RICE UNIVERSITY

NOVEL METHODS FOR REDUCING BREAST DOSE DURING COMPUTED TOMOGRAPHY SCANS

by

Kelsey Boitnott Mathieu

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APPROVED, THESIS COMMITTEE

Dianna Cody, Professor of Imaging Physics, The University of Texas MD Anderson Cancer Center

Dennis Cox, Professor of Statistics

Michael McNitt-Gray, Professor of Radiology, David Geffen School of Medicine at UCLA

Antonios Mikos, Louis Calder Professor of Bioengineering

Rebecca Richards-Kortum, Stanley C. Moore Professor of Bioengineering

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ABSTRACT

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Pediatric female and young adult female patients who undergo computed tomography (CT) scanning may be at higher risk for developing radiation-induced breast cancer later in life. Thus, the purpose of this thesis was to both accurately quantify dose and explore new strategies for CT breast dose reduction. In order to determine dose reduction, dose quantification was first assessed through the development and validation of an empirical model for describing attenuation in CT and second through evaluation of the precision of dosimetry-related measurements obtained using three different models of CT scanners. Breast dose-savings was evaluated using CT dose index phantoms, anthropomorphic phantoms, and Monte Carlo computer modeling. Modifications to current scanning procedures, such as proper patient centering and beginning data acquisition with the x-ray tube facing a patient’s posterior, were shown to minimize breast dose. Novel techniques, including varying the x-ray tube voltage during scanning and incorporation of a dynamic x-ray beam filter over the breasts, were also found to successfully reduce breast dose.
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To my parents for teaching me “where there’s a will there’s a way”… perhaps too well

To my dogs, Link and Lady, for providing comic relief and keeping me company during my all-nighters

Finally, to my brother, Josh, a cancer survivor, who makes me believe in the importance of this research
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<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Definition</th>
</tr>
</thead>
<tbody>
<tr>
<td>AIC</td>
<td>Akaike’s information criterion</td>
</tr>
<tr>
<td>ALARA</td>
<td>As low as reasonably achievable</td>
</tr>
<tr>
<td>ASIR</td>
<td>Adaptive iterative statistical iterative reconstruction</td>
</tr>
<tr>
<td>CT</td>
<td>Computed tomography</td>
</tr>
<tr>
<td>CTDI</td>
<td>Computed tomography dose index</td>
</tr>
<tr>
<td>CTDI&lt;sub&gt;100&lt;/sub&gt;</td>
<td>CTDI calculated from exposure detected by a 100-mm pencil ionization chamber</td>
</tr>
<tr>
<td>CTDI&lt;sub&gt;100, 9:00&lt;/sub&gt;</td>
<td>CTDI&lt;sub&gt;100&lt;/sub&gt; detected at the 9:00 peripheral chamber hole in a CTDI phantom</td>
</tr>
<tr>
<td>CTDI&lt;sub&gt;100, 12:00&lt;/sub&gt;</td>
<td>CTDI&lt;sub&gt;100&lt;/sub&gt; detected at the 12:00 peripheral chamber hole in a CTDI phantom</td>
</tr>
<tr>
<td>CTDI&lt;sub&gt;100, air&lt;/sub&gt;</td>
<td>Free-in-air CTDI</td>
</tr>
<tr>
<td>CTDI&lt;sub&gt;100, central&lt;/sub&gt;</td>
<td>CTDI&lt;sub&gt;100&lt;/sub&gt; detected at the central chamber hole in a CTDI phantom</td>
</tr>
<tr>
<td>CTDI&lt;sub&gt;100, peripheral&lt;/sub&gt;</td>
<td>CTDI&lt;sub&gt;100&lt;/sub&gt; detected at one of the peripheral chamber holes in a CTDI phantom</td>
</tr>
<tr>
<td>CTDI&lt;sub&gt;W&lt;/sub&gt;</td>
<td>Weighted CTDI</td>
</tr>
<tr>
<td>CV</td>
<td>Coefficient of variation</td>
</tr>
<tr>
<td>EM</td>
<td>Electromagnetic</td>
</tr>
<tr>
<td>FDA</td>
<td>Food and Drug Administration</td>
</tr>
<tr>
<td>FOV</td>
<td>Field of view</td>
</tr>
<tr>
<td>GE</td>
<td>General Electric</td>
</tr>
<tr>
<td>Gy</td>
<td>Gray</td>
</tr>
<tr>
<td>HU</td>
<td>Hounsfield unit</td>
</tr>
<tr>
<td>HVL</td>
<td>Half-value layer</td>
</tr>
<tr>
<td>ICRP</td>
<td>International Commission of Radiological Protection</td>
</tr>
<tr>
<td>ISO</td>
<td>International Organization for Standardization</td>
</tr>
<tr>
<td>keV</td>
<td>Kiloelectron volts</td>
</tr>
<tr>
<td>kVp</td>
<td>Peak kilovoltage</td>
</tr>
<tr>
<td>LNT</td>
<td>Linear no-threshold</td>
</tr>
<tr>
<td>mA</td>
<td>Milliamperes; units of x-ray tube current</td>
</tr>
<tr>
<td>mA·s</td>
<td>Current-exposure time product</td>
</tr>
<tr>
<td>MC</td>
<td>Monte Carlo</td>
</tr>
<tr>
<td>MDCT</td>
<td>Multidetector CT</td>
</tr>
<tr>
<td>MIRD</td>
<td>Medical Internal Radiation Dose</td>
</tr>
<tr>
<td>MRI</td>
<td>Magnetic resonance imaging</td>
</tr>
<tr>
<td>ms</td>
<td>Millisecond</td>
</tr>
<tr>
<td>NI</td>
<td>Noise index</td>
</tr>
<tr>
<td>PMMA</td>
<td>Polymethylmethacrylate</td>
</tr>
<tr>
<td>QA</td>
<td>Quality assurance</td>
</tr>
<tr>
<td>QVL</td>
<td>Quarter-value layer</td>
</tr>
<tr>
<td>R</td>
<td>Roentgen</td>
</tr>
<tr>
<td>R&lt;sup&gt;2&lt;/sup&gt;</td>
<td>Coefficient of determination</td>
</tr>
<tr>
<td>Abbreviation</td>
<td>Definition</td>
</tr>
<tr>
<td>--------------</td>
<td>------------------------------------------------</td>
</tr>
<tr>
<td>ROI</td>
<td>Region of interest</td>
</tr>
<tr>
<td>RPD</td>
<td>Relative percent difference</td>
</tr>
<tr>
<td>s</td>
<td>Seconds; units of x-ray tube rotation time or exposure time</td>
</tr>
<tr>
<td>SDCT</td>
<td>Single detector CT</td>
</tr>
<tr>
<td>SFOV</td>
<td>Scan field of view</td>
</tr>
<tr>
<td>SSE</td>
<td>Sum of squares error</td>
</tr>
<tr>
<td>Sv</td>
<td>Sievert</td>
</tr>
<tr>
<td>TCM</td>
<td>Tube current modulation</td>
</tr>
<tr>
<td>TLD</td>
<td>Thermoluminescent dosimeter</td>
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</tbody>
</table>
1.1. INTRODUCTION

The number of computed tomography (CT) scans performed annually in the United States grew from 3 million in 1980 to 62 million by 2006.\(^1\) Based on an estimated rate of 10% increase per year, approximately 100 million CT scans will be performed in the U.S. in 2011.\(^2\) For certain clinical applications, CT use has increased at an even faster rate; for instance, the number of annual CT scans performed in the emergency department reportedly increased by approximately 16% per year between 1995 and 2007.\(^3\) Additionally, the percentage of CT scans performed on pediatric patients (15 years or younger) increased from 4% of all CT scans in 1989\(^4\) to 11% by 1999.\(^5\) Use of CT has continually increased because CT images have notably improved soft tissue contrast compared to conventional x-ray images\(^6\) and enhanced spatial resolution compared to magnetic resonance imaging (MRI);\(^7\) increased speed of imaging provides another advantage of CT over MRI. Because CT imaging can depict both soft tissue and bone details within seconds, it has broad clinical applications, ranging from cancer diagnosis and treatment planning to assessing emergency room trauma patients.\(^8\)

Although CT scans are being prescribed at an increasing rate, until recently the potential risk of radiation-induced cancer associated with these scans was
given little attention. According to a 2004 study, only 9% of emergency department physicians and 47% of radiologists surveyed at one hospital believed that there was an increased risk of cancer from CT.\textsuperscript{9} Furthermore, only 22% of the emergency department physicians and 15% of the radiologists were able to accurately quantify the amount of radiation dose estimated for the hospital's abdominopelvic CT protocols relative to the dose received from a chest radiograph. Awareness of the potential risks of exposure to medical radiation among physicians and the general public alike has increased with media coverage of recent incidents across the United States that have resulted in hundreds of patients receiving excessive levels of medical radiation, causing radiation burns and hair loss.\textsuperscript{10, 11} While guidelines and regulations exist for monitoring radiation exposure to nuclear industry workers and healthcare providers, such regulations do not exist for exposed patients. A study that reviewed the health insurance records of almost 1 million Americans estimated that over 1800 of the study participants received very high levels of radiation, which were estimated to exceed the annual limit for nuclear power plant workers.\textsuperscript{12} Although CT use is currently not federally regulated and instead voluntary programs provide the standards of care (e.g., accreditation though the American College of Radiology), recent overdose incidents have prompted the Food and Drug Administration (FDA) to launch an initiative aimed at preventing unjustified CT examinations and unnecessarily high levels of radiation exposure.\textsuperscript{13} Upon investigation of radiation overdoses during CT brain perfusion examinations, the FDA discovered that improper use of CT scanners, rather than
equipment malfunction, was responsible for the incidents; these findings were summarized in a press release issued in November of 2010.\textsuperscript{14} The press release also stated that the FDA is currently working with CT scanner manufacturers to make scanners safer by adding alerts to warn the CT technologists performing the examinations, many of whom are inadequately trained on radiation protection,\textsuperscript{15} when the prescribed dose is considered to be too high.\textsuperscript{16}

