Development of a Fan-beam Optical Computed Tomography
Scanner for Three-Dimensional Dosimetry

by

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BSc, Thompson Rivers University, 2007

A Thesis Submitted in Partial Fulfillment of the Requirements for the Degree of

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in the Department of Physics and Astronomy

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Abstract

The current state of a prototype fan-beam optical computed tomography scanner for three-dimensional radiation dosimetry has been presented. The system uses a helium-neon laser and a line-generating lens for fan-beam creation. Five photodiode arrays form an approximate arc detector array of 320-elements. Two options of physical collimators provide two levels of scatter-rejection: single-slot (SS) and multi-hole (MH). A pair of linear polarizers has been introduced as a means of light intensity modulation. This work examined: (i) the characterization of system components, (ii) data acquisition & imaging protocols, and (iii) the scanning of an nPAG dosimeter. (i): The polarizer-pair method of light intensity modulation has been calibrated and the polarization sensitivity of the detector array was evaluated. The relationship between detected values and both light intensity and photodiode integration time was examined. This examination indicated the need for an offset correction to treat all data acquired by the system. Data corruption near the edges of each photodiode
array was found to cause ring artefacts in image reconstructions. Two methods of extending the dynamic range of the system—via integration time and light intensity—were presented. The use of master absorbent solutions and spectrophotometric data allowed for the preparation of absorption-based and scatter-based samples of known opacities. This ability allowed for the evaluation of the relative scatter-rejection capabilities of the system’s two collimators. The MH collimator accurately measured highly-attenuating solutions of both absorption-based and scatter-based agents. The SS collimator experienced some contamination by scattered light with absorption-based agents, and significant contamination with scatter-based agents. Also, using the SS collimator, a ‘spiking’ artefact was observed in highly-attenuating samples of both solution types. (ii): A change in imaging protocol has been described that greatly reduces ring artefacts that plagued the system previously. Scanning parameters related to the reference scan \( (I_o) \) and data acquisition were evaluated with respect to image noise. Variations in flask imperfections were found to be a significant source of noise. (iii): An nPAG dosimeter was prepared, planned for, irradiated, and imaged using the fan-beam system. In addition to ring artefacts caused by data-corruption, refractive inhomogeneities and particulates in the gelatin were found to cause errors in image reconstructions. Otherwise, contour and percent depth dose comparisons between measured and expected values showed good agreement. Findings have indicated that significant imaging gains may be achieved by performing pre-irradiation and post-irradiation scans of dosimeters.
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As a graduate student, I do occasionally get brief moments to sit back and think about things other than research. Although, life and research often find ways to mingle with one another. It’s just an occupational hazard, I guess. While I was working through this thesis, the symbolism of beam attenuation was not lost on me. A flux of photons being sent through a medium, potentially being thrown off their path by any interaction at any point along the way. I think of those photons as having a starting point and a destination. If I were them, I would want to reach my destination. Yes, I know the speed of light. Yet, in my mind, this imagery plays slow enough to feel their tension and their worry. The tension of trying to reach a destination, and the worry of being thrown off path. I’ve felt this tension and I’ve had this worry. I’ve also had destinations, and I’ve changed destinations. Along the way, any number of interactions have led me to my current position. When I graduated high school, I could not have predicted that I would reach my current position. I cannot know my destination. But, wherever I am currently, I am sure it is on my path. I have a number of people to thank for their role in getting me here.

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Chapter 1

Introduction

In its early years, Radiology—the use of radiation for the diagnosis and treatment of disease—was referred to as Roentgenology. This branding was in reference to Wilhelm Conrad Röntgen, who had discovered X-rays somewhat accidentally in November of 1895 [1]. Scientists were swift to employ these new kinds of rays in creative ways. Doctors found means to study anatomy like never before [2]. Surgeons were able to locate bullets and plan for their extraction [3]. Quickly, the use of radiation for the treatment of cancer would emerge. In 1901, upon witnessing the “remarkable improvement” Roentgen rays had produced in a case of breast cancer, Dr. Andrew Clark felt that these observations seemed “worthy of putting on record” [4]. A flood of interest in the therapeutic use of radiation would soon follow.

In the summer of 1910, at the 78th annual meeting of the British Medical Association, physicist Sir J. J. Thomson borrowed from his own experience with Roentgen radiation to comment on its therapeutic applications [5]. His remarks would embody concerns that are still the essence of modern radiation therapy: “When using these hard rays practically the whole body is exposed to the same influence, and to a layman this procedure looks rather like flogging the whole school because one boy has committed a fault. Parts of the body other than the one concerned are irradiated.” A century later, the goal is still the same: to treat the disease while avoiding undue harm to
healthy tissues. As radiation therapy has evolved, the methods of achieving this goal have grown more and more complex.

Today, state-of-the-art radiotherapy techniques strive to deliver highly conformal dose distributions. While ambitious, these state-of-the-art techniques demand appropriate methods of dose verification. For one, highly modulated dose depositions inherently become more sensitive to positioning errors and patient motion, especially when situated near critical anatomical structures. Furthermore, if one is to diligently champion new advances in radiation delivery, one must capably demonstrate that better dose distributions can be realized. Thus, developments in radiation delivery must be accompanied by corresponding developments in radiation dosimetry. The following work is intended to contribute to developments in dosimetry.

1.1 Radiation Therapy

The following subsections outline key factors involved in classical and modern external beam radiation therapy. These reviews aim to provide a sense of how treatment planning was initially approached, as well as a sense of the motivation behind emergent treatment technologies. This brief overview is meant to emphasize just how complex modern radiation therapy has become.

It must be noted that brachytherapy techniques using sealed radioactive isotope sources—such as iodine-125 and iridium-192—also fill a vital role in the treatment of cancers. These radioactive sources are positioned internally within the patient to deposit dose locally. The following discussion will examine only teletherapy techniques—treatments performed using an external beam of radiation.

1.1.1 Basics of Radiotherapy

The therapeutic use of radiation garners a biological response to dose. Dose—defined as the amount of energy deposited in a given mass—has units of J/kg or Gray (Gy). A main challenge in radiation therapy (RT) involves the accurate delivery of dose and
the disconnect between the dose intended and the actual dose delivered. There are no means of confirming that the dose prescribed was in fact realized within the patient. A subset of measurements may be taken during treatment, and those measurements can be used to infer what was most likely delivered to the patient. Yet, one can never be absolutely certain. Therefore, radiation oncologists and physicists must rely heavily on the decades of work that have gone into understanding the behaviour of radiation.

The simplest dose distribution example in teletherapy radiation would involve a single, square-shaped beam of photons. Even so, there are a number of parameters that will affect how such a beam will deposit dose. Beam size, beam energy, source size, the distance between the source and the patient, and whether the source is from a linear accelerator or a $^{60}\text{Co}$ unit. These will all affect the beam that emerges from the head of a treatment unit. Once this beam confronts the patient, anatomical factors become involved. These include the contour of the patient and the heterogeneous tissues that make up the patient. A given beam of photons will deposit dose at different rates as it traverses fat, muscle, bone, or lung.

![Figure 1.1: Classical radiation therapy techniques: (a) physical compensators for fluence modulation, (b) field aperture shaping, and (c) multiple fields for dose sparing of healthy tissues.](image)

Understanding the factors involved in dose deposition allows treatment planners to modify their approach. Photons of higher energies can be used to target deeper-seated
tumours. Physical compensators can be used to accommodate irregular surfaces of the patient (see Fig. 1.1 (a)). The aperture of the beam can be collimated to match the shape of the tumour to avoid critical structures (see Fig. 1.1 (b)). Multiple beam angles can be integrated so that their combined doses culminate upon the treatment volume while their entry and exit doses remain relatively low (see Fig. 1.1 (c)).

The approaches discussed so far are performed using static setups. Once the equipment is in place, the radiation source is activated for a calculated length of time. Advances in mechanical fidelity and computational power would allow newer techniques to be pioneered.

1.1.2 Advanced Radiotherapy

Recent developments in radiation therapy have sought to apply the basics of radiotherapy, but in more sophisticated and individualized ways. Fluence maps can be modulated dynamically. Positioning errors can be minimized by ensuring correct field placement during treatment. Every plan is tailored to the individual patient to effect a better outcome. The following section describes a few prominent methodologies in radiotherapy today. It is by no means a comprehensive survey.

Intensity Modulated Radiation Therapy (IMRT)

A physical compensator (see Fig. 1.1(a)) is one method to modulate the intensity of a 2D beam fluence. Thicker layers of material (e.g. lead or tungsten) are used to attenuate the beam, thereby decreasing the flux that reaches the patient [6]. However, because lower energy photons are more likely to be attenuated, the spectrum of photons that pass through the compensator is of higher mean energy, or “harder.” Thus, in this fashion, one cannot simply modulate the number of photons in a given region of the field without affecting the response behaviour of those photons as well. This method is also very labour intensive when unique physical compensators are used for each patient.
In order to modulate the 2D map of a beam without tampering with its spectrum, one can employ dynamic or static electronic compensation [7]. With dynamic compensation, a multi-leaf collimator (MLC) is used, which features multiple moveable leaves of highly-attenuating material (typically tungsten). Using an MLC, one can either simply shape the field to delineate the target (as illustrated in Fig. 1.1(b)), or one can modulate the intensity across the field by using computer-controlled motion of the leaves while the beam is on. In this manner, regions of higher fluence are achieved by leaving these regions open for longer durations while the beam is activated. Areas where lower fluence is desired are achieved by blocking these areas for the duration of the exposure. With static compensation, two fields of different shapes are delivered subsequently to superimpose their effects.

Using this method of fluence modulation, elaborate field maps are achievable. This technique is often demonstrated by calculating an x-ray fluence that creates a familiar image (see Fig. 1.2(a)). More practically, this technique can allow for the blocking of areas in the centre of the beam, a task that effectively corresponds to a “floating block” compensator (see Fig. 1.2(b)). One disadvantage to modulating a beam in this fashion is that it requires longer treatment times.

![Figure 1.2](image_url)

**Figure 1.2:** Beam’s eye view of two radiation fields created using dynamic MLC collimation. In (a), a recognizable image is used to exhibit the modulation achievable with this method. In (b), a floating block is created in the middle of the beam. Note that darker regions correspond to higher fluence.
The term Intensity Modulated Radiation Therapy (IMRT) conventionally applies to treatment plans that are delivered using MLC compensation and are calculated by inverse-planning [8]. That is, once the target volume is defined, an optimization algorithm uses data about the target to calculate the proper MLC-motion routine to achieve the best dose distribution.

**Volumetric Modulated Arc Therapy (VMAT)**

Treatment plans designed using IMRT techniques typically employ 5–10 fields to achieve desirable results. Volumetric Modulated Arc Therapy (VMAT) goes one step further and performs IMRT in a continuous arc [9]. Using an inverse-planning approach similar to IMRT, one is able to calculate the appropriate MLC-motion routine that is to be executed while the treatment gantry simultaneously rotates about the patient. This results in entry doses that are lower and more evenly distributed, and high-dose volumes that are more localized to the tumour site. An added benefit to implementing such a technique is a significant reduction in both beam-on time & treatment time [9].

*Figure 1.3:* Schematic of an arc treatment using Volumetric Modulated Arc Therapy. Beam fluences are dynamically modulated while the treatment gantry is rotated. Although only a partial arc is illustrated here, a full arc or multiple arcs may be used to achieve better dose distributions.
Image Guided Radiation Therapy (IGRT)

As seen with IMRT and VMAT, methods are emerging that are capable of formulating complex dose distributions. These complex distributions provide the opportunity for: (i) doses to be escalated to improve tumour kill, and (ii) treatment to be attempted in sensitive regions where radiation would have previously been considered too risky. While these advances in treatment planning are encouraging, they force a heavy reliance on accurate patient alignment. Treatment plans are being designed to conform dose closely to the tumour volume. As such, misalignment can result in not properly treating the disease, and/or highly irradiating an organ at risk (e.g. parotid glands, rectum, or spinal cord), which can result in debilitating complications (e.g. inability to produce saliva, rectal ulcerations, or paralysis).

![Diagram showing spinal cord and treatment volume](image)

**Figure 1.4:** The compelling need for patient-setup verification. In order to safely take advantage of highly-modulated dose distributions, accurate alignment is essential. Here, a 5 mm lateral shift causes a threat to the spinal cord.

Image Guided Radiation Therapy (IGRT) uses onboard imaging equipment to validate patient alignment and modify the treatment plan if deemed necessary [10]. Yet, IGRT methods can be expanded to go beyond just patient setup. Using real-time imaging data, adaptive treatment methods could potentially modify dose delivery on-the-fly [11]. Methods that acquire real-time data are necessary in order to develop treatment plans in 4 dimensions—3-dimensional dose distributions that fluctuate over time [12]. Cases that exemplify the need for 4D treatment plans are tumour sites in
the upper torso, which move due to breathing motion.

1.2 Point & 2D Dosimetry

A wealth of research has gone into understanding how photons interact in a medium. Specifically in medical physics, dosimetrists are interested in how photons and charged particles traverse water, as the soft tissues of the human body are considered near water-equivalent. Using dosimeters in water tanks or water-equivalent phantom materials, dose measurements have been obtained for a large array of different beam geometries and beam energies. So much is understood about the relationships between materials, photon energy, and interaction coefficients that treatment planning system algorithms are sufficient enough for calculating the expected dose for simple treatment plans [13].

There is a limit to how much credence can be lent to these algorithms, however. In some instances, measurements are becoming necessary. For more complex cases, such as IMRT, manual checks of the doses calculated by commercial treatment planning systems would prove difficult and time-consuming [14]. The reliability of mechanical systems adds another factor, which introduces more opportunity for inaccuracy. As a result, measurements are becoming necessary to validate demanding treatment plans. The following is a survey of dosimetry options that are being used clinically.

1.2.1 Point Measurements

Ion Chambers

Capable of precise radiation measurements, ionization chambers are routinely used to assure consistently accurate output from high-energy radiation treatment units. The ionization chamber is the most common dosimeter in use in radiation therapy departments, in part due to its recommended use set forth by Task Group 51 (TG-51) of the American Association for Physicists in Medicine (AAPM) [15].

The ionization chamber contains a mass of gas, $m$, between two electrodes. As
the gas is exposed to ionizing radiation, charges are freed in the gaseous volume. By applying a known voltage across the electrodes, precise measurements of the dose to the gas are calculable by considering the charge collected, \( Q \), and the average work required to free a single electron from the gas, \( \frac{W}{e} \):

\[
D_g = \frac{Q}{m} \left( \frac{W}{e} \right)_g
\] (1.1)

Knowing the absolute dose delivered to the gas allows the calculation of the absolute dose that would have been delivered to the volume of water that the ion chamber displaces. This is done through a number of correction factors described by TG-51, which need not be discussed here [15].

Ion chambers could be used to acquire 3D dose distributions, albeit onerously. To do so, the ion chamber is scanned step-by-step throughout the volume being examined. Typically the chamber is waterproofed so that it may scan a volume of water. This can take a significant amount of time, and the radiation source must be “on” for data collection at each point in the scan. Additionally, the volume of the active region can blur the dose distribution being scanned, which limits the resolution achievable by an ion chamber scan. Obtaining 3D dose distributions using an ion chamber would not be practical to say the least.

**Thermoluminescent Dosimeters**

A thermoluminescent dosimeter (TLD) is a crystal, typically calcium fluoride (CaF\(_2\)) or lithium fluoride (LiF), that contains ‘traps’ in higher energy bands where electrons excited by radiation are detained for long periods of time. These traps hold electrons in their excited states until the crystal is heated, which allows electrons to return to the ground state while emitting visible photons. The amount of light emitted during a precise heating routine provides an indication of radiation exposure. TLDs only provide one opportunity to measure their absorbed dose, as the heating process
essentially erases the dosimeter.

TLDs are available in many forms—granular, chip, pellet, disc, plate, or powder enclosed in transparent plastic tubing [16]. To be used dosimetrically, a TLD’s luminescence-response must be calibrated. They can be calibrated in batches, or individually for increased accuracy. A benefit to using TLDs is that they do not require a power source. TLDs are commonly used for point-measurements during treatment and for personnel monitoring [16]. Their implementation beyond point-measurements is uncommon because the process of reading numerous TLDs for a single dose distribution becomes cumbersome.

**Diodes**

Semiconducting materials, such as silicon (Si) and germanium (Ge), can also be used for radiation measurement. When assembled in a p-n junction, charges released in the material are collected. Measurement of this charge provides a proportional measurement of the energy deposited in the semiconductor. While charge collection using diodes is similar to measurements using ion chambers, the higher densities and Z-values of semiconductors make diodes much more sensitive to radiation. For x-ray energies $>100$ keV, a diode may be used as a substitute for an ion chamber [16]. In cases where resolution is important, such as small-field dosimetry and penumbra measurements, diodes are often preferred over ion chambers because of their small size, although they still require a power source and water-proofing to be used analogously.

**1.2.2 Planar (2D) Measurements**

**Films**

Traditional photographic film uses a layer of silver bromide (AgBr) granules, typically 1–2 micron in diameter, to measure the spatial distribution of radiation [17]. When a granule is exposed to ionizing radiation, charge pairs begin converting $\text{Ag}^+$ ions into Ag atoms. Once a few silver atoms are converted on a given granule, it can be
chemically developed. During development, the bromine in the granule is removed, leaving behind a granule consisting entirely of silver. Unexposed granules are washed from the film.

Areas of the film treated to higher exposures retain more granules of Ag. Consequently, these areas are more opaque. In order to quantify the exposure received at a given point on the film, one considers the optical density (OD)—a value obtained by comparing incident light, $I_o$, and light transmitted through the film, $I$:

$$OD = \log_{10} \left( \frac{I_o}{I} \right)$$  

Radiochromic films—films based on dyes rather than AgBr—were later developed [18]. These new types of film are near tissue-equivalent and do not require chemical development [19]. In order to be used dosimetrically, both types of films rely on calibration curves of OD vs dose. A dose map can thereby be obtained from a map of optical density. In portal imaging, an imaging unit is placed distal to the patient for setup verification purposes [17]. However, the time required to use films for such a purpose does not allow for practical pre-treatment setup verification. Portal imaging with film provides data post-treatment, which allows for subsequent treatments to be modified to compensate for treatment errors.

**Electronic Portal Imaging Devices**

If a portal image is to be used to ensure accurate patient alignment on a day-to-day basis, it must be immediately available. This is possible with electronic portal imaging devices (EPIDs). Relative dose maps are obtained with an array of electronic dosimeters, each acting as a single pixel. These dosimeters can be liquid ion chambers, semiconductors, or the more common amorphous silicon (a-Si) technology [13]. Using photodiodes mounted upon an a-Si panel, a scintillator allows the indirect detection of x-rays by first converting them into visible light. These devices have been shown
to be suitable for clinical dosimetry and are currently the most widely available type of EPID [20, 21].

Both films and EPIDs are valuable clinical tools for providing relative 2D dose distributions with good spatial resolution. For treatment plans that leave the treatment head in a static position during irradiation (e.g. IMRT), 2D measurements will provide good, quantifiable data that may be used to validate that the intended fluence is achieved. However, films are not ideal for VMAT treatment delivery schemes for example, due to the fact that films are integrating dosimeters and VMAT treatment verification using 2D dosimetry would require time-resolved measurements. Such measurements may be obtained, for example, using EPIDs [22]. However, using 2D measurements to reconstruct an expected 3-dimensional dose distribution does not verify that said 3D dose distribution can be achieved. Rather, it only verifies that the treatment unit is able to perform as instructed. To truly determine that a 3D dose distribution may be realized, one needs a 3D dosimeter.