Although the exact cancer risk to patients undergoing CT examinations is uncertain, it is the responsibility of medical professionals to minimize this risk by preventing overdose incidents and by developing and implementing strategies to lower patient doses. Furthermore, when patient dose is of concern (e.g., when scanning pregnant women or young children), non-ionizing imaging modalities, such as MRI or ultrasound, should be considered as alternatives to CT.\textsuperscript{7} A 2009 study suggested that 3 - 77\% of CT scans performed on young patients (under 35 years) were unjustified or could be replaced by a non-ionizing imaging modality.\textsuperscript{17} While there is uncertainty regarding cancer risks associated with CT, medical professionals agree that it is important to consider radiation exposure to individual tissues given that carcinogenesis is often organ-specific. Many epidemiological studies have associated exposure to ionizing radiation, including medical sources of radiation (e.g., CT), with increased breast cancer risk.\textsuperscript{18, 19} Recent data indicates that 1 out of every 8 American women will be diagnosed with breast cancer during her lifetime, making it the most common cancer in women (excluding nonmelanoma skin cancer).\textsuperscript{20, 21} The risk of developing radiation-induced breast cancer is especially high when radiation exposure
occurs during childhood.\textsuperscript{18, 22} Although most CT scans are performed on adults, children with cancer often receive high levels of radiation exposure from recurrent CT scans, as well as radiation therapy. Such exposure incurs a lifelong risk to these patients; in fact, statistically significant rates of secondary cancer have been reported throughout the lifespan of childhood cancer survivors.\textsuperscript{23, 24} Although childhood cancer is rare (<1% of cancer diagnosis), an estimated 10,700 new cases of childhood cancer (among children 0 – 14 years old) were expected in 2010.\textsuperscript{21} Furthermore, the percentage of childhood cancer survivors is increasing, and based on the 5-year survival rate of 77.9% for these patients,\textsuperscript{21, 25} over 8,000 children a year may be at risk for secondary cancer; thus, it is important to consider levels of radiation dose to this patient population.

The combination of high breast cancer incidence, radiation sensitivity of breast tissue (especially in pediatric cancer patients), and increasing use of CT provides motivation for development of breast dose reduction strategies. Thus, the goal of this thesis was to develop new strategies for reducing breast dose to pediatric and young adult female patients during CT scanning.

1.2. FUNDAMENTALS OF COMPUTED TOMOGRAPHY

The invention of computers allowed for the development of computed tomography – a medical imaging technology capable of producing a three-dimensional view of the human body’s internal anatomy.\textsuperscript{8} In 1971, the first clinical
CT scanner (developed by Godfrey Newbold Hounsfield) was installed at Atkinson-Morley's Hospital in London and its clinical potential was immediately apparent. Initially, CT was used to image the brain by passing a pencil-like beam of x-rays through the head from multiple angles around its periphery. The x-rays that were transmitted through the head were measured on the other side by a detector and relayed to a computer which reconstructed a cross-sectional image of the scanned region.\textsuperscript{6} Within 10 years of introduction into clinical practice, CT use rapidly increased and many technological advances were made, including development of a body scanner.\textsuperscript{7} Furthermore, the number of images acquired during modern scanning has increased dramatically from early CT examinations during which only 20 – 50 images were acquired.\textsuperscript{26} While modern CT scanners differ greatly from the brain scanner initially developed by G.N. Hounsfield, the basic principles of operation remain the same.

1.2.1. X-RAY PRODUCTION

In Hounsfield's original prototype, the radioactive isotope Americium 95, which emits gamma rays, was used. Both x-rays and gamma rays are types of electromagnetic (EM) radiation, or energy that travels through space or matter, and can be described both as waves and as particles. When considered to behave as a wave, EM radiation is characterized in terms of wavelength and frequency and appears as such in the EM spectrum. Gamma rays and x-rays are both on the high-frequency end of the spectrum and thus are ionizing. Ionizing radiation interacts with matter by removing bound electrons from the electron
shell of an atom. From a particle perspective, EM radiation behaves as photons, or quanta of energy. Although gamma rays and x-rays are both forms of EM radiation, they are produced differently. X-rays are produced outside the nucleus, while gamma rays are produced within the nucleus of a radioactive atom. Therefore, production of x-rays can be controlled, while gamma ray production is dictated by the decay time of the radioactive isotope emitting the gamma radiation (e.g., Americium 95). Because the time required to generate enough gamma rays to form an image resulted in extremely long scan times, Hounsfield soon replaced the isotope source with an x-ray tube to generate ionizing radiation in CT. 7

X-rays are produced inside an x-ray tube by energy conversion. A fast-moving stream of electrons is generated at the filament, which is part of the cathode, and is directed towards the anode by applying a high potential difference. When an electron in the stream interacts with a specific target area on the anode, its kinetic energy is either lost in the form of heat (>99% of interactions) or in photon generation. Because such a large amount of heat is generated during this process, rotating anodes are employed to minimize heat loading by allowing for heat distribution over a larger target area. Additionally, the anode target is typically made of tungsten because it has a high atomic number, which allows for efficient x-ray conversion, and a high melting point. 6 Figure 1.1 shows the basic components of the x-ray tubes used in CT.
FIGURE 1.1. Basic components of an x-ray tube. In CT, the x-ray beam emitted from the tube is filtered before exposing patients.

Two types of interaction are responsible for x-ray generation in an x-ray tube: bremsstrahlung and characteristic x-rays. Bremsstrahlung radiation constitutes the majority of the x-ray beam and is produced when an electron comes in the proximity of the nucleus of a tungsten atom on the anode target region. Electrical attractions between the electron and the positively-charged tungsten nucleus pull the electron off course, causing it to decelerate. A photon, whose energy corresponds to the kinetic energy lost by the electron during deceleration, is emitted; the amount of kinetic energy lost depends on the distance between the electron and the nucleus. When an electron collides with
the nucleus, which occurs only rarely, all of its kinetic energy is transferred to the emitted photon. Characteristic x-rays are a result of the photoelectric effect. The photoelectric effect occurs when an electron in the stream strikes an inner-shell electron of the tungsten atoms that compose the anode target, uses some of its energy to remove the electron from the electron shell, and then transfers the remainder of its energy to the ejected electron. When this happens, there is a vacancy in an inner-electron shell, which the atom attempts to fill through an electron cascade that moves electrons from outer to inner shells. Energy equal to the difference in the binding energies of the inner and outer shells is released during the cascade as a characteristic x-ray or is transferred to an Auger electron, which is an outer-shell electron that is ejected from the atom but does not produce x-rays.

The amount of current that is allowed to flow through the filament is measured in units of milliamperes (mA) and determines the number of electrons that flow from the cathode to the anode. Therefore, the x-ray tube current also determines the intensity (quantity) of x-rays produced. The exposure time (typically expressed in seconds [s]) determines how long current is flowing and x-rays are being produced. Energy (quality) of the x-rays is controlled by the potential difference across the x-ray tube, which pulsates between a minimum and maximum voltage (known as voltage ripple); the maximum voltage is called peak kilovoltage (kVp). Instantaneous voltage at the time of flow dictates the kinetic energy of each electron and is expressed in units of kiloelectron volts (keV). Varying kinetic energies of electrons in the electron stream, along with
differences in the amounts of energy transferred from the electrons to the emitted photons, causes the x-ray beam to be composed of photon of a spectrum of energies. Spikes from the contribution of characteristic x-rays appear at specific keVs in the spectrum, which are related to the binding energies of electrons in the shells of tungsten atoms. The energies of photons in the spectrum range from approximately 0 keV (in an unfiltered spectrum) to the kVp. Filtration increases the minimum energy and narrows the range of energies in the spectrum by preferentially removing the low-energy photons. CT scanners have a constant amount of inherent filtration, which is shown in Figure 1.1. An example of an unfiltered spectrum and the same spectrum after passing through filtration appears in Figure 1.2.

![Diagram showing the unfiltered and filtered x-ray spectrum](image)

**FIGURE 1.2.** The unfiltered spectrum of a 100 kVp x-ray beam and the same spectrum after applying filtration.
1.2.2. X-RAY INTERACTIONS IN CT

Interactions of the x-ray beam produced by a CT scanner’s x-ray tube as it passes through a patient are necessary for producing CT images. When traversing matter (e.g., the human body), x-ray photons will either penetrate through the matter unchanged or interact with the matter. Whether or not a photon is able to ionize matter is determined by both the photon’s energy and matter’s physical properties (e.g., density and atomic number).\(^8\)

Attenuation is the interaction and consequent reduction in intensity of an x-ray beam as it passes through matter. Before an x-ray beam passes through a patient, it is attenuated as it travels through the x-ray tube’s output port and beam filtration, which affects both the number and energy of photons in the beam. Attenuated photons are removed from the primary x-ray beam by either absorption or scattering. In the case of absorption, interacting photons lose all of their energy and are stopped inside the matter. Alternatively, scattered photons are deflected off their original course during the interaction and may lose some of their energy as a result.

In diagnostic CT imaging, three primary attenuation interactions occur between the x-ray beam and the human body: coherent scattering, Compton scattering, and the photoelectric effect, which was described in 1.2.1. Coherent (Rayleigh) scattering is the only non-ionizing interaction and occurs when a photon is scattered but its energy does not change. In Compton (incoherent) scattering, a photon is deflected after striking and ejecting an outer-shell electron.
from an atom; some of the photon’s original energy is transferred to the ejected electron, which then interacts with surrounding atoms. Of these three interactions, the contribution of coherent scattering is often less than 5%. The relative contributions of Compton scattering and the photoelectric effect vary based on photon energy and physical properties of the interacting matter. However, in the energy range used for diagnostic CT, Compton scattering is the dominant interaction in both soft tissues and bone.6

1.2.3. X-RAY DETECTION

X-rays that penetrate through the patient and reach an array of detectors (mounted directly across from the x-ray tube) are recorded and used to create CT images. The number and energy of photons varies across detectors in the detector array according to differences in attenuation in the anatomical regions that different photons pass through; different tissues in the body interact with x-rays differently. For example, bone readily attenuates x-rays and, as a result, few photons reach the detector; on the other hand, air in the lungs attenuates very few x-rays and thus many photons are transmitted and detected.

Modern scanners incorporate multiple rows of detectors placed side-by-side. This is referred to as multidetector CT (MDCT) and can be contrasted with single detector CT (SDCT), which features a single row of detectors. In SDCT, only one image can be reconstructed for each rotation of the x-ray tube; on the other hand, MDCT allows for multiple planar sections of a patient to be acquired concurrently. Figure 1.3 highlights the difference between SDCT and MDCT. The
advantages of MDCT over SDCT are that there are more options for the slice (image) thickness and more anatomy can be imaged during each rotation, which correlates to faster imaging times, particularly when acquiring submillimeter images, allowing small anatomic details to be resolved.27

![Diagram showing single detector array vs. multiple detector array](image)

**FIGURE 1.3.** SDCT only allows for acquisition of one image per rotation, and thus has been replaced by MDCT, which allows multiple images to be captured per rotation. The MDCT scenario shown here has four rows of detectors, which translates to reconstruction of up to four images per rotation.

The slice capability of MDCT scanners (i.e., the number of detector rows and the nominal beam width) is continuously increasing. Currently, 64-slice MDCT scanners are able to capture a 40-mm thick region of a patient in one rotation.28 The region of coverage (e.g., 40 mm) is called the nominal x-ray beam width and equals the number of images acquired multiplied by the thickness of
Because each row in the detector array in MDCT can acquire data for (at most) a single image, the thickness of each image is dependent on the width of each detector row. The number of images that can be simultaneously captured depends on the total number of detector rows. Thus, a 40-mm coverage region for a 0.625-mm detector width can be subdivided into 64 0.625-mm images. Data acquired from multiple detector rows can be combined, either before or after scanning, to form images that are thicker than the width of individual rows. For example, 32 1.25-mm thick images can be reconstructed from 64 0.625-mm thick images by combining the raw data from pairs of neighboring detector rows. Likewise, 10 3.75-mm thick images can be generated by binning together groups of six rows of 0.625-mm detectors prior to acquisition. However, when detector rows are binned before scanning, they are treated as a single detector row; thus, the detected signal cannot be retrospectively separated across individual rows and reconstructing the data into thinner slices is impossible.

1.2.4. IMAGE RECONSTRUCTION

Data is collected by the detector array many times during each rotation around a patient, generating different projections, or views, of the patient. The data collected by the detectors is transferred to a computer for processing, and subsequently a cross-section of the patient is reconstructed into an image by filling in an empty matrix (illustrated in Fig.1.4) with the data. The profiles of photon detection across detector elements in the array are sent through the
image matrix in the opposite direction from which they were originally received by the detector array; this process is known as back projection. For each angular position of the x-ray tube where data was sampled, the corresponding profile is back projected into the image matrix; Figure 1.5 shows the profile across the detector array for two angular positions of the x-ray tube. Each transmission profile is back projected into the image matrix such that the value detected by each detector element is filled in across an entire row or column in the image matrix. Once profiles from every angular position that was sampled have been back projected into the matrix, an image can be resolved. Overlap between profiles occurs and different regions of the image matrix appear either dark or light according to the x-ray interactions that occurred at the corresponding anatomical location. For instance, bone appears white in CT images, while air appears black.

**Figure 1.4.** The cross-sectional anatomy of a patient is converted into an image by filling in an empty image matrix. The thickness of each image (shown in this figure as the z-dimension) is determined by the single or the binned detector row width.
**FIGURE 1.5.** An image is produced by detecting transmission of x-rays through a patient at many angular positions of the x-ray tube as it rotates with detector array around the patient. The detected transmission profiles are then back projected into an image matrix and the overlap in profiles eventually forms an image. Note that the shading in this figure is opposite of what would be observed in the actual CT image. In reality, fewer photons would reach the detectors when traveling through the patient than through air and the patient would consequently appear lighter/brighter than air in the CT image.

While two-dimensional images are displayed on the scanner console, the image matrix is actually three-dimensional. Therefore, the elements in the matrix that appear as two-dimensional pixels are, in reality, volume elements called voxels; the numerical value of each voxel is determined during back projection.
All voxels in the matrix have the same dimensions, where each voxel is square in the x- and y-dimensions and sized according to thickness of the cross-section imaged in the z-dimension (see Fig. 1.4). The size of the image matrix is called the field of view (FOV) and is defined as the voxel size in the x- or y-dimension multiplied by the number of voxels per row or column, respectively. Because the image matrix is square, the FOV is the same for both the x- and y-dimensions. The two-dimensional views shown in each image provide three-dimensional views of patient anatomy when multiple images are viewed together.\(^8\)

Voxels in the image matrix are assigned a numerical value according to their level of attenuation; tissues in the body attenuate x-rays to varying degrees depending on their density and atomic number. For each detector, the left side of the following equation is calculated:

\[-\ln \left( \frac{I}{I_0} \right) = \mu x, \quad (1.1)\]

where \(I\) is the intensity of photons transmitted through an attenuating material of thickness \(x\) and \(I_0\) is the intensity with no attenuating material as measured by a reference detector. Because \(x\) is the same for every voxel, it cancels out of the calculation, and thus the calculated values for \(\mu\), which are known as the linear coefficients, are filled into voxels in the image matrix. Figure 1.6(a) shows a projection through one row in the image matrix, Figure 1.6(b) shows how voxels in a 2 x 2 image matrix would be filled in with data from two projections. The calculated \(\mu\) value is scaled according to the following equation and displayed as
such in the patient images:

\[ CT(x,y) = 1000 \cdot \frac{\mu(x,y) - \mu_{\text{water}}}{\mu_{\text{water}}} \]  \hspace{1cm} (1.2)

where \( CT(x,y) \) is called the CT number and is the value assigned to each voxel in the CT image in Hounsfield units (HU), \( \mu(x,y) \) is the \( \mu \) value for the given voxel determined through back projection, and \( \mu_{\text{water}} \) is the linear attenuation coefficient of water.\(^8\)

\[ (b) \]

**FIGURE 1.6.** (a) Transmission through one row in an image matrix for one projection angle; (b) shows a 2x2 matrix using transmission profiles from two projections.

Images generated through back projection are blurry, and so the transmission profiles are filtered before being back projected; this is known as filtered back projection. Modern scanners offer several filtering algorithms, which are designed to enhance specific details of the image (e.g., bone or soft tissue). Besides using filtered back projection to reconstruct CT images, iterative
reconstruction methods have also been developed. Iterative reconstruction is capable of producing CT images with less noise (when scanning the same object for the same x-ray tube exposure) than those generated by filtered back projection; however, long reconstruction times have historically limited this method's use. Recently, more efficient iterative reconstruction techniques have been developed and various forms of iterative reconstruction (known by the trade names ASiR, iDose, IRIS, AIDR) have been implemented across CT scanner manufacturers. The strategy developed by General Electric (GE) Healthcare is called adaptive statistical iterative reconstruction (ASiR) and combines iterative reconstruction with filtered back projection to shorten reconstruction times while reducing image noise. Although, an unfamiliar, pixelated appearance has been reported by some radiologists in images reconstructed with ASiR, who are accustomed to seeing images reconstructed from pure filtered back projection, it has not been shown to affect the diagnostic quality of the CT images.30

1.2.5. MODERN CT

Modern CT scanning begins with patients being positioned (either head-first or feet-first) on a table inside the bore (opening) of the CT scanner gantry (framework), as shown in Figure 1.7. Patients are then centered by a CT technologist within the bore using a laser-based guidance system; the patient table position is fixed in the horizontal plane, but can be vertically adjusted. The CT technologist leaves the patient room and performs a localizer scan, which is similar to a low-resolution chest radiograph, to help plan the examination. After
the localizer is reviewed, if the CT technologist is dissatisfied with the patient's centering, the patient's position can be adjusted and additional localizers can be acquired. However, the process of centering and re-centering the patient is time-consuming and demands are placed on technologists to increase patient throughput. As a result, patients are often improperly centered. At one hospital, it was reported that 95% of patients undergoing chest and abdominal CT examinations were off-center by more than 5 mm.\textsuperscript{31}

\textbf{FIGURE 1.7.} The basic components of a modern CT scanner gantry. The reference axes and tube angles shown are used by GE Healthcare CT scanners, which were the primary scanners involved in the research described in this thesis.
After using the localizer to determine the patient table positions (along the z-axis) where the scan should start and end to capture the desired image length, the technologist enters the scan parameters (e.g., kVp, mA, detector configuration, etc.) on the scanner console. The patient is then scanned according to the parameters specified. Once the scan begins, the patient is exposed to x-rays as the x-ray tube rotates (in the x-y plane) around the patient’s body. Modern CT scanners use a fan beam of x-rays coupled to an array of detectors, rather than a pencil beam coupled to one or two detectors, as in Hounsfield’s original CT scanner. The fan beam of x-rays is produced by an x-ray tube mounted inside the CT scanner gantry. The detector array is mounted on the opposite side of the scanner gantry and rotates synchronously with the x-ray tube. Since portions of the fan beam outside the detection region do not contribute to the images, width of the fan beam is collimated at the output of the x-ray tube to match that of the detector array. The shape of the fan beam is also controlled using a bowtie-shaped filter (see Fig. 1.1). The bowtie filter is used in addition to inherent filtration to further filter the spectrum and shape the beam to reflect the cylindrical shape of most patients by preferentially filtering the outer edge of the fan beam which exposes the relatively thinner regions of the patients. Many scanner models offer several bowtie options which can be selected to best match individual patient size.