1.3 3D Dosimetry

This section will examine measurement-based methods of 3D dosimetry. These sit apart from computational methods, most notable of which are Monte Carlo techniques. While Monte Carlo remains a valuable tool in radiation therapy, it ultimately insists that a set of treatment parameters are assumed in order to simulate dose delivery. Consequently, this means that Monte Carlo methods provide information about the dose expected. Measurement-based methods provide information on the dose realized.

A key requirement of a 3D dosimeter is that it should be water-equivalent, which is not required for point and field dosimeters. For 1D & 2D dosimeters, measurements are taken at a single point or cross-section of the beam. They represent a single instance as the beam traverses the material. Yet, because most of these dosimeters
are not water-equivalent, they cannot simply be arranged to fill a volume that is to be examined. Their very presence perturbs the behaviour of the beam. Even if one were to use a near tissue-equivalent radiochromic film, stacking numerous films would not be practical because the unavoidable air gaps between films can cause an angular-dependent response to dose [23]. What is necessary is a volumetric dosimeter that is water-equivalent, responds in a measurable way to radiation, and accumulates the dose to which it is exposed.

All water-equivalent 3D dosimeters that have been developed to date are chemical dosimeters. That is, they rely on chemical reactions, which occur due to radiolysis—the breakage of chemical bonds by radiation. Understanding the fundamental reactions that occur in these dosimeters is important, both for quantitatively evaluating their changes and modifying their performances. As a result, 3D dosimetry has become a collaborative area of research, involving physicists, chemists, and clinicians alike.

1.3.1 The Fricke Dosimeter

Fundamentally, the Fricke dosimeter is a simple solution of ferrous sulfate (FeSO$_4$). When irradiated in the range of 40–400 $Gy$, ferrous ions are converted to ferric ions ($Fe^{2+} \rightarrow Fe^{3+}$) linearly with dose [16]. Originally, solutions could be evaluated by chemical titration or by absorption spectroscopy. Whether evaluating the change in molarity ($\Delta M$) or the change in optical density ($\Delta OD$), the response is linear with dose. However, when used as a solution, energy deposited in the Fricke dosimeter does not maintain its spatial distributions. Ferric ions are free to move throughout the solution. Therefore, a single Fricke solution dosimeter provides a single measurement for its entire volume.

In order to use ferrous sulfate dosimeter for 3-dimensional measurements, the solution must be set into a stable matrix. The concept of a gelatin-fixed Fricke dosimeter (Fricke-gel) was first introduced by Gore et al in 1984, when they imple-
mented nuclear magnetic resonance (NMR) for measurement of radiation exposure [24]. This measurement is based upon the differences in spin relaxation times for Fe$^{2+}$ and Fe$^{3+}$. Unfortunately, the diffusion of ferric ions over time within the gel matrix inevitably degrades the spatial distribution of the dose deposition, which places time constraints on how soon a Fricke-gel dosimeter must be evaluated after irradiation [25]. Even with specially designed recipes, NMR measurement is required within 2 hours of exposure [26].

1.3.2 Polymer Gels

Dosimeters based on polymer gels consist of a solution of monomers that is spatially stabilized within a gelatin matrix. The basis of their reaction is the polymerization of these monomers due to radiation exposure. Polymerizations result in precipitates being deposited locally within the gel. The first demonstration of this was by Maryanski et al in 1993 using nuclear magnetic resonance (NMR) to evaluate changes to the dosimeter [27]. Although it was initially intended to serve solely as the matrix for the dosimeter, gelatin unavoidably participates in the chemical reactions of a gel dosimeter [28]. The first polymer to be used with the potential for radiation dosimetry was poly(methyl methacrylate), investigated by Alexander et al in 1954 [29]. However, with poly (methyl methacrylate), radiation breaks down the polymer instead of producing it. Today, two key polymers provide the basis for gel dosimetry research—poly(methacrylic acid) and polyacrylamide [28].

Gel dosimeters that use poly(methacrylic acid) are referred to as MAG dosimeters, in short. These dosimeters contain a single monomer, which polymerizes when exposed to radiation. To avoid inhibition of this polymerization, it is important that the dosimeter be void of oxygen. This can been done by manufacturing the gel in an environment devoid of oxygen, or by using an antioxidant [30, 31]. Gels that use an oxygen scavenger are manufactured in normoxic conditions. Thus, acronyms for these dosimeters are given an ‘n’ prefix (i.e. nMAG). MAG dosimeters have been
shown to be sensitive to dose-rate and temperature during irradiation [28].

Polyacrylamide gel dosimeters are denoted as PAG and nPAG dosimeters (oxygen also inhibits their polymerization). These dosimeters are based upon the copolymerization of acrylamide (or an acrylamide analogue) and a crosslinker, in this case N,N’-methylenebisacrylamide (bisacrylamide for short). Acrylamide is a toxic substance, so added care must be taken in the manufacture of a PAG dosimeter [32]. As a result of its toxicity, efforts have been made to seek out a safer replacement for acrylamide [33]. Unlike MAG dosimeters, PAG dosimeters show no sensitivity to temperature during irradiation, and have little dependence on dose-rate [28, 31].

A variety of methods have been implemented to quantitatively determine the distributions of precipitates in irradiated polymer gel dosimeters. Changes in NMR relaxation rates have been exploited for magnetic resonance imaging (MRI) [27]. Visual changes in opacity (by means of light scattering) have led to the use of optical computed tomography (CT) for scanning of polymer gel dosimeters [34–38]. A number of different scanning designs have been explored, which will be examined in more detail in Chapter 2. Density changes in polymer gels can be evaluated using x-ray computed tomography [39]. Also, ultrasound has been considered [40]. Initially, this modality saw little development, but has recently received a spur of interest that may revitalize its potential [41, 42].

1.3.3 Dye-based Dosimeters

The first dosimeter to exhibit a colour response to dose was tested in 1950 by Day and Stein using radiosensitive dyes in gels [43]. Subsequently, another gel based on a chloral hydrate solution (Cl₃CCH(OH)₂) was formulated by Andrews et al in 1957 [44]. With chloral hydrate, exposure to ionizing radiation results in the production of hydrochloric acid (HCl). To observe a dye-based response, a pH indicator can be added to the gel recipe. Electrical conductivity was also used to measure the HCl content of samples extracted from the gel. Susceptible to the same ion diffu-
sion as Fricke-gel dosimeters, measurements were taken immediately post-irradiation. Dosimeters based upon chloral hydrate were not pursued any further, possibly due to higher interest in the ferrous sulfate-based Fricke dosimeter.

Later, a dye-based polyurethane dosimeter was developed [45, 46]. Adamovics’ PRESAGE™ dosimeter is an optically-based 3D dosimeter that relies on the absorption of light rather than scatter. The active agents of the plastic are radiochromic dyes and free radical initiators. The PRESAGE™ dosimeter is slightly less water-equivalent than polymer gels, having an effective-Z value 16.5% higher and electron density to mass ratio ($\rho_e/\rho$) 1.8% lower than water [47]. Brown et al found that, for the therapeutic range (1–20 MeV), the effective radiological differences between PRESAGE™ and water were less than 5%, calculated using Monte Carlo simulations [47]. They also proposed that these differences might be overcome through the use of a dosimetric correction factor.

Due to the fact that the PRESAGE™ dosimeter was designed specifically as an optical dosimeter, the only means of evaluating it is by optical CT scanning. This dosimeter has quickly gained popularity, motivating a considerable amount of research into its characterization and clinical potential [47–55].

1.3.4 Gel Dosimetry

As this work most specifically addresses the needs of polymer gel dosimeters, a brief overview of the gel dosimetry workflow is warranted. The method allows for a wide variety as far as choice of gel recipe, evaluation modality and irradiation technique are concerned. Nevertheless, the general routine remains fairly consistent.

The first step in gel dosimetry is choosing a recipe that is appropriate for the modality that will be used for its analysis. By purposefully preparing the gel in this manner, one promotes optimum readout of dose information. Next, a treatment plan is devised to use for the irradiation of the dosimeter. This plan represents the dose intended, against which dosimeter readings will be compared. Next, the dosimeter is
irradiated according to the plan. Lastly, irradiated gels are imaged; typically, this is done by either MRI, x-ray CT, or optical CT.

\[\text{Figure 1.5: Schematic of the gel dosimetry workflow routine: (a) fabrication, (b) treatment planning, (c) irradiation, and (d) evaluation.}\]

With regards to evaluating the dosimeter post-irradiation, MRI was the first modality implemented for this purpose \[27\]. This modality benefits from a substantial library of research that has gone into understanding the technique and optimizing its results. However, limited availability and high cost significantly hampers the use of MRI for dosimetric purposes. Time with an MRI unit is valuable if one is available at all. In response to this reality, alternative methods are being explored.

Dosimetric information can be obtained using x-ray CT, and its wide-spread accessibility bodes well for its easy adoption if a proper dosimetry routine is realized. Yet, the density changes that are observed with x-ray CT are slight, so the resulting images are low in contrast and require multiple scans of each slice be averaged \[39\]. As well, beam-hardening of the x-ray spectrum can cause crippling imaging artefacts if proper container materials are not chosen. Lastly, while time with an x-ray CT is certainly less expensive than with an MRI unit, it is still a clinical piece of equipment.
that is primarily dedicated to the imaging of patients.

Optical CT is being explored to potentially establish an imaging system that is dedicated to 3D dosimetry and is relatively inexpensive. As this is a new imaging technology, it does not benefit from the wealth of efforts that have already been invested in MRI and x-ray CT. Currently, optical CT systems are being designed for short imaging times to encourage their adoption in the clinic. For the fastest systems to date, this has resulted in an adoption of absorption-based dosimeters. Scatter-based dosimeters present a challenge with regard to speed as their attenuation mechanism is problematic when more of the dosimeter is illuminated simultaneously. This issue will be discussed in detail in Chapter 2.

1.4 Thesis scope

This work examines an optical CT scanner that is currently in its development stages [38]. Its intended purpose is for the scanning of 3-dimensional dosimeters that exhibit increased optical density when exposed to radiation. Of these, polymer gels and PRESAGE™ dosimeters are the most viable for clinical use, as Fricke-gels still demand timely evaluation post-irradiation.

Former University of Victoria student David Rudko carried out the design, construction, and preliminary testing of the prototype scanner. Updates to the previous design and imaging protocols will be presented and discussed. A new circuit board and a set of daughter boards were developed at the University of Victoria, which provide profound noise-reduction, as well as significant increases in scan speeds over the first version. A pair of linear polarizers is introduced as a means of allowing in-alignment light intensity control. Imaging protocols that were used previously have been re-evaluated, resulting in the elimination of ring artefacts caused by reflection. In addition, investigations into the image reconstruction functions used previously have revealed deficiencies; alternative methods are presented.
New investigations have been performed to evaluate the current capabilities and deficiencies of the device. Detectors have been characterized using varying levels of light intensity. A method of extending the dynamic range of the system is introduced. Two collimators are evaluated using scatter-based and absorption-based solutions of known absorbances. Data acquisition parameters and their imaging ramifications are examined. A hardware-based issue has been recognized, which results in limited corruption of detector data. Finally, an nPAG dosimeter was manufactured, irradiated, and imaged.

Chapter 2 of this work provides an overview of optical computed tomography, with the specific aim being to give a sense of where this scanner’s design fits within the realm of optical-CT for 3D dosimetry. Its strengths and weaknesses relative to other designs will be outlined. Chapter 3 provides specific scanner and imaging protocol details regarding: components of the scanner, a fan-alignment routine, gel manufacture, gel irradiation, optically-absorbent solutions, and the steps used for data acquisition & image reconstruction. Chapter 4 includes the characterizations of a number of system components (most notably, the detectors and collimators), as well as a description of the method used for extending dynamic range. Chapter 5 is dedicated to imaging protocols and the effects of data acquisition parameters on scan results. Chapter 6 presents and evaluates the imaging results of a gel dosimeter. Finally, Chapter 7 summarizes the current capacity of the system and outlines directions for further development.
Chapter 2

Optical Computed Tomography

The most well-known implementation of computed tomography (CT) is the use of x-ray CT for medical imaging. By considering x-ray transmission values through a patient from many angles, one is able to reconstruct the patient’s 3D anatomy. For x-ray CT, this 3D data represents the physical densities of the patient’s tissues. However, the use of CT is not restricted to x-rays. Analogous to x-ray CT, optical CT uses visible light to evaluate the optical densities of an object. The following chapter reviews CT Theory, discusses issues in optical CT, surveys existing scanner designs, and introduces the scanner design that is the subject of this work.

2.1 CT Theory

2.1.1 Transmission & Optical Density

Beer’s Law allows one to obtain information about a specimen by observing light intensity as a beam of light (a ‘rayline’ or ‘ray’) traverses the specimen (see Fig. 2.1). As light travels through an opaque medium, it will be attenuated according to an exponential decay function. Beam attenuation is determined by the medium’s level of opacity, which is the product of linear attenuation coefficient, $\mu$ ($\text{cm}^{-1}$), and path length, $x$ (cm). This product is referred to as the optical density (OD) or absorbance (A). A higher OD means that a photon is more likely to interact with
The probability of a photon interacting increases as it travels deeper into a medium. By considering the input intensity of photons ($I_o$) and the output intensity ($I$), the transmittance value ($T = \frac{I}{I_o}$) represents the fraction of photons that have not interacted with the medium. Knowing this probability value allows the calculation of OD. For a homogenous medium, if the pathlength through the sample is known, one can calculate its linear attenuation coefficient. If a spectrum of light is used, the linear attenuation coefficient can be dependent on photon energy. If a monochromatic light source is used, all photons will have equal probabilities of interaction in a given medium.

\[ I = I_o 10^{OD} \]

\[ OD = \mu x = \log_{10} \frac{I_o}{I} \]

**Figure 2.1:** Optical density of a sample can be determined by comparing input and output light intensities. If the length of a homogenous sample is known, its linear attenuation coefficient, $\mu$, can be calculated.

For heterogenous samples, the OD for a given rayline is the sum of all optical densities along its path. See Figure 2.2. In this case, one cannot discern the different linear attenuation coefficient values along the rayline by only considering $I_o$ and $I$. One transmission value provides one OD value for the entire pathlength. In order to allocate different values to the various components of the specimen, one must acquire transmission values from multiple angles. This is the basis of computed tomography, and is discussed further in the next subsection.

For visible light, there are two main interaction mechanisms for opaque media: $i)$ absorption and $ii)$ scatter. With absorption, the entire energy of the photon is
absorbed within the medium. With scatter, the photon is elastically scattered away at a random angle. As illustrated in Figure 2.3, for samples with equal ODs, roughly the same number of photons will interact with each sample. If one is to accurately evaluate the opacity of a sample, only photons that have not interacted with the sample should be detected. Therefore, one would prefer to work with absorption-based attenuators, as they eliminate photons rather than deflect them. With scatter-based attenuators, multiple photon scatter can occur, potentially resulting in the contamination of the transmission signal. Through methods of scanner geometry, collimation, or data correction, the presence of scattered photons must be addressed when working with these types of attenuators.

\[ \text{OD} = \mu_1 x_1 + \mu_2 x_2 + \mu_3 x_3 + \mu_4 x_4 = \log_{10} \frac{I_0}{I} \]

**Figure 2.2:** When interrogating a heterogeneous sample, the resulting optical density is the sum of its component ODs. The linear attenuation coefficients for the various regions in the sample cannot be determined by comparing input and output light intensities.

**Figure 2.3:** Illustration of two samples of equal optical density but different interaction mechanisms: (a) absorption and (b) scatter. With absorption, photons are absorbed within the medium. With scatter, photons are deflected at a random angle. Working with scatter-based samples presents a challenge, as scattered photons can potentially contaminate the output signal.
2.1.2 The Sinogram & Image Reconstruction

The aim of computed tomography is to acquire a quantitative map for a slice of an object. The mathematics behind this task were first solved by Johann Radon in 1917 [56]. He showed that, if one could acquire an infinite number of projections through an object from an infinite number of acquisition angles, one can determine a 2D map of that object using the data acquired. Practically speaking, infinite acquisitions cannot be taken; but, an adequate number of measurements can suffice.

![Figure 2.4: A set of rays is used to evaluate an object at a single projection angle. By comparing input and output light intensities, an absorbance (OD) projection is attainable. In order to reconstruct a slice of the object, projections from numerous rotation angles must be acquired.](image)

![Figure 2.5: A test image (a) and its representation in sinogram space (b). Here, fan-beam projections through the image are taken as a simulated CT system rotates about the central pixel. In (b), vertical lines represent projections, and the horizontal axis represents the angular position from which each projection was taken.](image)

To reconstruct the optical density map for a slice of an object, numerous parallel raylines are used to interrogate the object. A full set of rays is referred to as a
projection or a view. By considering the transmission values of these raylines, an absorbance (A) profile is calculated (see Fig. 2.4). In order to reconstruct the slice being interrogated, absorbance profiles from numerous projection angles must be acquired. A collection of projections over multiple angles is referred to as a sinogram (see Fig. 2.5). Each pixel in the sinogram represents a single rayline. Sinograms can represent rayline data for \( I, I_0, T, \) and \( A \).

For image reconstruction, a sinogram of \( A \) data is needed. With this, slices may be reconstructed through a method referred to as filtered backprojection. As the name implies, absorbance values are projected back across the reconstruction field according to the geometry that was used for their acquisition. As multiple raylines from multiple acquisition angles are contributed, \( A \) data superimposes on itself. The result is a 2D map of the relative \( A \) values for the slice of the object that was examined. Prior to reconstruction, sinogram data is typically filtered with a convolution kernel (ergo, ‘filtered’). These filters are implemented in Fourier space (for computational efficiency; the calculation becomes a product rather than a convolution) and have various designs for various intended purposes. Although they will eventually require further consideration, reconstruction kernels are not examined in this work. Benefits that may be realized with proper filter choice are minor gains when compared to the issues that are being addressed at this point in time.

2.1.3 CT Geometries

To acquire 3-dimensional sets of data, one must acquire transmission values through the entire volume of the object. This can be done by acquiring multiple slices individually, or by using a broader beam which illuminates more of the object simultaneously. When deciding upon a scanner geometry, there are certain trade-offs to be considered. For optical CT, the main balancing act exists between accuracy and speed.

Consider a scanner design analogous to the first-generation x-ray CT: a translating
pencil-beam (see Fig. 2.6 (a)). A light source and detector are mounted on opposite sides of the object being scanned. In any given position, this system acquires data for a single rayline. In order to obtain raylines for a full projection, the source & detector system is translated in unison. Once an entire projection is acquired, the system is rotated relative to the object and acquisitions are repeated. To scan different slices, either the source & detector system or the object is shifted vertically. Such a design requires lengthy acquisition times. Yet, because this setup allows for the use of a physical collimator and only illuminates the specimen one ray at a time, it offers the best geometry to limit scatter contamination.

![Diagram](image)

**Figure 2.6:** A variety of geometries may be used to acquire transmission values through an object. In (a), a pencil-beam and detector are translated in unison across the object to obtain a full projection. In (b), a fan-beam can acquire a full projection for a single slice simultaneously. In (c) & (d), entire volumes of the specimen are irradiated simultaneously by parallel and cone beams, respectively. These allow projections through the specimen to be simultaneously acquired for multiple slices by area detectors. In (c), lenses are used to create parallel raylines. Once projections are acquired, the specimen is incrementally rotated and measurements are repeated.