To image different sections of the patient’s anatomy, the patient table is moved along the z-axis through the scanner bore. In early CT, the x-ray tube rotated around one section of the patient while the patient table was stationary.
Then, the patient table was moved either forward or backward and another rotation was performed to image the next section of the patient. This method of scanning is known as axial, or “step-and-shoot.” Because exams performed in axial scan mode consist of sections of a patient being successively imaged, it can take several minutes to scan over a large region of the patient. Because axial scanning is time-consuming it creates clinical disadvantages. For example, patients are typically asked to hold their breath for as long as possible during examinations over the trunk region because patient motion results in the blurring of anatomy in the CT images. Thus, the length of time a patient is able to hold his or her breath, which is typically less than 30 seconds, determines the number of consecutive images that are not degraded by respiratory motion; this issue is particularly important in thoracic imaging. The need for shorter scan times to capture larger regions of a patient’s anatomy within a single breath-hold motivated development of continuous acquisition scanners, which emerged in the early 1990s. Continuous CT acquisitions are performed in what is known as helical, or spiral, scan mode. Helical scanning allows the x-ray tube to continuously rotate as the patient table simultaneously moves through the scanner bore at a constant speed. Development of helical scan mode has resulted in faster scan times compared to axial scanning because table movement does not require additional scan time. Although both axial and helical scan modes are available on modern CT scanners, most patient examinations are performed in helical mode because of the time-saving advantages that helical scanning offers.
When scanning in helical mode, technologists must specify a pitch, which is defined as the length the patient table travels per rotation of the x-ray tube divided by the nominal beam width.³⁵ A helical scan with a pitch of 1 is equivalent to an axial scan in that the beam profiles from each rotation are contiguous; a pitch less than 1 results in beam overlap and a pitch greater than 1 results in gaps between the beam profiles from consecutive rotations. Figure 1.8 shows the beam profiles for axial and helical scanning (at different pitches). Although a spiral path is traced around a patient when scanning in helical mode, the acquired data can be reconstructed into consecutive planar images (like those acquired in axial mode) by first interpolating between data points in the z-direction.³⁶ Longer image reconstruction times, due to the additional processing time required for interpolation of helical data, are compensated for by shorter data acquisition time compared to axial scanning.⁷

**Figure 1.8.** The beam profiles “traced” around a patient during axial and helical scanning (for various pitches).
1.3. RADIATION DOSE

The intensity and energy of radiation emitted by a CT scanner is affected by the spectra of the x-ray beam, as well as other scanning factors. Output of a CT scanner is expressed as exposure, which is the ability of x-rays to ionize air. The unit of exposure is the roentgen (R), where 1 R produces $2.58 \times 10^{-4}$ coulombs of ionizing radiation per kilogram of air at standard temperature and pressure. Exposure measured in air can be converted to absorbed dose, which is the energy deposited to matter (e.g. the body), using appropriate conversion factors. Absorbed dose is expressed in units of gray (Gy) or rad, where 1 Gy equals 1 joule per kilogram and 1 rad equals 10 mGy. Compared to other radiographic imaging modalities, absorbed dose in CT imaging tends to be much higher.

Tissues in the body are not equally sensitive to the effects of ionizing radiation, thus the International Commission of Radiological Protection (ICRP) has assigned tissue weighting factors to tissues and organs based on their relative radiosensitivities; the most recent tissue weighting factors appear in Table 1.1. Because most CT scans do not expose the entire body to radiation, but rather irradiate a specific anatomic region (e.g., chest, head, abdomen) and the organs within that region, effective dose is calculated as the weighted
average of the dose absorbed by the irradiated organs.\textsuperscript{39}

\[
\text{Effective dose} = \sum w_T \cdot w_R \cdot D_{T,R}, \quad (1.3)
\]

where \( w_T \) is the tissue weighting factor, \( w_R \) is the radiation weighing coefficient, and \( D_{T,R} \) is the tissue absorbed dose. Units of effective dose are the sievert (Sv). Because \( w_R \) equals 1 for x-rays, absorbed and effective dose have the same units and therefore 1 Sv equals 1 Gy.\textsuperscript{8}

\textbf{Table 1.1.} The most recent tissue weighting factors (\( w_T \)) published in a 2007 report by the ICRP.\textsuperscript{38} The sum of all \( w_T \) values is 1.0.

<table>
<thead>
<tr>
<th>Tissue</th>
<th>Tissue weighting factor ((w_T))</th>
</tr>
</thead>
<tbody>
<tr>
<td>Breast</td>
<td>0.12</td>
</tr>
<tr>
<td>Colon</td>
<td>0.12</td>
</tr>
<tr>
<td>Lung</td>
<td>0.12</td>
</tr>
<tr>
<td>Red bone marrow</td>
<td>0.12</td>
</tr>
<tr>
<td>Stomach</td>
<td>0.12</td>
</tr>
<tr>
<td>Gonads</td>
<td>0.08</td>
</tr>
<tr>
<td>Bladder</td>
<td>0.04</td>
</tr>
<tr>
<td>Esophagus</td>
<td>0.04</td>
</tr>
<tr>
<td>Liver</td>
<td>0.04</td>
</tr>
<tr>
<td>Thyroid</td>
<td>0.04</td>
</tr>
<tr>
<td>Bone surface</td>
<td>0.01</td>
</tr>
<tr>
<td>Brain</td>
<td>0.01</td>
</tr>
<tr>
<td>Salivary glands</td>
<td>0.01</td>
</tr>
<tr>
<td>Skin</td>
<td>0.01</td>
</tr>
<tr>
<td>Remainder (Adrenals, etc.)</td>
<td>0.12</td>
</tr>
</tbody>
</table>
Average effective doses for various radiographic imaging examinations are listed in Table 1.2. These values can be compared to the mean effective dose to adults living in areas of normal background radiation, which is estimated to be 3 mSv/year. Background radiation comes from natural sources, including cosmic radiation from the sun and radioactive nuclides in the earth’s crust and in the air; therefore, levels of background radiation vary across geographical regions.

**Table 1.2.** Average effective dose for various radiographic examinations, including CT. Note that the CT doses are (for the most part) higher than those of other radiographic imaging modalities.

<table>
<thead>
<tr>
<th>Radiographic examination</th>
<th>Average effective dose (mSv)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intraoral radiograph (dental)</td>
<td>0.005</td>
</tr>
<tr>
<td>Posteroanterior chest radiograph</td>
<td>0.02</td>
</tr>
<tr>
<td>Cervical spine film</td>
<td>0.2</td>
</tr>
<tr>
<td>Mammogram</td>
<td>0.4</td>
</tr>
<tr>
<td>Head CT</td>
<td>2</td>
</tr>
<tr>
<td>Pelvic CT</td>
<td>6</td>
</tr>
<tr>
<td>Chest CT</td>
<td>7</td>
</tr>
<tr>
<td>Abdominal CT</td>
<td>8</td>
</tr>
<tr>
<td>Tumor (18F-FDG) (Nuclear medicine [PET])</td>
<td>14.1</td>
</tr>
<tr>
<td>Coronary angiography (CT)</td>
<td>16</td>
</tr>
</tbody>
</table>
1.3.1. **Radiation-induced Cancer**

While CT scans are useful diagnostic tools, they may put patients at risk for subsequently developing radiation-induced cancer.\(^{43}\) Specifically, it has been estimated that 1.5 – 2% of all cancers in the United States can be attributed to CT and that 1 out of every 500 children or 2,000 adults who receive a scan will develop fatal cancer.\(^{44,45}\) These rates can be compared to the estimate that 1 out of 4.7 deaths in the U.S. is due to cancer (for all causes).\(^{25}\) The estimated incidence of fatal cancer in exposed children is higher because sensitivity to radiation is inversely related to age.\(^{43}\) Children are more radiosensitive than adults because their cells divide more rapidly, making their DNA more vulnerable to damage, and because they have longer life expectancies over which to express this damage.\(^{46}\) Additionally, cancer risk from radiation exposure varies across gender and CT examinations. A study of atomic bomb survivors reported that women experience about twice the risk compared to men.\(^{47}\) Furthermore, a study of the health insurance records of 952,420 Americans reported that a higher percentage of women than men underwent at least one radiographic imaging procedure over a three-year period (78.7% versus 57.9%).\(^{12}\)

When examined concurrently, age and gender risk differentials appear even higher; lifetime cancer incidence estimates attributable to CT coronary angiography examinations indicate that a 20-year-old woman has 23 times the risk of an 80-year-old man, while a 20-year-old man only has 5 times the risk.\(^{48}\) One study estimated that a 20-year-old woman has 23 times the lifetime risk of
cancer attributable to CT coronary angiography compared to an 80-year-old man, while a 20-year-old man only has 5 times the risk.48

The CT examination type, which affects the amount of radiation exposure and which organs are exposed, also affects cancer risk. For example, a CT scan of the heart (coronary angiography) may lead to cancer in 1 out of every 150 women (aged 20 years), while routine head CT carries a lower risk of 1 cancer per 4,360 women.49 Effective dose is often used to estimate cancer risk; however, the tissue weighting factors used to calculate effective dose are constant regardless of a patient’s age or gender. Also, because effective dose combines dose for all organs into a single value, it has limited value in predicting cancer risk. For example, a CT scan of the pelvis with an effective dose of 50 mSv carries a lower risk of breast cancer than a chest CT scan with an effective dose of 5 mSv because, even though the effective dose is higher, breast tissue is not directly exposed during a pelvic scan.