Scan speeds may be increased by acquiring full projections of a slice simultaneously. Figure 2.6 (b) illustrates a fan-beam geometry. This has a set of raylines that spans the entire object. A consequence of illuminating multiple raylines at once is
Figure 2.7: Before a fan-beam sinogram is reconstructed, it is converted into a parallel-beam geometry. This involves re-binning each ray from the fan-beam, (a), to its corresponding parallel-beam projection, (b). Each ray in a given fan-beam matches individual rays in parallel beams at multiple angular positions. Here, 3 such matches are illustrated that the presence of stray photons from other raylines increases the probability of signal contamination. This is especially problematic with scatter-based attenuators. Fortunately, if only one slice is evaluated at a time, this provides space above and below the detector array to support a physical collimator for scatter-rejection. It should be noted that sinograms with parallel geometries can be interpolated from sinograms with fan geometries by re-binning ray values (see Fig. 2.7). It should also be noted that, according to the current literature, the scanner presented in this work is the only known attempt at a fan-beam system.

To further increase scan speeds, full projections for multiple slices may be acquired simultaneously. Consider parallel-beam and cone-beam geometries, as respectively shown in Figure 2.6 (c) & (d). Much of the object is illuminated at once, which leads to increased issues related to scattered light. The use of an area detector limits the ability to use a physical collimator. As a result, these systems choose to converge their beams towards the detector. This allows for the use of telecentric optical collimation to limit the amount of stray light that reaches the detector.
2.2 Optical CT

2.2.1 Issues in Optical CT

There are a number of issues that are common to both x-ray and optical CT. For example: streaking artefacts, ring artefacts and artefacts related to scatter are all realities for x-ray CT devices. However, there are some additional issues that are unique to optical CT. These include refraction-related issues such as rayline distortion, flask-edge effects, and schlieren. The following section surveys issues that must be acknowledged in optical CT.

Streaking Artefacts

If errors arise in the data of a sinogram, they will cause complications in the reconstructed image. Streaking artefacts are caused when raylines in a profile are not representative of the true absorbance value for the object at that position. When these errors are isolated to a single projection, the result is a single streak across the image corresponding to the angle of the projection and the position of the erroneous data. In some cases, these are caused by random errors that arise during acquisition. In other cases, errors may be caused due to challenging attributes of the object being scanned.

In optical CT, streaking artefacts often arise from: (i) refractive complications due to irregular surfaces of the object being scanned, (ii) misalignment of $I_o$ and $I$ data, or (iii) schlieren effects (discussed below). If a source of refractive error remains in a constant position relative to the image being scanned, the resulting streaks will align with the error source. For example, Figure 2.8 (a) shows a reconstructed image of a scatter-filled flask that had an air-bubble of $<1$ mm in diameter which remained on the surface of the flask for the full duration of the scan. In this case, the streaks are a result of the spherical surface of the bubble displacing raylines. Localized rayline displacement effectively creates a false opacity in the sinogram data. Minor
streaking artefacts may be removed from an image by filtering the sinogram data. Severe streaking artefacts can be impervious to filtering.

![Figure 2.8: Examples of (a) streaking and (b) ring artefacts.](image)

(a) ![Streaking Artefact](image)

(b) ![Ring Artefact](image)

**Figure 2.8:** Examples of (a) streaking and (b) ring artefacts. In (a), an air bubble of diameter less than a millimeter sticks to the flask wall, causing significant errors. Ring artefacts are apparent in both images, but more recognizable in (b). It should be noted that the outermost ring in each image is the cylindrical wall of the flask.

**Ring Artefacts**

In some cases, errors can remain in constant positions relative to the source & detector system for multiple projection angles. If these errors persist throughout the entire scan, they result in ring artefacts. Errors occurring only during segments of the scan result in segmented rings or arcs. These are most commonly a result of drifting detector response values, although ringing may be caused by a number of problems. For instance, because many dosimeters are cylindrical in shape, ring errors can be caused due to reflection. In these cases, the positions of the reflection interfaces remain relatively constant throughout the scan as the dosimeter is rotated. This allows for reflection-based errors to persist along relatively constant raylines.

An example of ring artefacts due to detector error is shown in Figure 2.8 (b). Post-processing methods to suppress ring artefacts may be performed. Ring artefacts may either be addressed in the sinogram space or in the reconstructed image space. In sinogram space, the data along the problematic rayline positions are corrected by
raising or lowering their intensity. In the reconstructed space, calibration images of uniform objects can provide correction maps which may be used to treat subsequent images. More sophisticated techniques may involve converting the reconstructed image into polar coordinates and analyzing data of common radial distances [57].

**Rayline Distortion**

Visible light refracts when it strikes the interface between two materials of differing refractive indices at an off-normal angle. This causes an issue in optical CT, as the majority of rays will reach off-normal interfaces during the scan. Thus, for all optical CT systems, a matching medium bath must be used. This is done by mixing water with a matching agent that has a refractive index greater than water. Figure 2.9 provides an illustration of how raylines are disturbed by unmatched refractive indices. If the matching medium and the dosimeter material do not share a common refractive index, radial compression of raylines results and raylines do not follow parallel paths. This causes radial compression of object attributes towards the center of the dosimeter in the final image.

![Figure 2.9: Illustration of rayline distortion created by a mismatched refractive medium. Ideal raylines are shown in green, unwanted raylines in red. With a mismatched medium, the bad rayline (i) traverses an off-angled path through the dosimeter, and (ii) compresses towards the centre of the dosimeter. Rays can also experience reflection at each of the interfaces encountered across the specimen. Here, the first reflected ray is illustrated.](image-url)
Flask Wall Effects

For dosimeters that are housed within a container, such as polymer gels, a materials quandary arises. Ideally, the entire dosimeter would consist of materials with a single refractive index. If this were so, the matching bath could be prepared to have a refractive index equal to both the gel and the container. However, with polymer gels, the material housing the gel often has a higher index. In reality, one is unable to match to both materials. Therefore, slight rayline compression is unavoidable.

Some researchers have shown that choosing a container material nearer to the refractive index of the gel increases the usable diameter of the dosimeter [58]. Unfortunately, these tests were performed on mock dosimeters—gels doped with scattering agents—using open containers which could not practically be used to house a polymer dosimeter due to the oxygen contamination that would result. Fortunately, the thickness of container walls are often relatively small. As a result, radial distortion is kept acceptably small (sub-millimeter) as long as the bath is matched with the dosimeter material [59].

Schlieren

When a matching bath is used, refractive inhomogeneities in the bath (a result of incomplete mixing of water with the matching agent) can cause light intensity and distortion errors during scanning. These errors result due to the redirection of light rays, which cause false opacities in the regions rays have been deflected from. These inhomogeneities are referred to as schlieren, from the German word ‘schliere’ meaning ‘streak.’ Schlieren may be caused by chemical or temperature inhomogeneities. An example of schlieren in an optical CT matching bath is shown in Figure 2.10. Errors due to schlieren can be minimized by allowing the bath to settle and equilibrate to room temperature.
Figure 2.10: Here, schlieren are observed in transmission scans through a matching bath. Refractive index inhomogeneities in the bath cause the displacement of light rays, resulting in false opacity values.

The Cupping Artefact

When scatter-based dosimeters are used, the possibility of scatter contamination increases as dosimeters increase in opacity. For highly attenuating scatterers, a cupping artefact results due to the overstatement of transmission values. This overstatement of transmission values correlates to an understatement of optical density values in the center of highly scattering regions. When contaminated data is reconstructed, regions of understated optical densities manifest as a depressed region, or ‘cup’ (see Fig. 2.11).

Scatter contamination can be avoided if a physical collimator is used. If a system’s geometry does not allow for the use of a physical collimator, scatter contamination may be reduced by lowering the dosage delivered to these types of dosimeters, thereby reducing scatter levels. This creates a difficult situation, as scatterers represent the very signal being examined. By lowering the dosages delivered to these types of dosimeters, quantum noise is increased and, correspondingly, the signal-to-noise (SNR) ratio is lowered. Ideally, scatter contamination would be addressed without
2.2.2 Current Optical CT Systems

Although the use of visible wavelengths of radiation introduces certain issues related to refraction, the easy manipulation of light rays allows for some creativity when it comes to designing an optical CT scanner. Not being limited to point sources, line and area sources can be easily realized. Mirrors may be used to efficiently redirect beams. As well, light sources that are used are typically coherent, thereby avoiding imaging issues related to beam-hardening (an issue in x-ray CT). The following section provides examples of some unique approaches that have been taken for optical CT scanner designs. For each design, their strengths and weaknesses are touched upon.

Pencil-beam (OCTOPUS™)

The first optical CT system used an identical geometry as the first generation x-ray CT system. Introduced by Gore et al in 1996, a translating pencil-beam was used to interrogate polymer gel dosimeters [34]. The greatest flaw to this design is scan speed—full volumetric scans can take many hours [60]. Nevertheless, it is

Figure 2.11: The cupping artefact. Consider a cylinder uniformly filled with a highly attenuating scatterer. Transmission profiles are shown in (a), with green indicating correct values and red indicating values contaminated with scattered light. Absorbance profiles, (b), show the resulting underestimation of absorbance due to signal contamination. When images are reconstructed, profiles through these reconstructions, (c), show the resulting ‘cupping’ which gives this artefact its name.
capable of providing accurate results using both absorption-based and scatter-based dosimeters [53, 61, 62]. A commercial version of this system is marketed under the name OCTOPUS™ (MGS Inc., Madison, CT, USA). A recent modification of this system provides ~5× reduction in scan time (8–9 minutes per slice), but its basic design ultimately limits the scan speeds that can be realized [63]. Some researchers consider results obtained with the OCTOPUS™ scanner as the ‘gold standard’ when evaluating the results obtained with an optical CT prototype [64]. For a schematic of this design, see Figure 2.12.

![Figure 2.12: Schematic of the pencil-beam (OCTOPUS™) scanner.](image)

Fast-scanning Laser

Some researchers have explored the possibility of fast-scanning pencil-beam systems. The first scanners of this type used rotating mirrors for rapid redirection of the beam [65–67]. In 2007, Krstajić and Doran presented a system that uses galvanometer-controlled mirrors, paraboloidal mirrors, and lenses [36]. See Figure 2.13 for an illustration of the schematic. The only moving components of the system, the galvanometer mirrors are rotated to reposition the entry point of the beam, each of them
dedicated to motion along an axis. A pair of lenses is used to ensure that beams (i) stay parallel within the tank, and (ii) converge upon a transmission detector after exiting the tank. Unfortunately, only initial characterizations have been performed with the device so far [68]. Still, if a well-performing version of this design can be realized, it could potentially provide results similar to the OCTOPUS\textsuperscript{TM} scanner in a fraction of the time.

Figure 2.13: Schematic of the fast-scanning pencil-beam scanner from Krstajić and Doran. Reference and transmission detectors simultaneously monitor the laser source. To collect rays for an entire 2D projection, galvanometer mirrors are used to deflect the beam position vertically (galvo 1) and horizontally (galvo 2). Two lenses, $L_1$ and $L_2$, are used to produce parallel beams and converge the beam towards the transmission detector, respectively.

### Cone-beam (Vista\textsuperscript{TM})

The first full-field optical CT design was introduced by Wolodzko et al in 1999 [35]. The cone-beam design uses a diffuse light panel as a source and observes a conical arrangement of raylines through the dosimeter using a CCD camera (see Fig. 2.14). A lens and aperture work together as telecentric collimation to limit the detection of stray light. Because the system obtains full 2D projections at once, scan speeds
are its strongest asset. However, this geometry is susceptible to scatter-related artefacts [58, 69].

Alongside the OCTOPUS™ scanner, this scanner is the only other commercially available system, marketed under the name Vista™ (Modus Medical Devices Inc., London, ON, Canada). While accurate scans can be obtained with the Vista™ scanner using absorption-based dosimeters, quality results using polymer gel dosimeters remain to be seen [60, 70]. Scatter corrections and scatter modelling have been investigated, but their specific implementation would require that assumptions be made about the OD distribution being evaluated [71, 72].

**Figure 2.14:** Schematic of the cone-beam (Vista™) scanner from Wolodzko et al. Raylines from a diffuse light panel converge through the scanning tank. An area detector acquires full-field views of the entire scanning tank.

**Broad Parallel-beam**

A full-field scanner using parallel raylines was presented by Krstajić and Doran in 2006 [37]. In order to ensure parallel geometry, a lens setup identical to their fast-scanning laser design *(described above)* is used. Yet, here a light-emitting diode (LED) source is used to illuminate the entire scanning medium. See Figure 2.15 for a top-down illustration of the system. Similar to the Vista™ scanner, this design uses a lens and aperture for telecentric collimation to limit the detection of stray light. A similar system was presented by Sakhalkar and Oldham which uses a diffuse light panel in replacement of the LED and first lens [64]. So far, publications with these designs have
used the absorption-based PRESAGE™ dosimeter [64, 73]. Nevertheless, due to the
fact that this broad field geometry simultaneously illuminates the entire sample (much
like the Vista™ scanner), one can expect these systems to be similarly disadvantaged
when evaluating polymer gel dosimeters.

![Figure 2.15: Schematic of the broad parallel-beam scanner from from Krstajić and Doran. An LED source simultaneously illuminates the entire medium. Lenses are used to maintain parallel raylines through the scanning tank. An area detector is used to obtain entire, full-field views of the light transmitted through the tank.]

2.3 Fan-beam Optical CT Scanner

The system design for the fan-beam optical CT scanner, which is the subject of this
work, finds itself in the middle ground with regards to speed. Its intended purpose
is to improve scan times while maintaining accuracy with scatter-based dosimeters.
With this design, entire projections are acquired at once, so it is certainly capable of
scans faster than the original translating pencil-beam design. Also, because only one
slice is illuminated at a time, this geometry allows for the use of a physical collimator.
On the other hand, by only illuminating one slice at a time, this system is unlikely
to surpass the scan speeds of systems which use broad beams that illuminate the full
dosimeter. Nevertheless, this design stands to potentially become the fastest system
capable of accurate measurements with polymer gel dosimeters. A schematic of the
fan-beam system is illustrated in Figure 2.16.

The original, prototype scanner was first investigated by a previous University of
Victoria student, David Rudko [38]. Preliminary tests revealed two key issues with the prototype system. The first issue was a substantial amount of detector noise, which was attributed to electronic issues with the circuit board that was used for data collection. The second issue was that reconstructed images suffered from significant ring artefacts. At this time, the cause of these artefacts was not determined.

**Figure 2.16:** Schematic of the fan-beam scanner which is the subject of this work. The intensity of a laser source is modulated by the cross-angle of a linear polarizer pair. A fan-beam is created with a line-generating lens. The dosimeter is immersed in a matching tank with walls concentric to the fan vertex. Five detector arrays provide a total of 320 detector elements to acquire full projections simultaneously. Detectors are concentrically mounted onto one of two collimator designs.

Most of the system components have remained unchanged since the original design, with a few key exceptions. First and foremost, a new circuit board and a set of 5 daughter boards were constructed to address the noise issues that were experienced with the prototype. In addition to addressing noise, the new board introduces a few new benefits. Another change to the system is a new method for controlling the intensity of the fan-beam. The previous design used neutral density filters (NDFs) of discrete optical densities for this purpose. The current work introduces a pair of linear polarizers to modify beam intensity. Lastly, a new entry window was installed in the matching tank. Due to moisture-related warping that had gradually occurred to the
old window, entry angles for portions of the beam were off-normal, causing vertical displacement of the resulting fan-beam. Chapter 3 provides a detailed introduction to the design features of the current system.

With regard to ring artefacts, the significant ringing that was observed in the previous system has been addressed. These were not solved by hardware considerations, but by changes in imaging protocol. The ringing that plagued the prototype was being caused by a combination of fan-beam inhomogeneities and reflection. The nature of this issue is discussed in more detail in Chapter 5, where imaging protocol investigations are described. Nevertheless, the removal of the predominant ring artefacts has revealed another set of ring artefacts. These distinct rings are not due to reflection, but rather are caused by data-corruption which occurs at the edges of the five detector arrays that make up the detector system. The nature of this data-corruption is examined in Chapter 4, where data acquisition is characterized.
Chapter 3

Materials & Methods

3.1 Scanner Design

For two photographs of the fan-beam system, see Figure 3.1. The list of components for the design are mentioned here, while more detail for each component will be provided subsequently. The system uses a 543 nm laser light source (Research Electro Optics, Inc.; Boulder, CO, USA) and a 60° line-generating lens (Edmund Optics; Barrington, NJ, USA) to create the optical fan-beam. A pair of linear polarizers (Edmund Optics) is used in the beam-line prior to the line-generating lens to allow for granular control of the beam’s light intensity. The matching tank (constructed at the Vancouver Island Centre of the British Columbia Cancer Agency; Victoria, BC, Canada) has walls concentric to the fan vertex so that rays enter and exit the matching medium at normal incidences. Dosimeter motion is achieved by a pair of stepping motors (one rotational, one translating vertically) from Newport (Irvine, CA, USA), which are mounted above the tank on a crossbar frame. Dosimeters are held in place with a mounting arm, which is suspended above the tank by the rotational stage. The rotational stage is raised and lowered by the vertically translating stage. After exiting the outer walls of the tank, the fan-beam passes through one of two choices of physical collimators (one a single-slot, the other with holes cut for each detector element). Five 64-element photodiode arrays (Hamamatsu; Hamamatsu City, Japan)
are mounted directly onto the collimator chosen. The collimators arrange the detector arrays so that they form an approximate arc of 320 elements that is concentric with the rest of the system. As they are linear arrays, a continuous arc is not achievable, so the center of each array is aligned to face the fan vertex. Each 64-element array is directly attached to its own daughter circuit-board, each of which is then attached to the mother circuit-board by ribbon cable (designed, constructed, and programmed at the University of Victoria; Victoria, BC, Canada). The mother board is connected by a Universal Serial Bus (USB) port to a personal computer (PC). MATLAB software (MathWorks; Natick, MA, USA) is used to orchestrate data collection by sending instructions to and receiving data from the mother board. MATLAB is also used to (i) communicate with a Universal Motion Controller (UMC) from Newport, which instructs the motion of the dosimeter stepping motors, and (ii) perform data-processing and image reconstruction. All of the components aside from the PC and UMC are arranged on a Newport optical bench.

**Figure 3.1:** System photographs for the fan-beam optical CT scanner. For a setup illustration of only the main components, see Figure 2.16.
3.2 Fan Creation

Fan creation for the scanner is achieved by three key components: a laser, a linear polarizer pair, and a line-generating lens.

The laser is a 543 nm helium-neon (HeNe) laser with a 2 mW maximum output. Its circular beam has a 0.83 mm diameter and 0.84 milliradian divergence. The laser is horizontally mounted roughly 18.5 cm high off of the optical bench using a set of Newport 423 series mounts and a Newport ULM-TILT mount. This setup allows the laser to be adjusted so that it is accurately aimed and horizontally level. The tuning of this alignment is detailed in section 3.8 below.

The pair of linear polarizers are used to modulate the beam intensity before it reaches the line-generating lens. Modulation is achieved according to Malus’ Law by adjusting the relative polarization angles of each polarizer. Polarization angles are adjusted using the rotational stages each polarizer is mounted in (1° increment markings). The characterization of this intensity adjustment method and a test of the polarization sensitivity of the detectors are provided in Section 4.1.

![Figure 3.2: Line-generating lens schematic. The angled surfaces of the knife-edge spread the circular input beam into an output fan-beam. Here, the virtual fan vertex is indicated within the lens.](image)

After the beam leaves the polarizer pair, it strikes the knife-edge of a 60° line-generating lens. This lens creates the fan-beam for the system. The fan is created
by the refraction that occurs as the circular laser beam strikes the angled surfaces of the lens. This glass lens has an index of refraction of 1.805, relative to $n_{\text{air}} = 1.000$. Each angular surface of the lens redirects the beam as illustrated in Figure 3.2, creating a fan with a height matching the input beam diameter and a fan vertex that resides within the lens. This vertex is virtual, and therefore cannot be used for alignment purposes. Thus, appropriate steps must be taken during beam alignment to ensure that the fan is properly set up. The line-generating lens is situated within a circular optical mount (Edmund Optics) which is attached to a rotational stage (Newport). These are attached to a Newport UP-1A mount and a pair of 423 series stages. This setup allows for the lens to be rotated and tilted, as well as shifted side-to-side and depth-wise relative to the tank. Again, the proper alignment of the fan will be described in Section 3.8.