Interactions between ionizing radiation (x-rays) and atoms and molecules inside the human body are necessary for the generation of CT images. However, exposure to ionizing radiation is a risk factor for cancer because of the damage it may cause to cellular components during these interactions, resulting in chromosome breakage, cell death, and oncogenic transformation. Absorption of radiation by water molecules is the primary source of cellular damage because the human body is predominantly composed of water. Interactions between water and ionizing radiation form free radicals, which are highly reactive and can
damage DNA and RNA. Many types of DNA damage can occur after exposure to ionizing radiation; double-strand breaks are the most lethal.\textsuperscript{50} Once DNA is damaged, specific molecular sensors activate damage processing (i.e., DNA damage response). Outcomes of the DNA damage response include successful repair (and restoration to full functionality), partial repair (resulting in an increased risk of mutagenesis and carcinogenesis), or cell death.\textsuperscript{51} Carcinogenesis is a late (long-term) stochastic effect, thus the probability of developing cancer increases with exposure to increasing levels of radiation, but the severity of the cancer is independent of the level of exposure.\textsuperscript{52} Radiation-induced cancer is only perceived after a latent period (e.g., 10 years for breast cancer)\textsuperscript{43} and cannot be differentiated from cancer caused by other carcinogens.\textsuperscript{8} Early effects of radiation exposure include erythema (skin burns) and epilation (hair loss).\textsuperscript{53}

Several studies have provided justification for a linear, no-threshold (LNT) relationship between the amount and duration of exposure to ionizing radiation and the likelihood of developing cancer.\textsuperscript{54, 55} A no-threshold effect means that any level of exposure (e.g., a single CT scan) incurs a risk to patients. Based on the census population and the estimated number of annual scans, one study calculated that 1 out of every 4.4 people received a CT scan in 2004;\textsuperscript{56} however, this statistic overestimates the true proportion because it did not account for patients receiving multiple scans within a year. Skeptics of the LNT model, who have cast doubt on the model's biological basis and validation, believe that it overestimates risk from doses lower than 10 mSv.\textsuperscript{57} Thus, from this opposing perspective, patients receiving single CT scans have a very minimal (if any)
increased cancer risk. The LNT model also assumes that risk accumulates linearly with each scan performed. Therefore, lifetime attributable risk must be considered for patients who receive multiple lifetime scans.\textsuperscript{58} The medical records of over 30,000 patients revealed that (over a 22-year period) 33\% of the study’s patients underwent five or more CT scans and 5\% of patients underwent between 22 and 132 CT scans.\textsuperscript{59} However, the percentage of patients who received multiple scans reported in this study is likely higher than among the general patient population because many of the participants were adult cancer patients (who typically undergo regular CT scans). A chart review of 355,088 children (younger than 18 years) found that 7.9\% received at least one CT scan and 1.1\% received three or more scans during the 3-year study period.\textsuperscript{60}

The estimated cancer risks attributable to radiation exposure are largely based on epidemiological studies of atomic bomb survivors. From 1950 to 1997, 86,572 atomic bomb survivors from Hiroshima and Nagasaki were followed,\textsuperscript{47} during this period, 36.8\% and 10.8\% of the survivors died of non-cancer diseases and solid cancer, respectively.\textsuperscript{47} Of the solid cancer deaths, 5\% were attributed to radiation exposure; 14.7\% of the 272 breast cancer deaths reported were considered to be radiation-induced.\textsuperscript{47} Breast tissue is especially susceptible to the carcinogenic effects of radiation because, compared to other organs, breast tissue has a heightened radiosensitivity.\textsuperscript{18} When the atomic bomb survivors were also pooled across age groups, it was observed that young children incurred the highest risk.
Breast cancer incidence varies by race and nationality, and because Japan has lower breast cancer incidence than the United States, it is unclear whether the results from studies of Japanese atomic bomb survivors translate to American women. Therefore, epidemiological studies have been conducted in North America with medical sources of radiation to further characterize breast cancer risk. In a multicenter study of 6,068 childhood cancer survivors (in the United States and Canada), a statistically significant excess in the number of cancer cases were observed; 24.7 times more cases of breast cancer were reported among this patient population than would have been expected among the general public. In another study, a chart review of 5,573 female scoliosis patients (under the age of 20 at the time of diagnosis) in the United States was performed to estimate their cumulative radiation dose; many patients received frequent radiographic examinations to monitor the condition of their spine. Because the latent period for radiation-induced breast cancer is at least 10 years, patients were tracked for an average of 40 years. Patients in the cohort were estimated to have experienced 1.69 times higher breast cancer incidence than in women who were exposed to background radiation only. Breast cancer incidence was increased by 3.4-fold in women first exposed between the ages of 10 and 11 years old, which is consistent with the belief that the breast tissue of prepubescent girls and girls who are beginning menstruation is most susceptible to radiation damage. Other studies suggest that women over 50 years at the time of exposure experience a minimal increase in breast cancer risk from diagnostic levels of radiation exposure. Additionally, higher incidences of breast
cancer have been observed in exposed patients that have a family history of cancer,\textsuperscript{64, 67} this may be because women that inherit deleterious BRCA1 or BRCA2 mutations are more radiosensitive due to deficiencies in DNA repair.\textsuperscript{58}

1.3.2. Dose Quantification

Absorbed dose to patients undergoing CT scanning is difficult to accurately quantify because organ dose cannot be directly measured, except perhaps in the case of cadaver studies. While dosimeters can be placed on a patient’s surface to give an indication of entrance or skin dose, it is difficult to quantify the dose effect using different techniques or scan parameters in patient studies as this would require scanning the patient multiple times and needlessly exposing them to harmful radiation. Therefore, effective dose is typically quantified using both phantom measurements and/or computational modeling. Two categories of test phantoms have been designed to approximate the attenuation of x-rays through a patient; the first represents the most basic simplification of human anatomy, while the second attempts to accurately represent the size and shape of the human body, internal anatomy, and differences in attenuation across different organs. Dose is estimated through computational modeling by mathematically simulating a CT scan, including radiation transport through a patient. Both phantoms and computational modeling can be used to quantify effective dose; however, phantoms measurements cannot accurately estimate organ dose for all of the tissue listed in Table 1.1. Furthermore, while modeling is capable of producing more accurate organ dose
estimates than phantoms (particularly for individual patients), this approach is only available to a few experimental research groups.

1.3.2.1. **CTDI PHANTOMS**

CT dose index (CTDI) is the primary and accepted means of estimating dose in CT. CTDI is quantified through exposure detected inside cylindrical polymethylmelacralate (PMMA) phantoms, which are standardized by the FDA. The phantoms used to measure CTDI are known as CTDI phantoms and are commercially available in several sizes. The most common sizes have 16 cm and 32 cm diameters and are used to represent the head and body of teenage and adult patients, respectively; the 16-cm phantom is used to represent both the head and body of pediatric patients. Both sizes of CTDI phantoms are 15-cm long. Holes are predrilled along the length of the phantoms (at the center and around the periphery) and a 10-cm long pencil ionization chamber is placed inside the holes to measure exposure. The ionization chamber, which is a standard air-filled radiation detector, is connected to an electrometer that displays the exposure. Locations of the peripheral chamber holes (1 cm below the surface) are identified in the same manner as the face of a clock: 3:00, 6:00, 9:00, and 12:00. Both CTDI phantoms are shown in Figure 1.9 along with their respective chamber hole locations. Because typically only one ionization chamber is used at a time, solid PMMA filler rods are placed in the empty chamber holes.
FIGURE 1.9. The two sizes of CTDI phantoms (32-cm and 16-cm) typically used to estimate adult patient dose received from a (a) body and (b) head scan, respectively.