3.3 Scanning Medium

The tank used to house the matching medium was constructed at the in-house machine shop of the BC Cancer Agency’s Vancouver Island Centre. A to-scale illustration of the tank is shown in Figure 3.3. The outer curved wall (made of acrylic plastic) is 3 mm thick with an inner radius of 228 mm and outer radius of 231 mm. The inner wall (made of polycarbonate plastic) is 3.15 mm thick with an inner radius of 31.75 mm thick and an outer radius of 34.90 mm. The flat wall of the tank (made of acrylic plastic) is 9.5 mm thick.

A portion of the inner curved wall was cut out and re-sealed with a thin entry window on its outer radius. Originally, this entry window consisted of a 1 mm thick mylar sheet that was mounted in place with polycarbonate braces and water-resistant caulking. However, this mylar window warped over time due to the absorption of water from the matching bath, which caused problematic refraction issues. The mylar window was replaced with a 1.8 mm thick piece of Lexan polycarbonate resin.
Figure 3.3: The scanning tank. Top: a to-scale illustration of tank wall dimensions. Bottom: photographs of the (a) old, warped mylar entry window, and (b) new Lexan entry window.
thermoplastic (SABIC; Riyadh, Saudi Arabia), which could be glued directly onto the polycarbonate wall of the tank. This material is typically used for safety face-shields and should therefore be less prone to warping.

The semicircular walls of the tank are constructed to be concentric. The fan vertex is to be positioned at the point of concentricity. In doing so, rays of the fan-beam will enter the entry window and exit the outer wall at normal incidence, thereby avoiding angular displacement due to refraction.

![Figure 3.4: Empirical bath matching technique. Using a combed fan, raylines through the bath (green) and raylines through the bath with dosimeter (red) are observed as refractive agents are added. Note: flask walls are positioned at pixels ∼45 and ∼285.](image)

The matching of the bath to the dosimeter being evaluated was achieved empirically. The precise refractive index of the gel dosimeters used have not been measured, so observed raylines were used to indicate that the dosimeter and bath shared matching refractive indices. To do this, a combed fan-beam was used. A comb pattern using 0.75 mm wide teeth was printed on a piece of overhead transparency. This comb pattern was placed on the inner radius of the inner wall, just prior to the fan-beam entering the tank. First, the comb pattern was observed through a water bath \( (n_{H2O} = 1.33) \). The positions of the teeth and gaps were determined by choosing a threshold detector value and assigning boolean values across the detector profile. These positions through water were used for rayline matching. Next, the dosimeter to be matched was placed in the beam’s path. The initial refractive mismatch results in radial compression of the raylines (see Fig. 3.4). To increase the refractive
index of the bath, glycerol \((n = 1.47)\) was gradually added until raylines through the dosimeter matched the original rayline positions that were viewed through water (see Fig. 3.4). The resulting composition consisted of 8.1% glycerol and 91.9% water, which corresponds to a refractive index of \(\sim 1.34\), only slightly greater than that of water.

### 3.4 Collimators

The system’s detector array is mounted onto one of the two available collimators; there are currently no means of mounting the detectors without a collimator. The first collimator is a simple, single-slot (SS) collimator with a slot height roughly matching the height of the detector elements \((0.8 \text{ mm}, \text{ no septa})\). The other collimator is a multi-hole (MH) collimator with telecentric holes cut for each individual detector element in order to reject stray light \((H: 0.8 \text{ mm}, W: 0.7 \text{ mm}, D: 15 \text{ mm}, \text{ pitch: } 0.8 \text{ mm})\). The collimators place the central elements of each detector array 260 mm from the fan vertex. This source to detector distance means that each of the 0.8 mm wide elements represents 0.179° of a 57.28° arc of detectors. Both collimators are constructed out of black plastic to absorb stray photons. In Section 4.5, comparative tests of these two collimators evaluates their abilities to reject scattered light.

### 3.5 Photodiode Arrays

The detector array consists of five 64-element Hamamatsu S8865 photodiode arrays \((H: 0.8 \text{ mm}, W: 0.7 \text{ mm}, D: 15 \text{ mm}, \text{ pitch: } 0.8 \text{ mm})\). The silicon sensors are operated in photoconductive mode (i.e. reverse biased), which causes their response to be linear with respect to light intensity. This model of sensors is also capable of fast acquisition speeds (maximum sampling rate: 500 kHz). The light-response of the entire detector array is examined in section 4.2. With an active width of 0.7 mm and a pitch of 0.8 mm, there is a 12.5% inactive area along the linear sensor. The data obtained for each 0.7 mm width is assumed to be representative of the inactive
area. Five 128-element arrays have also been acquired and their physical dimensions allow them to be placed in the same arc arrangement. However, they have not yet been tested and will not be evaluated in this work.

### 3.6 Circuit Board & Daughter Boards

A significant advancement over the previous generation in the current system came with the new circuit board and a set of daughter boards. In addition to addressing noise issues and increasing the speed of the system, the new circuit board comes equipped with 4 megabytes of onboard memory. Scans may be stored locally on the circuit board before they are output to the PC via USB. A serial port was also included on the new mother board, but has not been used as it has slower transfer speeds. A photograph of the mother circuit board & daughter boards setup is shown in Figure 3.5.

![Mother circuit board, set of daughter boards, photodiode arrays and a collimator.](image)

**Figure 3.5:** Mother circuit board, set of daughter boards, photodiode arrays and a collimator. Note: photograph shows a setup with the tank removed.

A number of choices were made with the new circuit board to reduce noise. Firstly,
5 individual daughter boards are used to place decoupling capacitors in close proximity with each detector array. These decoupling capacitors are used to reduce noise, and work best when connected close to the photodiode arrays. Another option would have been to create a mother board shaped to the detector arc so that detectors would mount directly onto the board. Choosing to use the daughter boards allows the decoupling capacitors to be mounted closely without the need for rigid mother board mounts for the detectors. Also, to reduce noise, a more layered circuit board was used (from 2-layer with the original board to 4-layer with the current board). This allowed for better segregation of circuitry. In addition, multiple smaller power sources were used on the board rather than one large, central power source. Lastly, as would be expected, electrical components with quieter specifications were purposely chosen.

To increase the speed of the system, a few changes to the original design were made. First and foremost, quicker analog-to-digital converters (ADCs) were chosen. This allowed for higher sampling rates. The new ADCs are capable of meeting the maximum clock rates specified by the photodiode arrays (maximum sampling rate: 1 MHz). Next, local memory improves the system’s speed potential. By storing scan data locally on the circuit board, the new system avoids the need to upload this data to the PC each time a scan is acquired. Scans may be performed at faster rates until the local memory is full, at which point all scan data may be transferred. This does not shorten the overall acquisition time as much as it allows for ‘quick-fire’ scans to be taken in short succession. Lastly, the USB connection between the circuit board and the PC means increased communication speeds over the serial port that was used previously. This delivers ‘scan’ instructions to the circuit board faster. Also, when full memory is reached, less time is required to transfer the data.

The firmware installed on the circuit board allows for scan parameters to be adjusted using commands sent through MATLAB. Integration times are programmable,
ranging from 640 µs to 21.890 ms in increments of 83.33 µs. The board is also capable of recognizing both the 64-element arrays and the 128-element arrays.

To instruct the circuit board to order a scan of the detector arrays, a single character (‘S’) is sent via USB. This scans the array twice, sending the data from the second scan to the local memory. The first scan is used to clear any charge that may have accumulated on the detector elements. At any time thereafter, the entire memory bank may be requested for upload to the PC by sending a single character (‘G’).

Acquisitions are performed using one of two routines. The first (monikered ‘Scan-Get’) is when data is uploaded to the PC immediately after it was acquired. This routine does not take advantage of the onboard memory. The second routine (monikered ‘Scan-Scan-Get’) lets scan data be stored to the circuit board memory for a number of scans before the entire set is uploaded to the PC. With this routine, the data set is cataloged as a single array of values which must be dissected and organized with MATLAB once it is collected. Using the Scan-Scan-Get routine, a maximum of 6,500 projection views may be stored to the local memory before it must be emptied to the PC. A full-memory transfer via USB takes ~100 seconds.

When the full array is scanned, each 64-element array is activated one at a time. Each of these arrays has a video output connected to an operational amplifier (op-amp) in a simple voltage follower circuit. The goal of this circuit is impedance isolation. The output of the op-amp circuit is sampled for data collection. While one array is being sampled, the other four are deactivated. In moving from one array to another, a switch occurs where one array is deactivated and another is activated. This sequencing of data collection proves to have consequences, which will be observed in Section 4.2.
3.7 Dosimeter Motion

Rotational and vertical motion of the dosimeter is performed with two stages. The rotational stage is mounted upon the vertical translating stage, which is suspended above the tank on a cross-bar frame (see Fig. 3.6). Instructions for their motion are sent using MATLAB from the PC to the UMC via a serial port. The vertical stage has a resolution of 1 \( \mu \text{m} \), maximum velocity of 2.5 \( \frac{\text{mm}}{s} \), and maximum acceleration of 10 \( \frac{\text{mm}}{s^2} \). The rotational stage has a resolution of 0.0005°, maximum velocity of 80 \( \frac{\text{deg}}{s} \), and maximum acceleration of 320 \( \frac{\text{deg}}{s^2} \). Considering speeds of motion as well as scan acquisition and data transfer speeds, a brief discussion on the potential scan times achievable with this system will be provided in Section 4.6.

Figure 3.6: Dosimeter motion setup. Dosimeter mounted in mounting arm, which is connected to the rotational stage, which is translated by the vertical stage.
3.8 Fan Alignment Routine

Proper alignment of the fan-beam is essential to allow for proper reconstruction of absorbance data. The beam must be horizontal so that the same slice is scanned throughout the full rotation of the dosimeter. As well, the vertex of the fan must be aligned to the system’s concentricity point. Both of these conditions need to be achieved at the detector array height so that fan data can be collected. As results obtained with the system can be sensitive to misalignments, a description of the fan alignment routine is warranted.

First and foremost, the laser should be allowed to warmup for at least an hour to allow the light source to stabilize before any acquisitions are attempted. A quantitative and qualitative examination of this warmup is given is Section 4.3. Next, with the line-generating lens removed, the laser is aligned to target the central pixel of the detector array. Once the laser is centered, alignment is performed with a high laser beam intensity, which saturates detectors, but allows the beam to be visible to the naked eye. Vertical and horizontal alignment are achieved by ensuring that the beam’s reflection and entry position match on the entry window. Next, the line-generating lens is mounted in place. The horizontal position of the fan is centered and aligned with the collimator slot. To ensure that the fan’s vertex resides at the point of concentricity, the flanks of the fan are used. These edges are mutually centered and aligned to target the proper vertex position. Next, the light intensity is lowered so that detectors are no longer saturating, and more fine-tuned adjustments are made to the tilt of the line-generating lens. Lastly, a dosimeter is mounted in the rotational stage and the position of the stage is adjusted so that its center of rotation lies halfway between the fan vertex and the central detector (130 mm). This can be done by physical measurement and verified by the projection size detected by the scanner. A 95 mm diameter flask centered at a distance of 130 mm creates a projection ~241
3.9 Gel Dosimeter & Absorbent Phantoms

All gel dosimeters and simulated phantoms being used with the current system are contained in 1L polyethylene terephthalate (PET) flasks (Modus Medical Devices Inc.; London, ON, Canada). These flasks are 95 mm in diameter and have ~1 mm thick container walls. An unfortunate attribute of these flasks are a pair of vertical seams along their sides. These seams can cause refraction errors in projections, which lead to streaking artefacts in image reconstructions that correspond to the seam positions. In the current work, these flasks have been used for: (i) an normoxic polyacrylamide gel dosimeter, and (ii) scatter-based and absorption-based solutions of known opacities. The following subsections describe the production of these materials.

3.9.1 Gel Dosimeter

For the gel dosimeter used in preliminary imaging, an nPAG recipe was chosen. This recipe consists of: (i) two co-monomers, (ii) the gelatin matrix, and (iii) an oxygen scavenger. Respectively, these are: (i) 20 g [2.0%] acrylamide and 20 g [2.0%] bisacrylamide, (ii) 920 mL [91.8%] deionized water and 40 g [4.0%] gelatin (from porcine skin), and (iii) 1.6 mL [0.2%] tetrakis(hydroxymethyl)phosphonium chloride (THPC). The percentages indicated are weight/weight percentages.

Of the water used, this was divided into 700 mL and 220 mL portions to aid in the fabrication of the gel. First, 700 mL of water was placed in a glass beaker and brought to 43 °C using a laboratory hot plate (Fisher Scientific International Inc.; Hampton, NH, USA). Once heated, the gelatin was added and mixed until dissolved using a magnetic stir bar. Then, the bisacrylamide was added and allowed to dissolve. Next, the acrylamide was added and dissolved. Once the solution was homogenous, the THPC and remaining water were mixed in a separate beaker. Immediately, this
A mixture was added to the homogenous solution. Quickly, the now-prepared nPAG solution was poured into a 1L flask and sealed. Teflon tape was used on the seal of the flask to encourage an air-tight closure between the lid and the container. This sealed flask was placed in a laboratory fridge and allowed to gelate before irradiation. The entire preparation of the nPAG dosimeter took roughly 3 hours, the most time-consuming step being the dissolution of the acrylamide. Steps for the treatment planning and irradiation of this dosimeter are detailed in Section 3.10.

### 3.9.2 Absorbent Phantoms

The first step for controlled imaging tests with uniform solutions of known opacities was the preparation of base concentrates that could be used to prepare solutions as needed. Two concentrates were prepared. One, an absorption-based concentrate, was a solution of blue food colouring (Club House; London, ON, Canada). The other, a scatter-based concentrate, was a solution of Duramax B-1000 polymer (Rohm and Haas; Philadelphia, PA, USA). In order to evaluate the known opacities for these concentrates, 9 dilutions of varying concentrations were prepared for each concentrate type (see Fig. 3.7). These dilutions were prepared using a laboratory scale by mixing portions of each concentrate with deionized water.

![Figure 3.7: Blue dye and Duramax B-1000 polymer. Master concentrates & dilution samples used for spectrophotometric analysis.](image)

To evaluate the opacities of the diluted samples, a Cary 500 UV-VIS-NIR Spectrophotometer was used (Varian, Inc.; Palo Alto, CA, USA). For evaluation by the spectrophotometer, samples were placed in cuvettes of 1 mm pathlength. For all 18 samples, absorbance spectra were obtained across the full visible range (300 nm – 900 nm) (see Fig. 3.8). Relevant to the current scanner, only the relationships of absorbance versus concentration (w/w%) at 534 nm are important, for these describe relationships that will be observed using the HeNe laser source. These relationships for each type of solution are shown in Figure 3.9. For the dye concentrate, the full range of dilutions maintains a linear relationship. For the Duramax concentrate, higher concentrations exhibit a slight sublinearity due to signal contamination by scatter. Therefore, to establish a relationship between concentration and absorbance, only the lowest six concentrations were used. By establishing this relationship for a 1 mm pathlength, expected values for longer pathlengths may be calculated. For instance, using the 1L flasks with a 93 mm central pathlength, expected absorbance values for a given concentration would be $A_{93\text{mm}} = 93 \times A_{1\text{mm}}$, assuming samples are uniform.

![Dye Spectrum](image1.png) ![Duramax Spectrum](image2.png)

**Figure 3.8:** Visible spectra for blue dye and Duramax B-1000 polymer.
3.10 Dosimeter Irradiation

As was described in Chapter 1, gel dosimetry consists of four main steps: (i) fabrication, (ii) treatment planning, (iii) irradiation, and (iv) evaluation. The previous section described the fabrication of the nPAG dosimeter. This section details the treatment planning and irradiation of the gel dosimeter.

The first step to treatment planning is to obtain a 3D rendering of the size, shape, and density of the object to be irradiated. A rendering of the dosimeter was obtained by x-ray CT. A 1L flask was filled with water, mounted in a horizontal arm, and scanned using a GE HiSpeed FX/i CT scanner (General Electric Medical Systems; Milwaukee, WI, USA). Water was chosen to act as an analog for the material of the gel dosimeter. A helical scan was performed using a 120 kVp tube voltage, 100 mA tube current, 1 \( \text{mm} \) slice width, pitch of 1, and reconstructed image separation of 1 \( \text{mm} \). The full length of the flask was obtained within 151 slices (i.e. 151 \( \text{mm} \)).

Once the 3D data is obtained, it may be used for treatment planning purposes. A relatively basic treatment plan was chosen for this dosimeter. As the system is still in its early development stages, attempting a more ambitious treatment plan would be premature. Using Eclipse\textsuperscript{TM} (Varian Medical Systems; Palo Alto, CA, USA), three 3\( \times\)3 \( \text{cm}^2 \) square treatment fields of 6 \( \text{MV} \) photons were placed at axial positions of

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**Figure 3.9:** Absorbance vs concentrations plots for dye and Duramax concentrates.
2.8 cm, 7.0 cm, and 11.2 cm relative to the 0 cm position of the isocentre (defined at the centre of the neck of the flask). Respectively, these fields targeted the flask from gantry angles 0°, 90°, and 180°. Also respectively, the maximum dosages for these beams were 600.0 cGy, 397.5 cGy, and 199.4 cGy. In monitor units (MUs, a measure of linear accelerator output), these dosages respectively corresponded to 612 MU, 407 MU, and 203 MU. The source-to-surface distance (SSD) for each field was 95.2 cm.

The dosimeter was irradiated on its day of manufacture. Treatment of the dosimeter began nearly 5 hours after the dosimeter was placed in the laboratory fridge for gelation. The same horizontal mount that was used for the x-ray CT of the water-filled flask was used to mount the dosimeter for treatment. Alignment lasers in the treatment room were used to position the dosimeter’s defined isocentre at the isocentre of the treatment unit. The treatment plan described above was executed using a Varian C-Series Clinac 6EX linear accelerator (Varian Medical Systems).

3.11 Data Acquisition

For this work, the last step to the gel dosimetry routine—evaluation—involves the imaging of the dosimeter using the fan-beam optical CT scanner. The effects of specific imaging parameter choices will be examined in Chapter 5. The following is meant to familiarize the reader with the data that is acquired and how it is used for imaging purposes.

Regarding information that is obtained by the detector array, two types of light data are acquired: \( I \) and \( I_o \). Respectively, these are attenuated and unattenuated fan-beam profiles. For \( I_o \) data, a scan can either be obtained through a bath with or without an unirradiated dosimeter in place. However, if the \( I_o \) scan is acquired through an empty bath, an amount of signal drop will occur due to reflection caused by the wall of the flask housing the polymer gel. While temporally stable, the fan-
beams attainable with this system are non-uniform. A combination of the unevenness of the fan and reflection off of the flask cause ring artefacts in image reconstructions. This issue of reflection-based ring artefacts is discussed in greater detail in Section 5.1.

**Figure 3.10:** Data acquisition samples: $I_o$ & $I$ light profiles for a uniform scattering solution. Note: the $I_o$ profile here uses a ‘blank’ flask. Flask walls are apparent near pixels $\sim 45$ and $\sim 285$.

**Figure 3.11:** Data acquisition samples: transmission & absorbance data profiles. ‘Horns’ can be seen in the $A$ profile near pixels 64, 128, 192, and 256. These cause ring artefacts in image reconstructions.