Exposure is measured inside the CTDI phantoms for a single axial rotation (i.e., stationary patient table) and converted to CTDI<sub>100</sub>, which is defined as the radiation dose (normalized to the beam width) measured for 100 mm along the length of the ionization chamber, using the following equation: \(^{72}\)

\[
\text{CTDI}_{100} = \frac{f \cdot C \cdot E \cdot L}{100 \cdot N \cdot T},
\]

where \( f \) converts in-air exposure to dose and equals 0.87 rad/R for dose to the air-filled chamber, \( C \) is the calibration factor of the exposure meter connected to the ionization chamber, \( E \) is the detected exposure value in units of R, \( L \) is the
active detection length of the chamber (100 mm), and \( N \cdot T \) is the nominal beam width in mm (\( N \) is the number of active detective rows and \( T \) is the detector row thickness). Detected exposure, and thus the calculated CTDI\(_{100}\) values, may vary across the phantom’s chamber holes; for instance, exposure measured at the peripheral chamber holes in a 32-cm CTDI phantom is typically around twice what is measured at the central chamber hole. Therefore, an average, or weighted, CTDI (CTDI\(_w\)) is calculated as follows:\(^{72}\)

\[
CTDI_w = \frac{1}{3} CTDI_{100, \text{central}} + \frac{2}{3} CTDI_{100, \text{peripheral}},
\]

where CTDI\(_{100, \text{central}}\) is the CTDI\(_{100}\) value calculated from exposure measured at the central chamber hole and CTDI\(_{100, \text{peripheral}}\) is the CTDI\(_{100}\) value calculated from exposure measured at one of the peripheral chamber holes. For the 16-cm phantom, exposure measured at the peripheral and central chamber holes is relatively consistent.

CTDI\(_{100}\) has been shown to underestimate dose deposited in CTDI phantoms by 10% - 37% because the 10-cm long pencil chamber can only detect dose for 10 cm in a 15-cm long phantom.\(^{73}\) It has been shown that a Farmer chamber, which is a small ionization chamber with a volume of 0.6 mm\(^3\), can be used to yield more accurate dose estimates.\(^{74, 75}\) In addition to issues with using CTDI\(_{100}\), dose measured in CTDI phantoms, which have a circular cross-section and lack internal organs or tissue differentiation, does not accurately reflect dose to patients. Furthermore, the CTDI values across two patients scanned with the same protocol parameter settings will not be equal although the absorbed dose will
vary across these patients, even when the patients of very similar in size. Despite these limitations, CTDI is the standard metric because it provides a simple way of quantifying scanner output and can be used to compare dose estimates obtained across scan parameters. Thus, in the future, the FDA may require CT scanner manufacturers to display CTDI calculated values on the scanner console to increased awareness of dose among CT technologists with the intention of preventing overdose incidents.

1.3.2.2. ANTHROPOMORPHIC PHANTOMS

Human-like anthropomorphic phantoms more accurately represent anatomy and tissue attenuation than CTDI phantoms. Anthropomorphic phantoms are commercially available in a variety of shapes, patient sizes, and anatomical complexities. These phantoms contain internal holes where point dose detectors can be placed to measure organ dose; thermoluminescent dosimeters (TLDs) are commonly used for this purpose. Unlike an ionization chamber, which detects dose within a volume, TLDs are so small that they are considered to detect dose at a single point. TLDs must be processed and read by trained personnel in order to obtain dose information. Additionally, large variations in detected exposure across TLDs (for a constant level of scanner output and at constant location on the phantom) are possible because of their point-dose nature. Therefore, despite the advantages of using anthropomorphic phantoms compared to CTDI phantoms (i.e., better representation of patient anatomy and the ability to estimate effective dose), limitations also exist.
1.3.2.3. **MONTE CARLO DOSE ESTIMATES**

Although a method for calculating effective dose has been established (Eq. 1.3), it relies on knowledge of the absorbed dose to each organ. Dose absorbed by each organ must be estimated because there is currently no way of directly measuring organ dose inside patients. Monte Carlo (MC) modeling seeks to improve upon the limitations of effective dose quantification in phantoms by providing organ dose information for humans of varying sizes scanned using varying protocols. Monte Carlo techniques may be used to estimate dose by mathematically simulating the passage of x-ray photons through a patient. 76

There are two approaches currently used to model a patient, mathematical and voxel-based. 78 The first approach simplifies human anatomy by using geometrical shapes to represent organs and tissues. These models were first developed at the Oak Ridge National Laboratory based on a “Reference Man” created by the ICRP to represent an average-sized male and are known as Medical Internal Radiation Dose (MIRD) models. 79 MIRD models are not anatomically realistic but are more accessible than voxelized models of actual patients because each voxel in a patient image must be identified as belonging to a specific organ or tissue through a process known as segmentation. Although segmentation can be semi-automated, some organs, and particularly organ boundaries, must be identified manually. This process can be time-consuming and requires extensive anatomical knowledge. For this reason, a limited
database of voxelized models exists, especially for full-body patient models with each organ identified. Some of the most widely-used voxelized patient models were developed by the Gesellschaft für Strahlen- und Umweltforschung (GSF) - National Research Centre for Environment and Health. Because internal anatomy varies greatly among individuals and organ dose estimates obtained in voxelized models are specific to the patient modeled, they do not reflect dose to a different patient undergoing the exact same scan. Nevertheless, Monte Carlo modeling’s ability to estimate absorbed dose to an entire organ provides an advantage compared to dose measurements obtained in phantoms.  

To estimate dose absorbed by specific organs, passage of individual photons through the patient is simulated. To do so, the kVp is input and a spectra is generated based on the scanner type. The spectra is divided into 1 keV bins from 1 to the kVp. Each bin is given a weight based on the likelihood of occurrence, which is determined from the spectra. The number of photons at each keV is normalized to the total number of photons in the spectrum such that the contributions at each keV will sum to 1.0. Then each keV is assigned to a range of numbers on a number line from 0 to 1.0 based on their relative contribution. A value from 0 to 1.0 is randomly generated by the MC model and then assigned a keV according to that number. Random numbers do not have the same likelihood of being assigned to each keV as demonstrated in Table 1.3. The shape and unequal energy distribution of photons in an x-ray spectra is taken into account in the MC model in this way because it is more likely to generate a number assigned to 60 keV than 1 keV. Because $10^6$ or more photons are followed
through the model, the energy spectrum of the beam can be accurately represented. However, because single photons are followed individually the MC model cannot directly incorporate or account for the current-exposure time product (mA·s).

**TABLE 1.3.** Example of the type of look-up table used when assigning photon energies, which range from 1 keV to the specified kVp, based on a random number generated by the MC model.

<table>
<thead>
<tr>
<th>Random number generated</th>
<th>keV value assigned</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 – 0.005</td>
<td>1</td>
</tr>
<tr>
<td>0.005 – 0.010</td>
<td>2</td>
</tr>
<tr>
<td>0.300 – 0.500</td>
<td>60</td>
</tr>
<tr>
<td>0.600 – 0.700</td>
<td>70</td>
</tr>
</tbody>
</table>

After a simulation is performed, output of the MC model is in units of mGy/photon and is converted to dose in units of mGy/mA·s by multiplying output by a normalization factor. The normalization factor is the ratio of free-in-air exposure measured using a pencil ionization chamber on the same as scanner simulated to the free-in-air exposure simulated by the MC model for the same scan parameters. Finally, dose in mGy/mA·s can be multiplied by the desired mA·s.
During CT scanning, it is important to minimize patient dose while maintaining adequate image quality. However, since reductions in radiation dose are typically accompanied by degradations in image quality, the "as low as reasonably achievable" (ALARA) principle should be followed to minimize dose while maintaining diagnostic image quality. This is particularly important for pediatric patients, who incur the greatest risk of future long-term effects and receive about 11% of CT scans in the United States. In 2002, the FDA issued a public health notification entitled "Reducing radiation risk from computed tomography for pediatric and small adult patients," which advised radiologists and medical physicists to prescribe CT sparingly to young children and to adjust scan parameters appropriately. Although other forms of medical doses (e.g., pharmaceuticals) are adjusted for children, prior to this notification, pediatric patients were frequently scanned using the same exam protocols as for adult patients. Release of the notification has led to increased awareness and to lower doses in pediatric scanning. In 2007, the Image Gently campaign was started to raise awareness and change practice such that lower doses are used when scanning pediatric patients and has outlined steps to consider when scanning pediatric patients. In addition to modifying protocols to reduce dose, several strategies have emerged in recent years, including tube current modulation (TCM), in-plane organ shielding, and new iterative reconstruction techniques (discussed in 1.2.4).
When considering ALARA for pediatric patients, it is important to keep in mind that if dose is too low there is the risk of missing a diagnosis. Thus, therapeutic or imaging goals (e.g., obtaining a diagnosis in cancer patients or assessing injury in emergency department/room patients) dictate the acceptable levels of dose reduction. In additional, image quality is affected by the size of the patient (the same protocol will give very different image quality across patient sizes and should be adjusted according to the patient’s size). Specifically, children tend to have a lack of visceral fat, which affects attenuation of radiation in the body, thus effecting both image quality and dose.

The effective doses listed in Table 1.2 represent averages, but because of the differences in imaging needs, dose calculated for different CT exam types has been shown to vary by a factor of 13 within and across hospitals. While variation in dose across hospitals due to differences in imaging needs is positive in terms of showing patient specification, it is important to ensure consistency in how scans are performed at a given hospital. In another study at a different hospital, compliance rates (i.e., technologists following the prescribed protocols) of up to 88% (but as low as 59%) were reported. Compliance ensures that the low-dose protocols are used, which helps to prevent overdose incidents.

1.3.2.4. Tube Current Modulation

Tube current modulation is a dose-reduction technology that has emerged in recent years as a result of the advanced capabilities of modern CT scanners. TCM is the CT version of automatic exposure control, which has traditionally
been used in conventional radiography systems as a means of preventing image overexposure. The delay in implementing such technology into CT was likely due to the fact that overexposure in CT actually improves image quality (which is not the case in conventional radiography). Thus, automatic exposure control is used in CT for regulating patient dose. TCM lowers dose by vary the x-ray tube current within a scan according to the patient’s anatomy; specifically, current, and thus radiation dose, is decreased when smaller or thinner parts of the patient’s body (e.g., the neck) are scanned and is increased when larger or thicker parts (e.g., across the shoulders) are scanned. Although multiple factors control the amount of radiation output by the CT scanner, the tube current is the easiest to modify and has the most predictable behavior.