Scans performed with this system acquire $I_o$ through a ‘blank.’ This may be a flask filled with pure deionized water when scanning solution based phantoms, or an unirradiated gel sample when scanning polymer gels. The $I$ fan-beam is always
acquired through the object. See Figure 3.10 for sample profiles for $I_o$ and $I$.

Having both $I$ and $I_o$ allows for the calculation of $T$ and $A$. Respectively, these are transmission and absorbance ($OD$) profiles. Transmission values are obtained by taking the ratio of attenuated and unattenuated light ($T = \frac{I}{I_o}$). Absorbance values are then obtained by taking the base-10 logarithm of the transmission value ($A = -\log_{10}[T]$). These absorbance values are the data used for image reconstruction. For profiles of each type, these values are calculated on a ray-by-ray basis (i.e. calculated for each detector element). Any temporal drift that occurs in the fan-beam between the acquisition of $I_o$ and the acquisition of $I$ can lead to errors in the $A$ values calculated. See Figure 3.11 for sample profiles for $T$ and $A$.

3.12 Image Reconstruction

Once data is collected and absorbance sinograms are calculated, cross-sectional slices are reconstructed using MATLAB’s inverse radon transform function, \textit{iradon}. This function performs filtered backprojection and allows for a variety of reconstruction parameters to be set. However, the input data required for \textit{iradon} is an absorbance sinogram of parallel-geometry. As data collected with this system has fan-geometry, it must first be converted to parallel.

MATLAB has functions that allow for the conversion between fan and parallel-geometry sinograms. Another function, \textit{ifanbeam}, is intended to: (i) take an input sinogram of fan-beam geometry, (ii) convert it into a sinogram of parallel-beam geometry, and (iii) use \textit{iradon} for image reconstruction. However, investigations into the performance of this function have revealed inadequacies in its execution. The following subsections will describe these inadequacies and introduce the work-around method that was used for image reconstruction.
3.12.1 Reconstruction Issues Using MATLAB’s Ifanbeam Function

MATLAB’s ifanbeam function ostensibly executes two functions: \textit{fan2para} and \textit{iradon}. The first, \textit{fan2para}, is used to re-bin rays from the input fan-beam sinogram into the corresponding rays of the equivalent parallel-beam sinogram. Once the parallel-beam sinogram is calculated by \textit{fan2para}, \textit{iradon} uses this sinogram for image reconstruction.

The issue with ifanbeam lies in the execution of \textit{fan2para}. In creating the parallel-beam sinogram, only the first $180^\circ$ worth of projections (i.e. half a rotation of the dosimeter) are calculated. These are then duplicated and mirrored for the back half of the sinogram. Effectively, this completely disregards the back half of the input fan-beam sinogram. As a result, 50\% of the acquired data is ignored using the ifanbeam function due to the steps performed by \textit{fan2para}.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{ifanbeam.png}
\caption{Illustration of compromised image reconstruction using MATLAB’s \textit{ifanbeam} function. Absurd data (\textit{rectangular block}) is used to track errors. Fan sinograms are converted into parallel sinograms by \textit{fan2para}. Parallel sinograms are used to reconstruct images with \textit{iradon}. Two scenarios are illustrated above. When absurd data is added to the fan sinogram’s back-half data, it is overlooked and does not affect the reconstruction. When absurd data is placed in the front-half of the fan data, consequences arise in the reconstructed image.}
\end{figure}
Two test sinograms were used to demonstrate these steps taken by fan2para. Both having fan-beam geometry, ‘absurd’ data was added to either the front half or back half of each sinogram. This data is deemed absurd because, in sinogram space, it does not represent a reconstructable object and would result in severe imaging aberrations if reconstruction were attempted. See Figure 3.12 for an illustration of each of these sinograms, their parallel-beam sinograms as calculated by fan2para, and the resulting images that are reconstructed from these data. In light of these inadequacies, ifanbeam was abandoned for image reconstruction and a MATLAB bug report was filed. MathWorks technical support have since confirmed that an enhancement request has been submitted to their Image Processing team.

3.12.2 Fan-beam Reconstruction with MATLAB

It is likely that MATLAB’s functions convert between sinogram geometries in the way that they do because of the fact that $360^\circ$ (i.e. full rotation) parallel-beam sinograms contain two redundant halves. That is, raylines in a parallel-beam follow parallel paths as raylines from the same beam rotated $180^\circ$. Ergo, if object data were acquired perfectly, the second half of the sinogram would be identical to the first half, only mirrored and reversed. However, actual results are not ideal, and data is not acquired perfectly. By taking advantage of the full sinogram, one can ensure that data from the entire rotation is considered.

To properly reconstruct images using full $360^\circ$ rotation data, ifanbeam cannot be used. Instead, its constituent functions, fan2para and iradon, are used individually. First, two versions of the fan-beam sinogram are created—one as it was acquired, the other mirrored across both axes of profile view and angular position. This way, the first $180^\circ$ of one sinogram contains the front-half data and the first $180^\circ$ of the other sinogram contains the back-half data. Next, these sinograms are converted to parallel-beam sinograms using fan2para. At this point, the data from the mirrored sinogram is flipped as needed so that it matches its partner sinogram. These two
sinograms are then averaged, as they represent redundant halves of parallel-beam data. Finally, this averaged sinogram is used with iradon to reconstruct by filtered back-projection (filter used: Ramp-Hann).

Using this method, data from the full rotation is combined into a 180° sinogram that is ready to be reconstructed. Whether the back-half data is taken advantage of or not, the subsequent reconstruction takes the same amount of time to produce an image. Currently, it is unclear whether the back-half data is essential to reconstructions. Future investigations may find that half-rotation acquisitions will suffice. In that case, it would provide substantial advantages with regards to scan speeds. However, for this work, full rotation sinograms were used.
Chapter 4

Instrument Characterization

4.1 Light Intensity Control

The laser beam intensity—flux of photons per second—is modulated using a pair of linear polarizers. The output intensity is determined by Malus’ Law, shown below. For two perfect linear polarizers, their output intensity \( I \) is determined by their input intensity \( I_o \) and the difference between their polarization angles \( \theta \). The \( \frac{1}{2} \) in equation 4.1 comes from the assumption that the input light intensity is of random polarization.

\[
I = \frac{1}{2} I_o \cos^2 \theta \quad (4.1)
\]

Yet, the theoretical relationship above is for ideal polarizers. In reality, polarizers do not perform ideally. With the current system, each polarizer has a polarization efficiency of 95%. Therefore, after the beam passes through the first polarizer (\( P_1 \)), 95% of the output beam will be polarized at an angle corresponding to the positioning of \( P_1 \). The remaining 5% will be randomly polarized, assuming that the laser beam is randomly polarized. After passing through the second polarizer (\( P_2 \)), 95% of the beam will be polarized to the angle of \( P_2 \). Of the remainder, 4.75% will have the polarization angle of \( P_1 \), and 0.25% will remain randomly polarized. Some detectors
can be susceptible to polarization sensitivities, where their responses can be a function of polarization angle. Acknowledging this possibility, the choice was made that $P_2$ remain stationary at a constant polarization angle while $P_1$ was adjusted for intensity changes. This way, 95% of the beam maintains a constant polarization as intensity is modulated.

In addition to the imperfect polarization efficiency, the polarizers also introduce additional absorbance to the system. As light passes through each polarizer, regardless of polarization angle, it is attenuated due to optical density values inherent to the polarizer material. Specifications for these lenses state that the output intensity for a single polarizer is 30% of the input intensity, compared to 50% in the ideal case. Transmission values for polarizer pairs are not provided. Therefore, it was essential that the relationship between output intensity and relative polarization angle be established. To establish this intensity relationship as well as evaluate the polarization sensitivity of the detector array, the following investigations were performed.

Due to the limited dynamic range of the photodiode detectors, the entire polarizer intensity curve cannot be characterized with a single experimental setup. As the polarizers are adjusted to increase the intensity of the output beam, detector values begin to saturate. Therefore, additional attenuation had to be added to the beamline to allow for the observation of higher portions of the calibration curve. This was achieved using neutral density filters (NDFs).

To obtain values for the relative changes in light intensity, entire light profiles were obtained at each angular position. Upon changing the beam intensity (i.e. $P_1$ angular position), ratios were taken on a ray-by-ray basis and the median value of these ratios was used. The entire curve was collected in four parts; for each part, three sets of data were collected. The lower portion of the curve used no NDF. The upper three portions used NDFs of optical densities 0.9, 1.5, and 2.5. See Figure 4.1 for the calibration curve with the four parts indicated.
Figure 4.1: Light intensity calibration curve for the linear polarizer pair. Relative intensity ranges from a minimum to maximum of $1 \rightarrow 10180$. Angular values denote the mounting position of $P_1$, and not the angular difference between the pair. Coloured regions indicate portions collected using different NDF values (blue, $OD=0$; green, $OD=0.9$; red, $OD=1.5$; magenta, $OD=2.5$).

The movement of $P_1$ through its necessary angular positions was repeated for each of the three sets of data. This was to ensure that positioning errors were represented in the data. For each of the four parts of the curve, mean values of the three sets collected were used. At the three intercepts between NDF values, 2–5 overlapping data points were collected. These overlapping data points were used to match the four parts of the curve into one set of relative data by means of normalization. To that effect, the nominal OD values for the NDFs are not considered in the calculation of the calibration curve.

Another test was performed to test the polarization sensitivity of the system’s photodiode array. To do this, $P_2$ was removed from the beamline, leaving $P_1$, an NDF, and the line-generating lens. This way, changing the angle of $P_1$ would change the polarization angle of the output beam and not its intensity. Fluctuations in detected values would then be caused by polarization sensitivity and not by intensity.
changes. The angle of $P_1$ was moved through 180° in 10° increments, and profiles were acquired at each step. These profiles were then normalized using the mean profile value across all angles of $P_1$. The relative fluctuations for all 320 elements are plotted together in Figure 4.2. However, because the choice was made to keep $P_2$ stationary, one needs only be concerned with the effective polarization sensitivity. That is, the relative fluctuations apparent in this test are scaled down to 4.75% to reflect values that are relevant to the performance of this device. Effective polarization sensitivity is also illustrated in Figure 4.2, and fluctuations to be expected due to polarization are less than 1% (SD = 0.14%).

![Figure 4.2: Polarization sensitivity and effective polarization sensitivity of the photodiode arrays. Adjusting only the polarization angle of the input light, relative intensities for all 320 detector elements are plotted together over 180° of polarization.](image)

One must be hesitant in relying solely on the calibration curve calculated here for the polarizer pair, as the system detectors themselves are being used for the characterization. Any dependent response on their part could blindly affect the shape of the curve. Because of this, an attempt was made to collect the same curve using an optical power meter (PM100 with S120B sensor; ThorLabs, Newton, NJ, USA). However, an offset of the power meter’s output proved to be more problematic. The conclusion was reached that the data collected by the photodiode array was more reliable than the data collected by the power meter. Data from Section 4.5,
which examines the effectiveness of the collimators, supports this conclusion. There, theoretical values provided by spectrophotometric data are compared against values measured by the photodiode array. Errors in the polarizer calibration curve would reveal themselves in that comparison.

4.2 Detector Characterization

4.2.1 Data Collection

In order to observe a linear detector response, the photodiodes are operated in a photoconductive mode. In this mode, a negative voltage is applied to the photodiode (i.e. reversed biased). As a result, the current produced by the photodiode has a nearly linear relationship with the intensity of light striking its silicon element. As shown below, this relationship is nearly linear because it is composed of two components: (i) a proportionate current $I_p$, and (ii) a dark current $I_d$.

$$I = I_d + I_p$$

(4.2)

The dark current is a constant component of the detector response that remains present even in the absence of light. The proportionate current is the linear component that results from charges which are released when visible photons interact with the silicon of the photodiode.

For all intents and purposes, an optical CT system only needs to perceive relative changes in light intensity (i.e. transmission ratios). To that effect, the only component of the detector signal that is required by the system is the proportionate current, $I_p$. Yet, extracting this information from electronic specifications would prove difficult. The current $I$ is produced by each of the 64 individual elements on each photodiode array. Each of these arrays has integrated processing chip that amplifies the signal, increasing their sensitivity. It is unclear how this processing treats the
raw current. So, the arrays output a treated current, $I_t$. This analog current is then converted into a 14-bit binary number, which is what is stored in the circuit board memory and ultimately transferred to the PC. The relationship between the 14-bit value (here onward referred to as ‘bin-value’) and $I_p$ is not always clear. Therefore, proper characterization of the detector system as a whole—silicon response, treated current, sampling, and analog-to-digital conversion—is required.

A few factors went into deciding how to characterize the detector responses. First, the relationship between bin-value and photodiode integration time was important. For one, there was some question around whether or not longer integration times would provide signals with less noise. As well, detector response versus integration time should be linear; so, this could provide an indication of the dark current component of the bin-values collected. The second factor that aided the design of the characterization tests was a desire for alignment-independence. It was important that the detector response would not rely on the shape of the fan-beam used. This would avoid the requirement of maintaining a constant fan-beam alignment. The last factor involved was temporal stability. If it could be established that the system response is temporally stable, this would avoid the need to perform detector calibration tests on a daily basis.

Keeping these factors in mind, the acquisition of 10 data sets on 10 different days was planned. Of these 10 sets, 5 were collected using the SS collimator and 5 using the MH collimator. For each of these sets, a constant alignment was created and bin-values were acquired across all integration times and varying light intensities. Light intensities started at the lowest intensity that would provide any signal, and these were increased through to an intensity where all detectors had become saturated. For each combination of intensity and integration time, a sampling of 50 profiles was acquired. Mean and standard deviation values for these acquisitions were stored into signal and noise data blocks, respectively. The end result was 10
pairs—signal and noise—of data blocks (detector elements $\times$ integration time $\times$ light intensity) that were acquired on 10 separate days. When collecting the 5 sets for each collimator, varying alignments were deliberately set up to allow any potential alignment-dependent effects to be observed.

### 4.2.2 Data Analysis

To aid a better understanding of the overall response of the detector system, one should first examine the response of a single detector element. Figure 4.3 below illustrates the bin-value response versus integration time. The multiple ramps shown represent data that were acquired at different light intensities. Ramps with higher slopes correspond to higher charge accumulation rates, which correspond to higher light intensities.

**Figure 4.3:** Output for a single detector element over all programmable integration times for a variety of light intensities. Each ramp represents a different input light intensity, with steeper ramps corresponding to higher light intensities. Expected 14-bit saturation levels (16,384) are not reached. An unexplained ‘rippling’ effect occurs in groups of three across programmable integration times.
We see that there is a range of bin-values that represent valid data points, along with a range of dark values and a range of saturated values. On the lower end of the scale, bin-values do not drop below $\sim 31$. On the higher end of the scale, bin-values approach the 14-bit maximum value of 16,384 before settling around $\sim 14,000$. Because of these observations, the decision was made to set a acceptance range for data collected. Bin-values above or below this range ($35 \rightarrow 13,000$) are deemed to be unreliable.

Next, one will notice the jaggedness of each ramp as photodiode integration time increases. The ‘ripples’ of these jagged ramps appear in sets of three programmable integration times. While the cause of the pattern is unclear, it is consistent and therefore reliable. One must acknowledge this behaviour when comparing detector values of different integration times. That is, comparisons should be made only between bin-values with integration times $3n$ apart from one another so that mutual response ramps are shared between the two points.

![Offset Extrapolation for a Single Element](image1.png) ![Offset Profiles: 10 Data Sets](image2.png)

**Figure 4.4**: Detector offsets: (a) sample of extrapolation for a single photodiode element, and (b) offset profiles from 10 sets of data for the entire detector array.

The last observation to be made from the response ramps is the offset that they all share. Extrapolating to an integration time equal to zero shows a negative offset value (see Fig. 4.4). Ideally, response with respect to integration time would be
perfectly linear and these ramps would intercept at zero. To get an estimate of the
offset value for an individual detector element, linear fits for each ramp were taken
and averaged. Although the value of this offset remains fairly consistent for a single
detector element, this offset value varies greatly among different elements across the
detector array. Figure 4.4 shows a plot of this offset value across the entire 320
element profile for all 10 sets of data.

Observing the shape of the bin-value offset across the entire profile, it becomes
apparent that the detector array consists of five smaller arrays. There are five re-
peated shapes as well as significant spikes where sampling switches between arrays.
The current hypothesis attributes this offset to two components. The first component
is from dark current values that are inherently persistent due to the reversed biased
mode these photodiodes are operating in. This component would be featured across
the entire profile. The second component is only featured at the edges of detector
arrays. Upon discussing this offset with the electronics department that developed
the circuit board, the conclusion was reached that the spikes seen in the offset are
caused by inadequate operational amplifiers. The model of op-amps chosen were
expected to be just barely able to keep up with the clock rate required of them. It
would seem that faster op-amps are necessary to prevent the spike effects apparent
in the sampling of the detector array.

In the meantime, one may try to correct this offset. Considering the consistency
of this shape over multiple days using multiple alignments, it seems that the offset
is constant and does not depend on the fan-beam alignment. This persistency allows
for the calculation of a data correction. Using data from all 10 days, median values
were used to calculate the offset correction shown in Figure 4.5.

The benefit of this correction is demonstrated when comparing fan-beams of vary-
ing light intensity. Figure 4.5 provides a comparison of corrected and uncorrected
‘drops’ in light intensity. Here, one would ideally observe perfectly flat transmission
Figure 4.5: The offset correction and the demonstrated benefits gained by correcting light data acquired by the detector array. For transmission 'drops,' uncorrected ratios of light intensities are shown in red, corrected ratios in blue. Correcting the data removes spiking at photodiode array edges and reduces effects caused by fan non-uniformities.

Transmission values in red are ratios of uncorrected light profiles obtained at different light intensities. These profiles suffer greatly from spiking at array edges and have jagged features across the entire profile which correspond to the non-uniform shape of the fan-beam. Transmission values in blue are ratios of the same data with the correction added. The benefits are considerable. Spiking at the array edges is eliminated and the alignment-dependent features throughout the profile are greatly reduced. It should be noted that this offset correction is used to treat all of the data acquired for all tests performed in the current work.

The last facet to be considered with these tests is detector noise. Unexpectedly, moving to longer integration times did not result in appreciable noise reduction. However, it was observed that certain integration times did exhibit quieter behaviour. These corresponded to the $3n$ pattern that was evident in the detector response ramps of Figure 4.3. Allocating the integration times into their mutual $3n$ groups, it was observed that the $2+3n$ programmable integration times (i.e. $t = 806.7 \, \mu s + n250 \, \mu s$) provided lower noise compared to the other 2 groups (see Fig. 4.6). Based on this information, further work with the system was performed using these more
Figure 4.6: Average relative noise for all elements in a given profile versus integration time is plotted for 3 sets of programmable integration times. These correspond to the sets of ‘ripples’ observed in the detector data ramps. Shown in green, the \([0.8067 \text{ ms} + n \times 0.250 \text{ ms}]\) set of integration times show relatively lower noise than the other two sets. The nature of this benefit is unexplained.

favourable integration times.

4.3 Fan Stability

The fan-beam used for the scanner is fairly non-uniform. Regardless of which collimator is chosen or whether the tank is in place or not, this non-uniformity is persistent. This implies that fan-beam non-uniformity is an issue inherent to the method of fan-creation used—laser beam & line-generating lens. See Figure 4.7 for samples of fan profiles obtained with each collimator with various tank setups. Nevertheless, beam uniformity is not a necessity for CT. As long as the shape of the beam remains stable between the acquisitions of \(I_o\) and \(I\), transmission values will prove valid. Therefore, the stability of the fan-beam is an important aspect of the system.

The first step to achieving a stable beam is to allow the laser beam to warm up for at least an hour. This warmup is looked at both qualitatively and quantitatively here. Having already achieved proper alignment the day prior, a continuous scan of the first
Figure 4.7: Sample light profiles using a variety of setups. Top: multi-hole collimator profiles with (a) no tank, (b) tank with air, and (c) tank with a matching bath. Bottom: single-slot collimator profiles for similar tank arrangements.

An hour of laser beam warmup was collected. The relative fluctuation of the fan-beam profile during this warmup period is shown in Figure 4.8. Notice that the fluctuation does not occur equally across the beam, but is more intense in certain portions. Allowing the beam to warm up, its erratic behaviour calms to a steady level of $\sim 2\%$ relative standard deviation ($\sigma/\mu$). The mean relative standard deviation across the entire profile for this same warmup period is plotted in Figure 4.8.

In addition to allowing the laser to warm up, it is important to allow the temperature of the scanning medium to stabilize as well. As the bath equilibrates with room temperature, air bubbles form on the inner walls of the tank. Once the bath has reached room temperature, these bubbles may be wiped clear from the walls and will not reappear as long as the bath holds its temperature. This is important because air bubbles can be problematic due to their refractive properties.

Two other factors of the scanning medium affect the stability of the fan-beam:
Figure 4.8: Laser warmup. Qualitatively, profile variation over the first hour of warmup exhibit ‘waves’ of intensity fluctuations. Quantitatively, average relative noise for this same data is plotted. These assert the need to allow the laser to warmup before use.

**particulates** and **schlieren**. Dust and other impurities in the water of the bath can come into the path between fan vertex and detector, obscuring the light from the beam. When a refractive matching agent is used, inhomogeneities in the mixture—schlieren—redirect light from its original path, causing false light levels to reach the detector elements. Errors caused by each of these factors may be minimized by allowing the bath to settle after mixing. This allows dust and refractive inhomogeneities to fall to the lower regions of the tank, out of the beam. Examples of each type of error are shown in transmission scans in Figure 4.9. Also shown in Figure 4.9 are plots of mean relative standard deviation for transmission scans for a water bath and a matching bath, respectively. For each bath type, they were stirred to cause a turbulent medium, then allowed to settle over a period of 3 hours. In order to achieve a noise level roughly equaling the noise that is inherent to the light source (\( \sim 2\% \)), a water bath should be allowed to settle for 3 hours and a matching bath should be left to settle overnight.

If acquisition errors due to particulates and schlieren cannot be prevented, certain measures may be taken to limit their effects. Post-processing filters may be used for data treatment. Although, filters can also cause unwanted blurring in the true scan
**Figure 4.9:** Particulates in water & schlieren effects in matching solution: (a) and (b) show transmission scans exhibiting the potential for error caused by particulates and schlieren, respectively; (c) and (d) show plots of average relative noise for each bath type as they are left to settle for 3 hours after being stirred.
data. Another option would be to acquire multiple scans of the same dosimeter. By acquiring extra samples, one may average these samples. If possible, one would take a median value from these scans to limit the affect of such outliers. The value of acquiring multiple scans will be considered in Section 5.2.2.

The fluctuation of global fan-beam intensity may also be addressed by post-processing. For each object scanned thus far with the current system, the 320 element detector array provides superfluous elements outside the object’s diameter. Roughly 20 elements on each side of the object may act as reference detectors to monitor intensity fluctuations of the beam. With this data, scans could subsequently be corrected to a constant intensity. However, one may question whether such a correction is necessary. Fluctuations in a stable fan are reasonably small. Also, because of the nature of backprojection reconstruction, intensity level fluctuations that are profile-wide would influence the entire reconstruction field, thereby balancing out in the final image. Examinations into the value of such a correction are not performed in this work and is left for future investigations.

### 4.4 Extended Dynamic Range

A method has been devised to extend the range of light intensity levels that the system is capable of perceiving. The unextended dynamic range (DR) inherent to the system is limited by: (i) the range of valid bin-values (35 → 13,000), and (ii) the need for the offset correction. Assuming an $I_o$ fan-beam with profile-wide detector values of 13,000 before offset correction, the maximum perceivable absorbance profile is shown in Figure 4.10. This profile was calculated assuming a light attenuation which results in profile-wide detector values of 35 before offset correction.

In practice, this maximum is not achievable. For one, such a fan is not attainable with the fan-creation method that is currently being employed. Also, cylindrical attenuators do not result in a uniform drop of light intensity. Central regions of
Figure 4.10: Maximum perceivable absorbance profile using an unextended dynamic range. Assumes uniform 13,000 values acquired for \( I_o \) and uniform 35 values acquired for \( I \). The need to use the offset correction creates a DR that is dependent on pixel position.

The fan will be attenuated faster due to the longer pathlengths in the center of the cylinder. Therefore, actual maximum absorbances observed with the system will be lower than the theoretical values depicted in Figure 4.10.

Absorbance represents the signal being evaluated by the scanner. As such, it is desirable to observe a higher amplitude signal to improve the signal-to-noise ratio. To observe increased absorbances, extended dynamic range (EDR) is necessary. Increased sensitivity can be achieved in two ways. First, by lengthening the integration time of the photodiode arrays. Second, by increasing the intensity of the input fan-beam.

For the first method, scans are acquired using multiple integration times. To start, the \( I_o \) scan is acquired using the shortest integration time. Then, the \( I \) scan is acquired using the same integration time. If at any point during the scan bin-values fall below the ‘dark’ detector values, these points are deemed to be invalid and need replacement via EDR. By shifting to longer integration times, new scans can provide valid data points, which are then normalized using the programmable integration times used for the respective scans. Figure 4.11 shows a transmission profile exam-
ple of multiple scans that were acquired and normalized using the integration time method of EDR.

![Light Profile: EDR via Integration Time](image1.png) ![Absorbance Profile: EDR via Light Intensity](image2.png)

Figure 4.11: Samples of normalized profiles using Extended Dynamic Range: (a) EDR via integration time shows transmission profiles using normalized data from 5 different integration times, and (b) EDR via light intensity shows absorbance profiles using normalized data from 4 different light intensities.

For the second method, scans are acquired using multiple light intensities. To start, the $I_o$ scan is acquired using a lower light intensity. Then, the $I$ scan is acquired at this same intensity. Like in the case above, if dark values occur during the scan, they require replacement. By increasing the light intensity, valid data points are acquired, which are then normalized using the calibration curve determined for the linear polarizer pair. Methods of EDR using light intensity have been implemented by other optical CT researchers working with highly-attenuating samples [58, 74]. Figure 4.11 shows an absorbance profile example of multiple scans that were acquired and normalized using the light intensity method of EDR.

In both methods, one acquires multiple $I$ scans. In these scans, portions of the sinogram will be dark while other portions will saturate. As a result, a rejection scheme is first implemented that flags any data which falls outside of the valid data range of the detectors. After this, the multiple $I$ scans are normalized. Then, these normalized and flagged scans are averaged to obtain a single sinogram that represents $I$. To do this, a ray-by-ray routine is used which counts the number of valid data
points for each ray and averages them appropriately.

Considering the integration time and light intensity ranges of the system, significant increases in maximum perceivable absorbance are possible. Going from integration times of 803.33 $\mu s$ to 21.807 $ms$, the maximum perceivable absorbance increases by 1.43. Comparing the relative light intensities of the lowest and highest positions of the polarizer pair ($\times 10,180$), the maximum perceivable absorbance increases by 4.01. Together, they combine for an increase of 5.44. Maximum perceivable absorbance profiles for unextended DR, EDR via integration time, EDR via light intensity, and EDR via both methods are shown in Figure 4.12.

![Maximum Perceivable Absorbance](image)

**Figure 4.12:** Maximum perceivable absorbance: unextended DR (red), EDR via integration time (green), EDR via light intensity (blue), and EDR using a combination of both methods (black).

### 4.5 Collimator Comparison

When given the option between the two collimators, one would expect that the MH collimator, which provides physical collimation for scatter-rejection, would be more beneficial. This should especially be the case when interrogating scatter-based dosimeters. However, the benefit of the MH collimator must be demonstrated. It is not clear whether it provides an appreciable benefit. Could the SS collimator perform
just as well? The following tests were designed to find out.

The design of these tests was inspired by work performed by Al-Nowais and Doran using the Vista\textsuperscript{TM} scanner [58]. In their work, they evaluated highly-attenuating absorption-based phantoms (blue dye in water). For their most opaque samples ($A>1.5$), they observed cupping artefacts. This was perplexing, as the previously believed cause of the cupping artefact was scatter contamination [69]. Because the samples they were examining were not scatter-based, they were unable to conclude what was causing the artefacts they observed.

For tests with the fan-beam scanner, both scatter-based solutions (Duramax polymer) and absorption-based solutions (blue food colouring) of varying concentrations were interrogated. As was described previously in Section 3.9.2, these solutions may be prepared to known opacities. Samples of both types of attenuators were each examined by the MH collimator (\textit{scatter-rejection}) and the SS collimator (\textit{relatively no in-plane scatter-rejection}). Employing these two levels of collimation allows for useful comparisons to be made.

First, tests for both the dye and Duramax were performed using the SS collimator. Then, these same tests were repeated using the MH collimator. For each set of tests, a 1L flask with a porthole cut into its upper wall was used. This porthole allowed portions of concentrate to be added incrementally to the container without the need to remove it from the mounting arm. Beginning with 250 mL of deionized water, set amounts of concentrates were added using a micropipette (Pipetman P200, Gilson, Inc.; Middleton, WI, USA). The porthole was large enough so that the solution could be stirred before it was scanned (see Fig. 4.13). For all of these tests, an initial light sinogram of the pure water was acquired to act as $I_0$. Subsequently, $I$ sinograms were collected for each concentration level. Multiple $I$ scans were compared to a common $I_0$. For scan parameters, 360 projections were acquired over 360$^\circ$ (i.e. in 1$^\circ$ increments). At each projection angle, 3 views were acquired and averaged. EDR
via integration time was used for the majority of the scans, as it was necessary for all but the lowest opacities. To observe higher opacity samples ($A>2$), EDR via light intensity was used.

Three types of data were used for analysis. *Measured* and *theoretical* absorbance values were calculated for the central pathlengths through the samples. For these, spectrophotometric data allowed the calculation of theoretical values (i.e. ideal values expected to be measured) for a $\sim 93 \text{ mm}$ pathlength, and the central 3 pixels of the fan-beam sinogram data were averaged for measured values. Secondly, the shapes of absorption profiles were observed as opacity increased. Lastly, image reconstruction profiles were taken and also observed as opacity increased. With all of these data types, equal concentrations of the same solution types were compared using different collimators. This way, their relative differences could provide insight into the relative performances of the two collimators.

Figure 4.13: Equipment used for collimator comparison tests (*as well as imaging protocol investigations later in Chapter 5*). In (a), the micropipette with disposable tips and the flask with rectangular porthole. In (b), a photograph of the highest concentration of dye imaged for the collimator comparisons (i.e. $A_{93\text{mm}}=4.09$).

Figure 4.14 (a) shows the measured versus theoretical absorbance values for increasing concentrations of blue dye using the MH collimator and the SS collimator.
Absorbances were accurately measured using both collimators up to $A = 2.2$. For the highest concentration, measurements using the single-slot collimator understated absorbance values ($A_{th} = 4.09$, $A_{mSS} = 3.71$). Measurements at this concentration using the multi-hole collimator maintained linearity ($A_{mMH} = 4.09$).

Considering the understated measurement of the highest concentration sample using the SS collimator, one may approximate the portion of light reaching the detectors that consist of the primary beam and the portion that represents contaminating light. Theoretically, at this level of absorbance, one expects to see $0.008\%$ transmission (i.e. $10^{-4.09}$). The value of transmission measured is $0.019\%$ (i.e. $10^{-3.71}$). This corresponds to a true $\frac{L}{L_0}$ value of $0.008\%$ being contaminated by an additional $0.011\%$ of light. Essentially, $\sim 58\%$ of the light reaching the central detectors consists of signal contamination by stray photons.

Figure 4.14 (b) shows the measured versus theoretical absorbance values for increasing concentrations of Duramax polymer using the MH collimator and the SS collimator. Absorbances were accurately measured using both collimators up to $A = 1.6$. For the highest concentration that provided measurements with both collimators, measurements using the single-slot collimator again understated absorbance values ($A_{th} = 3.84$, $A_{mSS} = 3.32$). Measurements at this concentration using the multi-hole collimator maintained linearity better than the single-slot, although slightly overstated this value ($A_{mMH} = 3.91$). Two higher concentrations also provided measurements using the SS collimator. These values were greatly understated ($A_{th} = 7.55, 14.54; A_{mSS} = 3.72, 3.87$). Such understatements were to be expected, as the attenuation mechanism of the Duramax polymer is scatter. Considering the understated measurements of the three highest concentrations using the SS collimator, they correspond to detected light signals respectively consisting of $\sim 71\%$, $\sim 99.99\%$, and $\sim 100\%$ scattered photons.

The slight overstatement of measured values using the MH collimator could be
Figure 4.14: Collimator comparisons: Measured versus theoretical absorbances along central raylines for blue dye & Duramax polymer are shown in (a) & (b), respectively. Respective absorbance profiles are displayed in (c) & (d) with single-slot collimator data shown in dashed red and multi-hole collimator data in solid black. Blue dotted lines in (a) & (b) indicate expected values.
caused by two factors. First, the calibration curve for the polarizer pair could be inaccurate, causing poorly normalized light values to be used for $I$ when implementing EDR via light intensity. Second, an error in the preparation of the Duramax solution could have accidently introduced extra scatterer into the sample. Data indicates that the latter is more likely. If a poorly calibrated light curve were the sole cause, a similar overstatement would have been observed with the highly-attenuating dye solutions.

The remaining types of data to be compared are more qualitative in nature, but still provide useful illustrations of data aberrations that can occur as opacity is increased. The first of these appear in Figure 4.14 (c). Here, absorbance profiles for increasing concentrations of blue dye using each collimator are shown. Profiles using the MH collimator maintain accurate shape for all concentrations examined. With the SS collimator, flattening is apparent with the highest concentration. Similar behaviours are seen with the Duramax samples (see Figure 4.14). While the MH collimator profiles maintain accuracy, SS collimator profiles flatten at higher concentrations. Flattening is especially apparent with the two highest concentrations. Note that with all of the data, slight ‘horns’ are apparent near the 150 and 270 pixels. These correspond to the array edge detectors; evidently, the offset correction does not completely compensate for the data-corruption which occurs here. Also, it should be noted that the absorbance profiles shown are taken from the parallel-geometry sinograms just prior to image reconstruction. This is why these profiles show a higher number of pixels than there are detector elements.

The last pieces of data to be observed come from the images reconstructed with the above absorbance data. Figure 4.15 (a) shows profiles taken through the center of the reconstructed cylinders of dye solutions. The ‘horns’ seen in absorbance profiles manifest as ring artefacts here for both collimators. It would appear that data corruption at the array edges must be better addressed, as this corruption causes
problematic issues in final images. Figure 4.15 (b) shows profiles taken through reconstructions of Duramax solutions. Here, ringing is also apparent for both collimators. For the highest concentrations of both attenuator types, profiles using the SS collimator data show increased understatement in the central regions of the cylinder (i.e. cupping).

![Figure 4.15: Collimator comparisons: profiles through reconstructed images of increasing concentrations of: (a) blue dye, and (b) Duramax polymer. Single-slot collimator data shown in dashed red, multi-hole collimator data in solid black. Cupping and spiking artefacts are observed in SS collimator reconstructions. Ringing artefacts near pixels 100, 205, 310, and 415 are apparent with both collimators.](image)

An additional, unprecedented artefact is seen in the SS collimator reconstructions of high concentration samples. Severe spikes begin to appear in the centre of these images and do not appear in images using MH collimator data. While the cause of this ‘spiking’ artefact is unclear, it may be due to the redistribution of light reflected off of the elements of our detectors. The multi-hole collimator would limit the detection of such stray light.

Upon considering the above data, one may conclude that the multi-hole collimator is best suited for imaging with the current system. The MH collimator provides improved scatter-rejection and is not susceptible to cupping or spiking artefacts. It is also rather impressive that the system maintains its linearity with the MH collimator to the full range provided by current EDR methods. That is, the current system is
limited by its dynamic range and not by scatter contamination even when scatter-based attenuators are used. This allows polymer gel dosimeters to be irradiated to higher dosages without their optical evaluation being compromised by effects of stray light.

4.6 Speed

The imaging protocol used as well as the scan orchestration code in MATLAB have not been optimized for the sake of speed. Inevitably, future investigations will become more dedicated to improving scan time. At this point in development, enough data is available to warrant a hypothetical discussion on the potential scan times that could be achieved by the system.

For this discussion, three system factors will be considered: (i) detector acquisition speeds, (ii) dosimeter motion speeds, and (iii) data transfer speeds. If a Scan-Scan-Get routine were used, where multiple scans are stored to local circuit board memory, the system is capable of acquiring \( \sim 186 \) projections per second. This rate could potentially be doubled if the charge-clearing scan were eliminated in the data sampling routine performed by the circuit board. For dosimeter motion, the maximum rotational speed of the current rotating motor is \( 80^\circ/s \), and the maximum translation speed of the vertical motor is \( 2.5 \text{ mm/s} \). With respect to data transfer rates, the current USB driver transfers at \( 350 \text{ kbps} \). This rate could be increased substantially if a custom driver were written (maximum speed: \( 12 \text{ Mbps} \)).

Now, consider a 10 cm high volume of a cylindrical dosimeter. If a 1 mm axial slice interval were chosen, this volume would consist of 100 two-dimensional slices. Using full 360\(^\circ\) sinograms for each slice, the dosimeter would have to be rotated 100 times for a single scan. The vertical motor would only operate after each rotation was completed. Thus, regarding motion of the dosimeter, rotational speed is the limiting factor. For 100 full rotations, the maximum speed of \( 80^\circ/s \) would execute this 10 cm
high scan in roughly 7.5 minutes. If pre- and post-irradiation scans were acquired for each dosimeter, this equates to 15 minutes in total scanning time (9 seconds per slice).

Considering the current detector sampling speed and the maximum dosimeter rotation speed, the system could acquire 837 projections for each rotation. At the moment, memory on the circuit board can accommodate only 6,500 projections. Therefore, if the scan described above were to be attempted, additional memory would need to be added to the circuit board to house the data from the entire scan. As well, once data has been stored to onboard memory, data transfer must follow. Currently, the USB connection can transfer roughly 65 projections per second. So, the 83,700 projections acquired in a single scan would take roughly 21 minutes to transfer (42 minutes total for both pre- and post-irradiation scans).

In the above scenario, the scan data for a 10 cm high portion of a cylindrical dosimeter could be obtained in just less than an hour (57 minutes). Speeds could be improved by increasing data transfer rates with a custom USB driver. Also, the sampling speed could be reduced, thereby reducing the total number of projections that need to be transferred. As well, the number of projections collected could also be reduced if it is found that full 360° sinograms are not necessary for slice reconstruction. Lastly, dose distributions calculated by treatment planning software typically has a resolution of 2.5×2.5×2.5 mm$^3$. Therefore, practically speaking, the axial slice interval could be increased, as fewer slices would be necessary for dosimetric comparisons between measured values and planned values.

Nevertheless, these scenarios are only hypothetical. The scan speeds of the system have yet to be prodded, and one must consider a number of issues before doing so. Firstly, a volumetric scan has yet to be attempted with the scanner. Thus far, only reconstructions of single slices have been performed. Secondly, if EDR methods were to be implemented, these would require repeated scans of the dosimeter. Finally, there
are issues regarding the mechanic stability of continuous dosimeter motion. It is not yet clear whether the maximum speed of the rotational motor is consistent enough to sustain accurate interrogation of the dosimeter. However, a test continuous scan has been performed at a lower rotational speed. This will be mentioned in Chapter 5, alongside other imaging protocol investigations.
Chapter 5

Imaging Protocol Investigations

For any optical CT system, there are a number of decisions that must be made regarding the data acquisition and imaging protocol. This chapter presents investigations exploring these decisions. Section 5.1 examines the path rays take as they traverse the scanning medium. This examination aims to better understand how ring artefacts are reduced by scanning a ‘blank’ dosimeter. Section 5.2 evaluates the effect of various imaging parameters have on image noise.