TCM promotes ALARA by enabling patient specificity between exams; the specificity of TCM across patients is demonstrated in Figure 1.10. Both angular (within the xy-plane) and longitudinal (along the z-axis) approaches to TCM have been developed and are typically combined (i.e., xyz modulation) in modern CT scanners to maximize dose-savings. However, each CT scanner vendor has a unique approach to accomplishing TCM.
Figure 1.10. TCM schemes (superimposed on the patients’ localization projection images) used when scanning two 8 year old female patients. Although both patients are very similar in size (the patient shown in [a] was 127 cm tall and weighed 22.0 kg at the time of scanning, while patient [b] was 128 cm tall and weighed 24.7 kg), they had noticeably different TCM patterns. The above images were collected as part of an IRB approved study.
Prior to the development of TCM, the x-ray tube current was constant (static) throughout the exam regardless of variations in patients’ anatomy. Compared to scans with a static mA, dose reductions of up to 40% have been observed (without a significant increase in noise) from applying angular TCM in simulations, phantom measurements, and cadaver studies. Similar results were also observed in a study comparing pediatric patients scanned with and without TCM.

1.3.2.5. **IN-PLANE ORGAN SHIELDING**

Radiation protection garments have been placed over patients undergoing radiological procedures for years (e.g., lead aprons for conventional and dental radiography). These garments are designed to protect the body from radiation and are used in CT to specifically shield superficial radiosensitive organs (i.e., the eye, breast, thyroid, and testes). Shielding the patient has the same effect as adding filtration to the x-ray beam, which is the preferential removal of low energy photons that are unlikely to penetrate through a patient and are instead likely to be attenuated upon entrance. Therefore, removing these photons from the x-ray beam, which are thought not to contribute to improved image quality, reduces patient dose. Lead shielding has been shown to reduce fetal dose during CT scanning in pregnant patients, another study also showed dose reduction of 87% to the male gonads during abdominopelvic MDCT.

While protective garments have historically been lead-based, most garments used in CT are now bismuth-based, because they conform better to the
body and are lighter-weight.\textsuperscript{104, 105} Bismuth garments were first introduced into CT in the 1990s and have since been approved by the FDA. Dose reductions ranging from 41 - 57\% to the breast (as measured with TLDs placed on the patients' skin surface) have been observed in patients scanned with bismuth breast shields in place.\textsuperscript{106, 107} Similarly, dose reductions of 18 - 48\% to the lens of the eye have been observed in patients and phantoms from using bismuth shields.\textsuperscript{108-110} Furthermore, bismuth shielding has also been shown to reduce dose to the testes and the thyroid.\textsuperscript{106, 109} Although dose-savings has consistently been reported with in-plane organ shielding, some studies have observed a negative effect on image quality.\textsuperscript{111-113}

1.4. \textbf{FORWARD TO THESIS}

The aim of this thesis was two-fold: (1) to accurately quantify dose and (2) to develop new strategies to reduce breast dose to pediatric and young adult female patients during CT scanning. These patient populations experience one of the highest risks related to secondary cancers due to their increased radiation exposure that is part of diagnosis and treatment of cancer in the patients at MD Anderson Cancer Center – the facility where this research was conducted.

In order to evaluate success of novel breast dose reduction strategies, the ability to accurately quantify relative dose is important. Thus, the first part of this thesis (i.e., Chapters 2 and 3) focused on quantifying dose in terms of both
physical measurement precision and gathering data for Monte Carlo modeling. The second part of the thesis (i.e., Chapters 4 – 6) focused on strategies for reducing breast dose. Initially, measurements in CTDI phantom were used to evaluate dose penalties (to regions of the phantom that best represent breast dose [i.e., the peripheral 12:00 chamber position]) associated with particular scanning techniques and protocols (i.e., axial versus helical acquisitions). After considering the dose effect of current scanning protocols and procedures, this thesis explored the efficacy of novel approaches (i.e., strategies not available on current scanners), including tube voltage modulation and built-in breast shielding, for reducing radiation dose delivered to the breast tissue of female young adult and pediatric patients undergoing CT scans. Both anthropomorphic phantoms and MC modeling were employed to estimate the amount of radiation dose deposited to the breast tissue before and after implementation of the novel techniques. Given the relationship between increased radiation exposure and cancer incidence, reducing breast dose in girls and young women will in turn decrease the rate of radiation-induced breast cancer in adult women who received CT scans as children or young adults.
CHAPTER 2

AN EMPIRICAL MODEL OF CT ATTENUATION UNDER NARROW-BEAM GEOMETRY

PURPOSE: The purpose of this study was to develop and validate a mathematical model to describe narrow-beam attenuation of kilovoltage x-ray beams for the intended applications of half-value layer (HVL) and quarter-value layer (QVL) estimation, patient organ shielding, and computer modeling.

METHODS: An empirical model, which uses the Lambert W function and represents a generalized Lambert-Beer law, was developed. To validate this model, transmission of diagnostic energy x-ray beams was measured over a wide range of attenuator thicknesses (0.49 mm Al to 34.00 mm) on a CT scanner. Exposure measurements were acquired under narrow-beam geometry using standard methods, including the appropriate ionization chamber. Nonlinear regression was used to find the best-fit curve of the proposed Lambert W model to each measured transmission versus attenuator thickness data set. In addition to validating the Lambert W model, we also assessed the performance of two-point Lambert W interpolation compared to traditional methods for estimating the HVL and QVL (i.e., semilogarithmic [exponential] and linear interpolation).

RESULTS: The Lambert W model was validated for modeling attenuation versus attenuator thickness with respect to the data collected in this study ($R^2>0.99$). Furthermore, Lambert W interpolation was more accurate and less sensitive to
the choice of interpolation points used to estimate the HVL and/or QVL than the traditional methods of semilogarithmic and linear interpolation.

**CONCLUSION:** The proposed Lambert W model accurately describes attenuation of both monoenergetic radiation and (kilovoltage) polyenergetic beams (under narrow-beam geometry).

### 2.1. INTRODUCTION

The Lambert-Beer law describes attenuation of a monoenergetic beam (composed of photons of a single energy) and is represented as

\[
I = I_0 e^{-\mu x}, \quad (2.1)
\]

where \(I\) is the intensity of photons transmitted through an attenuating material of thickness \(x\), \(\mu\) is the linear attenuation coefficient, and \(I_0\) is the unattenuated intensity. In a polyenergetic beam, the energy spectra of the photons changes as the beam passes through an attenuating material; as a result, the linear attenuation coefficient is not a constant value but rather varies with beam attenuation.\(^8\) Because x-rays are polyenergetic, attenuation of a clinical x-ray beam cannot be accurately modeled by Eq. (2.1).

Development of an equation to describe x-ray beam attenuation may benefit several applications, one of which is HVL and QVL estimation. Quality of an x-ray beam is characterized by its spectra, which is difficult to measure
directly. The HVL, which is defined as the attenuator thickness that reduces the measured exposure to exactly one-half of the unattenuated exposure, is the metric most often used to describe the quality of an x-ray beam. Therefore, the HVL is routinely estimated during quality assurance (QA) testing. The QVL is defined as the attenuator thickness that reduces measured exposure to exactly one-quarter of the unattenuated exposure. Although estimation of the QVL is not as routine as that of the HVL, it is necessary for determining the homogeneity coefficient, which is used to characterize the polyenergetic nature of an x-ray beam. Additionally, some MC computer models use both the HVL and QVL to construct virtual CT scanners. Typically, two-point interpolation is performed to estimate the HVL or QVL from two measured data points \([x_1, I_1], (x_2, I_2)\) bracketing the HVL or QVL; that is, with one thickness \(x_1\) less than and one thickness \(x_2\) greater than the HVL or QVL. Alternatively, two-point extrapolation uses two thicknesses \(x_1\) and \(x_2\), which are both less than or both greater than the HVL or QVL. Traditionally, either two-point semilogarithmic (i.e., exponential) or linear interpolation has been used for HVL and QVL estimation, despite the fact that neither function accurately describe attenuation in radiographic imaging. The measurements used to estimate the HVL or QVL are gathered under narrow-beam geometry, which refers to detection of photons from the primary beam and does not take into account the contribution of scatter; conversely, broad-beam geometry refers to radiation that is outside the primary beam (i.e., scatter).
Four mathematical models for describing attenuation of megavoltage x-ray beams (two of which were evaluated in this study) were discussed by Kleinschmidt, who suggested that a function $\bar{\mu}$ can be written to describe the mean attenuation coefficient of a polyenergetic beam.\textsuperscript{118} The model that Kleinschmidt determined to be optimal was excluded from our analysis because it contained three unknown variables and thus conflicted with the objective of this study, which was to develop an attenuation model that could be used to estimate the HVL and QVL by two-point interpolation. The first of the two mean attenuation coefficient functions $\bar{\mu}$ discussed by Kleinschmidt and considered in our analysis was developed by Bjärngard and Shackford as:\textsuperscript{119}

$$\bar{\mu} = \mu_0 - \lambda x, \quad (2.2)$$

where $\mu_0$ and $\lambda$ are unknown parameters. The second function we considered was proposed by Yu et al. as an alternative to Eq. (2.2),\textsuperscript{120}

$$\bar{\mu} = \frac{\mu_0}{1 + \lambda x}, \quad (2.3)$$

with all variables defined as above.