5.1 Light Ray Traversal Through the Scanning Medium

The imaging tests that were performed in Section 4.5 used a full sinogram scan of the water-filled flask to serve as $I_o$. This scan of the water-filled flask, which can be considered a ‘blank,’ was compared to scans of flasks with scattering solutions (i.e. $I$). Yet, this choice of $I_o$ has not been fully explained so far in this work. The tests comparing the relative effectiveness of the two collimators were performed after it had already been determined that imaging a ‘blank’ was necessary to avoid ring artefacts. This section details the reasoning behind this choice.

First, to better understand how fan-beam intensity is decreased, one must consider the path the beam takes between the fan vertex and the collimator & detector system. Along this path, light intensity can decrease by: (i) attenuation due to the optical densities (scatter-based or absorption-based) of media, and (ii) reflection which occurs
at the various interfaces along the path.

See the diagram shown in Figure 5.1. Assuming that this tank contains a bath that is matched to the refractive index of the gel dosimeter, two scenarios for identical rays are depicted. On the left, a ray traverses the tank from the vertex, to the entry window \((a)\), through a length of the matching bath \((l)\), exits through the outer wall \((d)\), and approaches the detector system. Exponential attenuation will occur along the pathlength \(l\) if there is absorption and/or scatter inherent to the matching bath. Additionally, reflection will occur at interfaces \(a\) and \(d\). On the right of the diagram, an identical ray will experience the same amount of reflection at interfaces \(a\) and \(d\), as well as reflection at interfaces \(b\) and \(c\) of the dosimeter wall. With regards to exponential attenuation, the beam will be attenuated more along pathlength \(l'\) through the dosimeter if the dosimeter is of a higher optical density than the matching bath.

![Figure 5.1: Rayline diagram through a matched bath with and without a dosimeter in place. With the refractive index of the bath matched to that of the gelatin, raylines are assumed to be unaffected by refraction. With regards to reflection, interfaces at \(a\), \(b\), \(c\), and \(d\) all occur for the ray through the bath with dosimeter. Without the dosimeter in place, interfaces \(b\) and \(c\) are not present. So, in addition to attenuation along the length of the dosimeter \(l'\), attenuation due to reflection at \(b\) and \(c\) must also be accounted for. As well, if the dosimeter is housed within a container, a short distance of the container wall is traversed at both \(b\) and \(c\).](image-url)
In reality, interfaces a and d actually represent a pair of interfaces. For a, this is from air to entry window material and from entry window to matching bath. A similar pair of interfaces is seen at d. For interfaces b and c, these represent pairs of interfaces if the dosimeter is held within a container (e.g. a flask), or they each represent a single interface if the dosimeter is a solid material (e.g. PRESAGE\textsuperscript{TM}). In this discussion, we designate reflection constants <1 for each interface to represent the portion of the incident beam that is not removed by reflection: \( c_a, c_b, c_c, c_d \).

Next, we designate a uniform linear attenuation coefficient to represent the opacity of the bath: \( \mu_{bath} \). Then, we designate a linear attenuation variable to represent the dosimeter which is being imaged: \( \mu_{gel} \). Lastly, we can define linear attenuation coefficients for the container material (\( \mu_{flask} \)) and the length travelled through the container walls at b and c (\( l_f \)). For the case of a container-less dosimeter, \( l_f = 0 \).

For the case of the containers used in this work, \( l_f = \sim 2 \ mm \), as each wall is \( \sim 1 \ mm \) thick. Now, consider the pathlengths from the initial intensity \( I_v \) at the vertex to the final intensity \( I \) at the detector system.

Through the bath only:

\[
I = I_v c_a c_d 10^{-\left(l \mu_{bath}\right)}
\]  

(5.1)

Through the bath with a dosimeter in place:

\[
I = I_v c_a c_b c_c c_d 10^{-\left((l-l'+l_f)\mu_{bath}+(l')\mu_{gel}+(l_f)\mu_{flask}\right)}
\]

(5.2)

Now, consider the scenario where \( I_o \) was chosen to be transmission values through the bath only (Eq. 5.1), and was compared to transmission values through the bath with dosimeter in place (Eq. 5.2), the transmission value required for imaging
purposes is given:

\[ T = \frac{I_v c_a c_b c_c d \cdot 10^{-(l-l')\mu_{bath} + [l'\mu_{gel} + l_f\mu_{flask}]}}{I_v c_a c_d d \cdot 10^{-(l\mu_{bath})}} = c_b c_c 10^{-(l'\mu_{gel} - \mu_{bath}) + l_f\mu_{flask}}} \] (5.3)

Subsequently, absorbance is calculated:

\[ A = \log_{10} \frac{1}{T} = l'\mu_{gel} - \mu_{bath} + \left\{ l_f\mu_{flask} + \log_{10} \frac{1}{c_b c_c} \right\} \] (5.4)

As can be seen in Equation 5.4, the reflection that occurs at the walls of the dosimeter and any attenuation due to absorbance in the flask material will be represented as an additive error in the final absorbance value. For imaging purposes, one is seeking the variable attenuation coefficient \( \mu_{gel} \) along the pathlength \( l' \). To remove the presence of this additive error, one does not choose \( I_o \) to be an empty bath, but rather a ‘blank’ dosimeter. For instance, if one were to scan a dosimeter pre-irradiation and post-irradiation, the resultant absorbance would be represented by the change in opacity of the dosimeter only: \( A = l'\mu_{post} - \mu_{pre} \). In this way, both reflection and the inherent attenuation caused by both the bath and the flask material are removed from the calculation completely. Also note that the value for \( I_v \), which is determined by the shape of the fan-beam, is also removed from the calculation.

The additive error in Equation 5.4, or some other flask-related issue, is evidently responsible for the ring artefacts that plagued the system previously. By choosing to scan a ‘blank’ to act as \( I_o \) rather than scanning an empty tank for \( I_o \), ring artefacts due to reflection are eliminated. A side-by-side comparison of images reconstructed from the same \( I \) data is shown in Figure 5.2. Note that ring artefacts due to array-edge data corruption remain present in the image that has been reconstructed with a ‘blank’ \( I_o \) sinogram. Also, it should be noted that remarkable noise reduction is also attained by acquiring a full sinogram of the flask for \( I_o \). The remaining sections in this chapter will examine the relationships between imaging protocol and noise.
Figure 5.2: Comparison of reconstruction using a ‘bath only’ $I_o$ (a), and one using the full sinogram of a ‘blank’ for $I_o$ (b). Ringing due to fan non-uniformities and reflection are eliminated in (b), while ringing due to array-edge data corruption remains. Also, considerable amounts of noise due to flask surface variances are eliminated by acquiring a full sinogram for the ‘blank’ $I_o$.

5.2 Protocol & Parameters

In order to investigate the influences of various scanning parameters, a solution of Duramax polymer was chosen ($A_{93mm} = \sim 1.2$). The concentration was high enough to provide an acceptable signal-to-noise ratio while not being so opaque that EDR via light intensity was required (though, EDR via integration time was necessary). Working with a constant opacity allowed for a variety of comparisons to be made regarding scanning protocol and imaging parameters.

For these sets of tests, only the multi-hole collimator was used. The same porthole flask that was used for the collimator comparison tests was used here, along with the same technique for creating solutions of known opacity. Yet, measurement accuracy is not being evaluated here. These tests were meant to examine the relationship between noise and imaging protocol. As such, a constant region of interest (ROI) was chosen to provide a quantifier for all of these tests. In order to avoid the regions in images affected by array-edge data corruption, the ROI was chosen to be a disc lying just outside of the inner rings and just within the outer rings (see Fig. 5.3).
This ROI sampled 46,192 pixels, and relative noise \( (\sigma/\mu) \) from this sampling was used as a quantifier.

Figure 5.3: Here, a typical image is shown in (a), exhibiting the familiar ring artefacts due to array-edge data corruption. To avoid these contentious regions of the image, a disc region of interest between these rings was chosen. The ROI—indicated in (b) in red—sampled 46,192 pixels. From this sampling, relative noise \( (\sigma/\mu) \) was used as a quantifier.

5.2.1 Choosing ‘\( I_o \)’

As has already been explained above, \( I_o \) needs to be acquired through a ‘blank’ dosimeter to avoid ring artefacts due to reflection. Yet, this raises other questions about how sensitive the system is to the ‘blank’ scan. \textit{Will a single profile or an average profile through a rotation of the flask suffice? Can an unirradiated slice serve as \( I_o \) for a separate, irradiated slice? Once \( I_o \) is acquired, how long does it maintain its validity?} This section examines these questions regarding \( I_o \).

The same slice of water is scanned before and after the scattering agent is added. Initially interrogating a water-filled flask, 360 projections were acquired over the full rotation (i.e. 1° increments) acquiring a single profile sample per angular position. This scan would provide \( I_o \) for the subsequent \( I \) scan. However, the \( I_o \) sinogram may be used in two fashions: (i) \( I/I_o \) comparisons may be made incrementally on a projection-by-projection basis where light profiles are compared for each angle of the sinogram, or (ii) an average of the \( I_o \) sinogram can provide a single profile against which all \( I \) profiles may be compared. Examining the same data in these two fashions,
it is evident that the former technique provides significant advantages (see Fig. 5.4).

**Figure 5.4:** Here, a ‘blank’ was used for \( I_o \) in both images. In (a), the \( I_o \) sinogram was used for comparison against \( I \) on a projection-by-projection basis, while in (b) an average projection for the \( I_o \) sinogram was used to compare against all of \( I \). It is evident that dramatic reductions in noise (from 22.4\% relative noise to 3.1\%, here) can be attained by using a projection-by-projection comparison.

Using a projection-by-projection method of comparison provides significantly reduced noise levels. Examining the disc ROI, relative noise was calculated at 22.4\% and 3.1\% for the average profile and projection-by-projection methods, respectively. This would imply that there are significant variations in the surface quality of the container flask that the projection-by-projection method accommodates. These are illustrated in Figure 5.5. In the top left, the relative fluctuations observed over the full rotation of the flask are plotted for each ray. These fluctuations are large in amplitude and introduce significant amounts of variation in the \( I_o \) data. In the top right, an error map plots all of the pixels that fall outside of the range 90\%<\( X <110\% \). These fluctuations, however, are reproducible. As shown in the bottom of Figure 5.5, scanning the same slice twice and comparing rays from the two scans reveals that the fluctuations observed are indeed due to constant qualities of the flask’s surface. The corresponding error map illustrates this further, having error pixels that are independent of the container. These pixels are most likely due to particulates in the bath. Schlieren effects would not be present here, as no refractive agents have been
Figure 5.5: Variations in the surface quality of the flask wall can cause considerable noise in image reconstructions. 360 projections were acquired through a 360° scan of a water-filled flask: (a) shows the rayline fluctuations relative to their average transmission values through the rotation (scale in %), and (b) shows an error map of values $>110\%$ and $<90\%$. When scans are repeated, these fluctuations are shown to be consistent. The same 360° scan of the water-filled flask was performed twice, and (c) shows relative changes observed pixel-by-pixel between the two scans (scale in %). (d): a corresponding error map with only errors due to particulates remaining.
added to the bath.

It has been shown that images reconstructed from $I$ and $I_o$ data at a given slice are very sensitive to the surface quality variations of that slice. Considering this, one would expect that this sensitivity is dependent on the uniqueness of the slice. Next, different vertical slices of the flask were scanned to see whether they shared common surface variations. Similar scan protocols as above were used (i.e. $n_{\text{samp}} = 1, n_{\text{proj}} = 360$). Yet, in these tests, full $I_o$ sinograms were acquired at different vertical positions of the flask before the scattering agent was added and a final $I$ scan was obtained. Using the multiple $I_o$ scans acquired, the $I$ scan could be compared on a projection-by-projection basis against slices at a distance $D$ away from its vertical position.

As one might expect, considerable amounts of noise are introduced to the final image when different slices of the flask are compared. Even a slight displacement between the $I$ and $I_o$ slices ($D = 1 \text{ mm}$) leads to a considerable jump in ROI noise (from 2.6% to 15.5%). Image reconstructions for various values of $D$ are shown in Figure 5.6. These jumps in noise are attributed to the differences in surface quality between the separate slices. Although projection-by-projection full sinogram comparisons were used for these tests, the value of these sinograms are less relevant when different surfaces are being compared. This is illustrated in Figure 5.7. Similar to the sinogram examinations shown in Figure 5.5, these examine a 50 mm vertical translation of a water-filled flask. Relative fluctuations over the vertical translation are not quite as severe as the relative fluctuations observed over a full rotation, but are nevertheless present. These fluctuations are also shown to be reproducible, as seen in the bottom of Figure 5.7. The relationship between ROI noise and the distance between slices compared is shown in Figure 5.8. All of these points indicate that full sinograms of $I_o$ and $I$ scans should, ideally, be collected for each slice of the flask. This would require pre- and post-irradiation scans of dosimeters and some sort of
Figure 5.6: Full $I_o$ sinograms of the water-filled flask were acquired at different slices. In (a), the $I$ and $I_o$ data was acquired from the same slice. In (b), (c), and (d), the $I_o$ sinogram acquired was a distance ‘$D$’ mm away, as indicated. Even at 1 mm displacement, considerable amounts of noise are introduced ($ROI$ relative noise respectively: 2.6%, 15.5%, 25.3%, and 28.3%).
Figure 5.7: Variations similar to those seen across rotations of the flask are seen in vertical translations of the flask. 500 projections were acquired through a 50 mm vertical scan of a water-filled flask: (a) shows the rayline fluctuations relative to their average transmission values through the translation (scale in %), and (b) shows a corresponding error map of values $>110\%$ and $<90\%$. Similar to the rotation case, these fluctuations are consistent. The same 50 mm scan of the water-filled flask was performed twice, and (c) shows the relative changes observed pixel-by-pixel between the two scans (scale in %), and (d) shows a corresponding error map with only errors due to particulates remaining.

Figure 5.8: Relative noise in the disc ROI is plotted for delayed scans and for comparisons between different slices. While the best results are observed with the shortest time between acquiring $I_o$ and $I$, penalties for delaying $I$ scans are minor. On the other hand, considerable amounts of noise are introduced when different slices of the flask are compared. Shifting only 1 mm away, relative noise jumps from 2.6% to 15.5%.
data registration method to mutually align the pre- and post-data.

The investigations above show that the scanned data chosen to represent \( I_o \) is very sensitive to variations in the physical quality of the flask. However, it was found that the data chosen to represent \( I_o \) is much less sensitive to temporal variations. Using a projection-by-projection method for the exact same slice of the flask, multiple \( I_o \) sinograms were collected over a period of an hour before the scattering agent was added and \( I \) was acquired. This way, the same \( I \) data could be compared against \( I_o \) data of varying temporal differences. As would be expected, the best images were reconstructed using \( I \) and \( I_o \) data with the least delay between their collection. However, the penalties to be paid for delaying the \( I \) scan were minor (see Fig. 5.8). This speaks to good temporal stability in the shape of the fan-beam.

5.2.2 Imaging Parameters

Once one has decided on the appropriate scanning protocol to be used for their \( I_o \) and \( I \) scans, imaging parameters for the collection of these scans must be considered. In this work, two key imaging parameters were examined: (i) the number of samples per projection angle \( (n_{samp}) \), and (ii) number of projection angles per rotation \( (n_{proj}) \).

The first set of tests—examining the effect of multiple sampling—used full \( I_o \) sinograms of the same slice as \( I \), acquiring 360 projections over the full rotation of the flask. For both \( I_o \) and \( I \), varying numbers of samples were acquired and averaged at each position in the scan. As can be seen in Figure 5.9, there is no significant relationship between acquiring extra samples and reduction in image noise. This speaks to the stability of both the fan-beam light source and the detector system. Essentially, these are reliable enough to allow for accurate data acquisition with a single sampling.

The second set of tests—examining the effect of projection angles—used full \( I_o \) sinograms of the same slice as \( I \), acquiring a single sample per projection angle. For both \( I_o \) and \( I \), the number of projection angles used for the full rotation of the flask
**Figure 5.9:** Relative noise in the disc ROI is charted with respect to number of samples acquired at each projection angle (using 360 angular positions per rotation, i.e. $1^\circ$ increments), and number of projection angles acquired per rotation (using 1 sample per angular position). Evidently, significant gains are not obtained by acquiring multiple samples at each angular position. Regarding number of projection angles, reductions in noise are achieved with increasing number of positions; although, these reductions are relatively minor beyond $n_{proj} = 360$.

**Figure 5.10:** Best spent acquisitions. Here, the image in (a) used twice as many views as the image in (b), yet exhibits a ‘rougher’ quality. In (a), 360 projection angles were used with 10 samples per angle ($ROI_{noise} = 3.1\%$). In (b), 1800 projection angles were used with 1 sample per angle ($ROI_{noise} = 2.5\%$).
was varied (i.e. angular increments). Figure 5.9 shows the noise measured in the disc ROI as the number of projection angles is increased. As it shows, undersampling of angular positions can result in significant increases in noise. However, above 360 projections per rotation, the advantages gained are not as significant.

With this information regarding sampling and projection angles, better scan protocols can be chosen to improve image quality through efficient data acquisition. For example, it has been shown that acquiring multiple samples per projection angle is not an efficient use of data acquisition. One is better off acquiring single views at an increased number of projection angles. This point is illustrated by an image comparison in Figure 5.10. There, the first image was created from twice as much data than the second, yet it exhibits a ‘rougher’ quality.

5.2.3 A Test Continuous Scan

Currently, the system uses a step-and-shoot scan routine where the flask is rotated in angular increments and allowed to come to a complete stop before the detectors are sampled. For exploratory purposes, a continuous scan was attempted. This scan used a ‘Scan-Scan-Get’ data acquisition routine, which stored multiple scans to the circuit board’s local memory prior to transfer to the PC. The rotational stage for the flask was set to a constant angular velocity, and profiles were acquired as the flask rotated. For this scan, EDR via light intensity was used rather than EDR via integration time. As a result, two light intensities were required for the $I_o$ scan, and three light intensities were required for the $I$ scan (i.e. the scan of the scattering solution). The same slice was examined for both $I_o$ and $I$, and these were compared on a projection-by-projection basis.

Empirically, it was found that higher rotation speeds were less consistent than lower speeds. Inconsistencies in speed meant that matching profiles between $I$ and $I_o$ proved to be somewhat difficult at $80^\circ/s$. Therefore, a lower rotational speed, $11.1^\circ/s$, was chosen which allowed for $I$ and $I_o$ sinograms to be easily compared. Using this
speed, 6000 projections were acquired for each rotation in 32.3 seconds. Thus, the total ‘scan’ time including all \( I_o \) and \( I \) scans was 2 minutes and 42 seconds. To lower this time, one must find a way to reliably operate at higher angular velocities.

![Figure 5.11](image_url)

**Figure 5.11:** A test continuous-scan acquisition was attempted. In (a), a step-and-shoot method of acquisition was used, acquiring views at 1800 projection angles (ROI noise = 2.5%, total scan time = 24.8 minutes). In (b), a continuous-scan method was used, acquiring views at 6000 projection angles (ROI noise = 4.8%, total scan time = 2.7 minutes). The continuous-scan image shows increased streaking, yet ringing seems less apparent.

A side-by-side comparison of an image using the step-and-shoot method and an image using the continuous-scan method is shown in Figure 5.11. The step-and-shoot image used 1800 projection angles with a single sample per projection and required a total scan time of 24.8 minutes. One will notice that streaking effects in the continuous-scan image are more prevalent. These streaks correspond to the seams of the flask. Streaking due to these seams are also present in step-and-shoot methods, though they are not as widespread. An interesting attribute in the continuous scan is the reduced appearance of ring artefacts. It is currently unclear why ringing would be reduced here, as the ring artefacts are a product of detector data corruption and not the physical attributes of the object being scanned. Two hypotheses are currently being considered: (i) the amount of streaking introduced is disruptive enough to mitigate the presence of these rings, or (ii) the difference in ringing is caused by differences in EDR via light intensity and EDR via integration time.
Chapter 6

Polymer Gel Imaging

An nPAG dosimeter was prepared and irradiated with a relatively simple treatment plan. The gel was treated by three square beams from three gantry angles in three different portions of the cylinder. See Figure 6.1 for a photograph of the irradiated dosimeter. These three beams are identical with exception of their peak dosage levels—6 Gy, 4 Gy, and 2 Gy.

The purpose of this dosimeter was to allow preliminary gel imaging tests to be performed. This chapter presents the current imaging results and provides new observations to inform the imaging of subsequent gel dosimeters.

6.1 Scanning Method

Indications seen in Chapter 5 suggest that huge benefits may be gained by obtaining $I_o$ and $I$ from the same slice on a view-by-view basis. Doing so is simple enough using a water-filled flask with a porthole. By adding scattering agent through the porthole, the flask need not be removed from its mounting arm. Sinograms for $I$ and $I_o$ are kept mutual by not disturbing the placement of the flask. However, for a gel dosimeter this is not so simple. Acquiring pre- and post-irradiation scans will require a method of positioning registration that has not been derived in development so far. Ergo, images to be reconstructed here will unavoidably suffer from quality variations seen on the container walls.
Figure 6.1: Photograph of the irradiated nPAG dosimeter illustrating the 6 Gy, 4 Gy, and 2 Gy beams.

The slight separation of each beam provides low-dose regions that may act as $I_o$. Between each beam pair is a slice of gelatin that has not been irradiated by the primary beam. However, these slices are still exposed to the integrated scattered dose from all three beams. Therefore, this scattered dose must be acknowledged in the analysis of imaging results. Slices for $I$ were acquired through the center of each beam, and slices for $I_o$ were acquired through the low-dose regions between beams (see Fig. 6.2). Therefore, one should not expect to measure the distribution of polymerization caused by each beam. Rather, one should expect to measure the difference in polymerization between the $I_o$ and $I$ slices (i.e. ‘beam dose’ − ‘scattered dose’). The expected dose maps to be measured are also shown in Figure 6.2. These were calculated by subtracting the dose map in the ‘blank’ slice from the dose map in the ‘beam’ slice, as calculated by the treatment planning software.

EDR via light intensity was not required for any of the scans, although EDR via integration time was. For imaging parameters, 1800 projection angles were used for
Figure 6.2: Scanning protocol for the nPAG dosimeter. Slice 2 is scanned to act as $I_o$ for slice 1 (for the 6 Gy beam). Slice 4 acts as $I_o$ for slice 3 and slice 5 (for the 4 Gy and 2 Gy beams, respectively). Due to the fact that scatter dose distributions are present in the $I_o$ slices, one expects to reconstruct the dose differences between the paired slices. Expected dosage maps for the 6 Gy, 4 Gy, and 2 Gy slices are shown above. These were calculated: $[\text{slice 1} - \text{slice 2}]$ for the 6 Gy beam, $[\text{slice 3} - \text{slice 4}]$ for the 4 Gy beam, and $[\text{slice 5} - \text{slice 4}]$ for the 2 Gy beam. Note that the scattered dosages in slices 2 and 4 are relatively low. Only the expected dose map for the 2 Gy beam shows a noticeable dip. This is due to scattered dose in slice 4 that results from the 4 Gy beam.
each scan, and 1 sample per projection was acquired. Slice 4 (illustrated in Fig. 6.2) was collected twice—once for each of the 2 Gy and 4 Gy beams.

6.2 Initial Observations

A few observations may be made about the reconstructed images before comparing them against their expected dose maps. An unfiltered (i.e. no sinogram processing or post-reconstruction processing) image of the 4 Gy slice is shown in Figure 6.3. This image displays three prominent issues that are apparent in all of the reconstructions—streaking, ringing, and particulates. Unlike in scans of scatter-based solution, the majority of streaking artefacts seen in images of the gel dosimeter do not correspond to the seams of the container walls. The streaks correspond to the outer portions of the gelatin. Ring artefacts due to array-edge data corruption are expectedly present in the reconstructions. Lastly, particulates—dust and other impurities—are revealed in the images. Whether each particulate results in a peak or a dip in the image depends on whether it was present in the I or Io slices, respectively.

The new sources of streaking can be observed in the sinogram scans of the gelatin. Figure 6.4 shows both of the I scans that were scanned at slice 4 of the scanning protocol (see Fig. 6.2). Woodgrain-like striations in the sinograms correspond to the positions of the streaking observed in these reconstructions. The patterns that these striations exhibit in sinogram space do not represent physically reconstructable objects. So, the current hypothesis charges these striations as being refractive inhomogeneities in the gelatin matrix. The fact that these striations are reproducible, in that they repeat in both sinograms, lends to the belief that these are physically constant attributes of the dosimeters. Potentially, these streak-causing striations and particulates could be addressed by acquiring pre- and post-irradiation scans at the same slice of the dosimeter.
Figure 6.3: Here, an unfiltered sample image is shown of the 4 Gy beam. Zoomed regions show streaking, ring artefacts, and particulates in the gelatin. Much of the streaking here does not correspond to the flask seams. Ringing due to array-edge data corruption remains present. Lastly, errors due to particulates in the gelatin can arise from particulates in either of the $I_o$ or $I$ slices.
Figure 6.4: Slice 4 of the nPAG dosimeter was scanned twice—once for each acquisition of the 2 Gy and 4 Gy slices. Both scans are shown above. Woodgrain-like striations are apparent between raylines 65→105 and 225→265. The persistence of these striations indicates refractive inhomogeneities in the gelatin. By comparing different slices of the gelatin for $I_o$ and $I_I$, different striation patterns compound to create considerable streaking in the image reconstructions. Also noteworthy is the sinogram path of a particulate that starts near ray 147.

6.3 Comparison Between Measured & Expected Values

Expected dose maps were calculated using dose distributions provided by Eclipse$^{TM}$ Treatment Planning System (Anisotropic Analytical Algorithm version 8.615, 1 mm$^3$ dose grid). The unfiltered images and expected dose maps for all three beams are displayed in Figure 6.5. The expected dose maps were scaled and aligned with measured dosages by visually matching the curved walls of the flask container. Aside from the three features described above, the main trend to be seen in these comparisons is the relative level of noise experienced by the three different dosages. The lowest of the three, 2 Gy, suffers greatly from streaking caused by the gel-based striations. These streaks are present for all three images, though their presence becomes less imposing as dose (i.e. signal) is increased.

To facilitate a comparison between measured values and expected values, images were filtered. To filter images, absorbance sinograms were filtered once using an adaptive wiener filter with a 5×5 pixel$^2$ filter window. Sinograms were then used
Figure 6.5: Unfiltered reconstructions and expected dose maps: 2 Gy slice in (a) and (b), 4 Gy slice in (c) and (d), and 6 Gy slice in (e) and (f).
Figure 6.6: Left: filtered reconstructions with overlying expected dose contours corresponding to 95%, 70%, 50%, and 10% of the max dose. Right: PDD curve comparisons between expected (plotted in black and white) and the 40 central PDDs from the filtered reconstructions (dose scale in cGy, beam depth indicated). These are shown for the 2 Gy beam in (a), 4 Gy beam in (b), and 6 Gy slice in (c). Note that the reconstructed data uses a common normalization constant for all three beams in the PDD comparisons.
for reconstruction and these reconstructions were filtered again using the same filtering method (i.e. wiener, 5×5). More sophisticated and well-purposed methods of data filtering will have to be investigated in the future. Yet, for these preliminary comparisons, this method will suffice.

Filtered images for all three beams are displayed in Figure 6.6 with overlying contours from the expected dose maps corresponding to 95%, 70%, 50%, and 10% of the maximum dose for each expected dose map. Alongside these images are percent-depth dose (PDD) curve comparisons for all three beams. The measured values for the 40 central PDDs for each beam are plotted in colour together with the expected values, which are plotted in black and white. It should be noted that the same pixel-to-dose normalization factor was used with all three sets of measured PDDs. That is, they were not individually normalized to be level with their respective dose maps. This normalization factor was determined based simply on visual alignment with the expected PDD data.

Overall, there is good agreement between measured and expected values. The majority of pixels along the PDDs for each beam follow a similar curve as the expected PDDs. Upon scrutinous comparison, one will notice that where the measured data deviates greatly from the expected values corresponds to imaging issues that were described in Section 6.2. Large variations near the peak of the 2 Gy PDD are due to severe streaking in that region. A tremendous dip near pixel 170 of the 2 Gy PDD curve is caused by a particulate that can be seen in the reconstruction. Also, while ringing is apparent in all three images, the horns are most prominently shown in the PDDs of the 6 Gy beam.

6.4 Current Imaging Issues

Following the examination of the irradiated nPAG dosimeter, three key issues are found to be undermining accurate reconstruction. These are: (i) streaking due to
refractive inhomogeneities in the gelatin, (ii) ring artefacts due to array-edge data corruption, and (iii) particulates in the gelatin. In addition to these three prominent issues, one would expect that streaking due to variations in flask surface quality remains a dormant issue that is not as imposing, relatively. One could find that flask surface quality would become an issue again once gel striations are addressed.

Two main steps may be taken to help address these issues. First, data corruption that occurs at the edges of each photodiode array needs to be better addressed. This will be addressed via a future hardware update that will replace the op-amps that are believed to be causing the corruption. Alternatively, the offset correction could be revisited. It is possible that improving offset correction could reduce the appearance of rings in final images. Another option would be to devise a sinogram treatment method that specifically targets these problem areas, however this could cause unwanted blurring in the corresponding radii of images.

The second step to address key issues is to develop a technique that allows pre- and post-irradiation scans to be acquired and registered so that both sets of data are mutually aligned in vertical and angular position. This would address both types of streaking since both flask surface quality and gel striations seem to be physically constant variables, which would be reproducible in pre- and post- scans. As well, particulates present in the gel would be accommodated by a pre- and post- scan.

Ideally, both particulates and gel refractive inhomogeneities would be prevented. To prevent particulates, better care could be taken to ensure that all equipment used in gel fabrication is clean. Also, impurities may be caused due to impurities in the ingredients themselves, so a better inspection of these ingredients warrants investigation. To prevent gel refractive inhomogeneities, better gelation of the gelatin matrix would be necessary. It is possible that a more controlled method of cooling could limit or even eliminate the inhomogeneities seen here.
7.1 Conclusions

The current state of a prototype fan-beam optical CT scanner has been presented. Its intent and purpose is to evaluate three-dimensional radiation dosimeters that exhibit increases in optical density when exposed to ionizing radiation. This work described a number of essential developments that contribute to both the capability and performance of the device.

A new method of light intensity modulation for the system was introduced. Using a pair of linear polarizers, the system is now capable of fine-adjustments in light intensity with minimal disruption of the alignment of the system. As well, this method of modulation provides a considerable range of relative intensities (1→10,180). To be certain that polarization changes that come with this method of modulation does not affect detected values, a test of the polarization sensitivity of the detector system was performed. Effectively, fluctuations due to polarization angle of light were found to be negligible (<1%).

The detector system as a whole was characterized with respect to both light intensity and the programmable integration time of the photodiode arrays. Observations found during this characterization revealed the need for an offset correction that must be applied to values detected by the system. The use of this correction significantly
improves the system’s ability to reliably measure relative changes in light intensity without exhibiting the shape of the 5 photodiode arrays used for detection. However, data corruption that persists near the edges of photodiode arrays continues to be an issue. This leads to distinct ring artefacts in reconstructed images.

Unfortunately, the need for a positive offset correction affects the dynamic range of the system. In order to perceive high levels of absorbance ($A > 1.25$) two methods to extend the dynamic range of the system were introduced. The first method uses longer photodiode integration times, allowing values that were otherwise ‘dark’ to be detected. The second method uses higher levels of light intensity—modulation allowed by the polarizer pair—to collect valid data. Together, these methods may be combined to significantly improve the dynamic range of the system.

Examinations of highly attenuating absorption-based and scatter-based solutions was made possible using a combination of both methods of dynamic range extension. This allowed for comparisons to be made between the relative effectiveness of the multi-hole (MH) and single-slot (SS) collimators for scatter-rejection. The MH collimator was found to accurately measure absorbance values of highly attenuating samples of both attenuator types ($A \sim 4$). This range of values was limited only by the maximum dynamic range of the system. The SS collimator experienced slight signal contamination with absorption-based samples and considerable signal contamination with scatter-based samples. Its range of accuracy was not limited by the maximum dynamic range of the system. Signal contaminations experienced by the SS collimator cause the understatement of absorbance values, which lead to cupping artefacts in image reconstructions. A spiking artefact was also observed with highly attenuating samples using the SS collimator. The cause of this artefact is currently unclear. These comparisons affirm the worth of the MH collimator, which is now the default collimator used with the system.

A number of investigations into imaging protocol have provided valuable informa-
tion for improving the quality of images obtained by the system. First, by obtaining the scan of a ‘blank’ to use for $I_o$, ring artefacts that previously plagued the system have been eliminated. These were evidently caused by reflection and/or the presence of dosimeter container walls. Second, examinations of container surfaces have revealed significant variations in surface quality that cause considerable amounts of noise in image reconstructions if not properly addressed. These variations may be addressed by comparing the same surfaces for both $I_o$ and $I$ scans. This requires a projection-by-projection comparison to be made through sinograms of the same physical slice for $I_o$ and $I$. Third, the stability of both the fan-beam and the detection system was supported by temporal variation and sampling accuracy tests. While it was shown that the shortest delay between the acquisitions of $I_o$ and $I$ is ideal, penalties paid by delaying $I$ scans are minor. Also, tests that acquired and averaged multiple projection samples for each projection angle show that improvements in image quality due to multiple sampling are insignificant. Finally, tests that acquired varying numbers of projections per rotation show that improvements in image quality may be attained by increasing number of projections. However, improvements seen with greater than 360 projections per $360^\circ$ rotation were minor.

An nPAG dosimeter was prepared, irradiated, and evaluated using the system. Images reconstructed from slices of the dosimeter exhibit three main artefacts that undermine their accuracy: (i) streaking artefacts caused by refractive inhomogeneities that are present in the gelatin of the dosimeter, (ii) peak and dip artefacts caused by particulate impurities that are also present in the gelatin of the dosimeter, and (iii) ringing artefacts caused by data corruption which occurs at the edges of the individual photodiode arrays. Comparisons between measured data and expected data indicate that accurate results may be obtainable if the above three artefacts can be prevented or properly corrected for.

In conclusion, a number of key investigations have been performed and presented
in the current work. These are promising developments, suggesting that high quality results may be obtained in the future by properly addressing current issues. Primarily, these issues are (i) hardware-based data corruption near photodiode array edges, and (ii) the need to obtain $I_o$ and $I$ data from the same slice on a projection-by-projection basis. Once these issues are addressed, priorities may begin shifting towards improving acquisition efficiency and benchmarking the system against other optical CT systems and other methods of dosimetry.

7.2 Priorities for Future Work

There are a myriad of aspects of the current system that call for attention as development continues. The first of which is the need for pre- and post-irradiation scans of dosimeters. By ensuring that the same portions of the dosimeter are being compared on a projection-by-projection basis before and after it is irradiated, a number of imaging artefacts may be avoided or significantly subdued. These are artefacts related to variations in flask surface quality, gelatin refractive inhomogeneities, and gelatin impurities. To achieve this goal, a method of properly registering the position of the dosimeter for both scans will have to be developed.

Modifications may be made to improve the data collection of the system. First and foremost, the data corruption which occurs at the edges of photodiode arrays must be addressed. By installing faster operational amplifiers that are capable of operating at current speeds, ring artefacts may be prevented without the need to slow down the system. Also, the onboard memory for the current circuit board could be improved. With the current hardware, memory may potentially be doubled by using memory more efficiently. This would require an update of the current firmware. Lastly, data transfer speeds may be increased significantly by writing a custom USB driver for communication between the circuit board and the PC.

A number of hardware updates may also be made to other components of the
scanner. For dosimeter motion, the scanner would benefit from a rotational stage that can reliably operate at high angular velocities. Yet, this should be considered after the data collection of the system has been optimized, as considerations with angular velocity are intertwined with the speed at which light projections can be collected. As well, a new multi-hole collimator that can accommodate the 128-element photodiode arrays could be developed. The benefit of physical collimation has shown it to be a valuable asset of the system. If a collimator could be created for a 640-element detector array, this could lead to admirable scanner resolution. Additionally, a mechanized stage for the first linear polarizer could improve the speed of the system. A mechanized stage could allow for fast and automated light intensity modulation and improved light accuracy. Lastly, although it has been shown that the shape of the fan-beam is not a vital aspect in optical CT, better methods of fan creation could be investigated. These investigations could examine line-generating lenses with higher wedge-angles or, alternatively, cylindrical lenses.

Significant consideration must also be given to the relationship between sinogram space and image reconstruction space. While the current work focused primarily on which data was acquired, future work will have to exploit opportunities to improve the processing of data once it has been acquired. These may be done by developing methods of sinogram filtering that are aimed at targeting specific problems (i.e. schlieren and particulates in the bath, light intensity fluctuations, ring artefacts). Also, a method could be developed in sinogram space that removes the seams of container walls, which can cause problematic streaking artefacts. Lastly, improved methods of image reconstruction will have to be considered. One could explore back-projection filters designed specifically for optical CT applications, as well as methods of masking and/or padding sinograms before reconstruction.

Once a number of dominant system issues have been dealt with, a set of tests will have to be performed to benchmark the current system with other optical CT sys-
tems and other methods of dosimetry. Phantoms can be constructed with needles of known diameters to evaluate the spatial resolution of the system. Simple treatment plans with stepped dosage levels can be designed to evaluate the dosimetric accuracy and precision of the system. This accuracy and precision will be unavoidably intertwined with the dosimeter chosen. Therefore, other gel dosimeter recipes and other dosimeter types may be need to be considered. Lastly, a novel irradiation could be designed to demonstrate the imaging capabilities of the system. This could provide interesting dose map comparisons between reconstructed dose distributions and dose distributions provided by treatment planning software. These may also be compared with another dosimetry method (e.g. film) to provide relatable comparisons for researchers who are not familiar with polymer dosimetry.

In summary, current developments are promising and the priorities for future research are clear. Yet, these priorities branch into two areas. The first of these areas involves establishing the system’s presence in the area of 3D dosimetry. Indications suggest that the fan-beam optical CT scanner stands to potentially establish itself as the world’s fastest system available for accurately imaging scatter-based 3D dosimeters. Although it is specifically equipped to image scatter-based dosimeters, it is just as well equipped to image absorption-based dosimeters. The system could become a viable option for researchers looking to begin investigations with measurement-based 3D dosimetry. The second area of priorities for the system involves actually using the system for 3D dosimetry research. Ultimately, the purpose of the scanner is to obtain measurements of 3D dose distributions. These may be used for the verification of complex radiation delivery methods, the evaluation of new methods as they are developed, and any implementation where high-resolution 3D dosimetry would be valuable.
Bibliography