Because Eqs. (2.2) and (2.3) were intended for use in describing high-energy (megavoltage) photon beams, an empirical model for $\bar{\mu}$ was proposed for diagnostic-energy (kilovoltage) x-ray beams with the intent that the model be utilized to describe narrow-beam attenuation through homogeneous aluminum,
which is typically used for HVL and QVL estimation:

\[ \bar{\mu} = \mu_0 + \lambda \left( \frac{I}{I_0} \right) \]  \hspace{1cm} (2.4)

Replacing constant \( \mu \) by function \( \bar{\mu} \) (i.e., Eq. [2.4]), Eq. (2.1) can be rewritten as (see Appendix A for derivation [Chapter 8]):

\[ I = I_0 \frac{W(\lambda x e^{-\mu_0 x})}{\lambda x} \]  \hspace{1cm} (2.5)

where \( I \), \( x \), and \( I_0 \) are defined as in Eq. (2.1) and \( W \) refers to the Lambert W function. The Lambert W function has been shown to have several applications in physics\(^{121}\) and is defined as the multivalued inverse of the function

\[ W(z)e^{W(z)} = z \]  \hspace{1cm} (2.6)

where \( z \) is a complex number.\(^{122}\) By L'Hopital's rule, Eq. (2.5) is reduced to Eq. (2.1) when used to model attenuation of a monoenergetic beam (\( \mu_0 = \mu \) and \( \lambda = 0 \)). Therefore, Eq. (2.5) represents a more general form of the Lambert-Beer law (Eq. [2.1]).

Although Eqs. (2.2) and (2.3) were not originally developed for or validated under the conditions of this study (i.e., kilovoltage x-ray beam attenuation), there were no constraints included in the publications describing their derivation that would specifically exclude them from being applied to this type of data. Therefore, the following attenuation models (Eqs. [2.7] and [2.8]) were derived (in
the same manner as Eq. [2.5]) by substituting Eqs. (2.2) and (2.3), respectively, in to Eq. (2.1):

$$I = I_0 e^{-\mu_0 x + \lambda x^2}, \quad (2.7)$$

$$I = I_0 e^{\frac{-\mu_0 x}{1+\lambda x}}. \quad (2.8)$$

The purpose of this study was to validate the proposed Lambert W model (Eq. [2.5]) using transmitted exposure versus attenuator thickness data obtained (under narrow-beam geometry) on a CT scanner. Versatility of the proposed model was evaluated by collecting data over a range of conditions, including two kVps and two experimental setups. After validating the Lambert W model over a broad range of conditions, we evaluated the accuracy of using two-point Lambert W interpolation to estimate the HVLs and QVLs compared to accepted methods (i.e., semilogarithmic and linear interpolation) and also to interpolations based on the models described by Yu et al. and Bjärngard and Shackford.

2.2. MATERIALS AND METHODS

2.2.1. DATA COLLECTION

Exposure measurements were collected through varying thicknesses of type 1100 aluminum alloy (Al 1100), which were measured using a digital caliper (with 0.01-mm accuracy). Two unattenuated exposure measurements ($I_0$) were initially collected and averaged for every data set. Data points were recorded as
measured exposure ($I$) versus attenuator thickness ($x$). Exposure was detected (under narrow-beam geometry) using an exposure meter (RadCal Corporation, Monrovia, CA) connected to a RadCal ionization chamber (either a 3-cm$^3$ pencil chamber or 0.6-cm$^3$ Farmer chamber).

Two experimental setups (shown in Figure 2.1) were used to gather exposure versus attenuator thickness data on a LightSpeed QX/i CT scanner (GE Healthcare, Waukesha, WI) so that the proposed Lambert W attenuation model could be evaluated using a traditionally-accepted method, as well as a more recent approach.$^{74,123}$ Exposure readings were measured at 80 kVp and 120 kVp (20-mm nominal beam width, large focal spot size, and “body” scan FOV [SFOV]) using a 10-cm pencil ionization chamber in the first (traditional) approach and a Farmer chamber in the second (more recent) approach. In both cases, the chambers were positioned free-in-air at gantry isocenter. Additionally, the x-ray tube was positioned at the bottom of the scanner gantry and held stationary at that location throughout testing; sheets of Al 1100 were placed at the bottom of the gantry so that they directly covered the x-ray tube output port.
FIGURE 2.1. Two experimental setups were used to collect data on a GE LightSpeed QX/i CT scanner. With the x-ray tube held stationary at the bottom of the gantry, varying thicknesses of Al 1100 filters (indicated by the arrows) were placed over the x-ray tube output port and transmitted exposure was detected with either (a) a 10-cm pencil ionization chamber (circled) or (b) a Farmer chamber (circled). Lead collimation was added at the bottom of the gantry for the Farmer chamber setup and the Al 1100 attenuators were placed directly over the lead collimators.
In the second setup, the Farmer chamber was suspended from a hole (with roughly the same diameter as the chamber) drilled into the side of a long, narrow foam block such that the inactive region of the chamber remained inside the hole to secure the chamber in place. The foam block was positioned on the patient table with the active volume of the chamber suspended free-in-air. Using the scanner’s built-in laser lights, the length of the chamber was aligned along the x-axis (rather than the z-axis, as was the case for the pencil chamber) and its midpoint was aligned at the intersection of the y- and z-axes (see Fig. 2.1b for orientation of the chamber and the axes). The Farmer chamber was suspended in this manner (orientation) because its length exceeded the widest beam available for the QX/i scanner and we wanted to ensure that the entire active detection region of the chamber was exposed. To avoid detection of scattered photons, the fan beam was physically collimated (along the x-axis) to match the length of the chamber. Collimation was achieved by placing two 1.5-mm thick sheets of lead at the bottom of the gantry and two foam blocks (cut to the length of the active detection region of the chamber) in between the lead sheets on each side of the x-ray tube output port. A plumb-bob was hung from the chamber to ensure that the opening between the lead sheets was properly aligned with the chamber. Upon alignment confirmation, the foam and lead were taped in place to maintain a constant opening throughout testing, during which the Al 1100 attenuators were placed directly on top of the lead collimators.

The thicknesses of added aluminum, along with other information related to the methods, appear in Table 2.1.
TABLE 2.1. Details of the experimental conditions under which exposure versus attenuator thickness data were collected using two experimental setups on a GE LightSpeed QX/i CT scanner.

<table>
<thead>
<tr>
<th>Experimental setup</th>
<th>kVp</th>
<th>Range of attenuator thicknesses (mm)</th>
<th>No. of data points collected</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pencil chamber</td>
<td>80</td>
<td>0.49 – 20.28</td>
<td>41</td>
</tr>
<tr>
<td></td>
<td>120</td>
<td>0.49 – 33.03</td>
<td>68</td>
</tr>
<tr>
<td>Farmer chamber</td>
<td>80</td>
<td>0.49 – 34.00</td>
<td>70</td>
</tr>
<tr>
<td></td>
<td>120</td>
<td>0.49 – 34.00</td>
<td>70</td>
</tr>
</tbody>
</table>

*Both setups used to collect the CT data are shown in Figure 2.1. The setup which used a 10-cm pencil ionization chamber did not feature any additional beam collimation, while the x-ray beam was collimated in the Farmer ionization chamber setup, using two 1.5-mm thick lead sheets, to match the length of the chamber (along the x-axis).

†For all data sets, average unattenuated exposure (I₀) was first determined and then Al 1100 filters were added in approximately 0.5-mm increments within the specified attenuator thickness range; the number of attenuator thicknesses tested within this range is equal to the number of exposure versus thickness data points collected.
2.2.2. Data Analysis

The proposed Lambert W model (Eq. [2.5]) was validated (within the context of this study) as an acceptable means of modeling attenuation of diagnostic x-ray beams. Additionally, the model's fit to the measured transmission data was compared to that of the monoenergetic Lambert-Beer law (Eq. [2.1]) and also to that of the polyenergetic attenuation models described by Eqs. (2.7) and (2.8). After validating and comparing Eq. (2.5) to other attenuation models, the accuracy of using two-point interpolation based on the Lambert W function to estimate the HVL and QVL was assessed. The Lambert W estimates were compared to estimates from the traditional methods of semilogarithmic and linear interpolation and also to estimates from interpolations based on Eqs. (2.7) and (2.8).

2.2.2.1. Validation of Lambert W Model

Prior to validating the Lambert W model (Eq. [2.5]), we will justify the reasoning behind its development. Since Eq. (2.4) was used to derive Eq. (2.5) and thus is the basis of the Lambert W model, evidence in support of a linear relationship between $\bar{\mu}$ and $(L/L_0)$ (with slope $\lambda$ and y-intercept $\mu_0$) could provide such justification. However, because $\bar{\mu}$ was not quantified in this study, we could not directly test for the presence of such a linear relationship. Instead, we rearranged Eq. (8.8) (derived from the substitution of Eq. [2.4] into Eq. [2.1], as shown in the appendix) such that the right side of the equation matched that of...
Eq. (2.4) and \( \bar{\mu} \) was replaced by a term which used the data at hand:

\[
-\ln\left(\frac{1}{I_0}\right) = \mu_0 + \lambda\left(\frac{1}{I_0}\right). \tag{2.9}
\]

For every data set, \( -\ln\left(\frac{1}{I_0}\right) \) versus \( \left(\frac{I}{I_0}\right) \) was plotted and goodness of fit was judged through least squares linear regression; Figure 2.2 shows one such plot along with the corresponding regression analysis. However, because these plots were generated by transforming the data to create a linear graph, some assumptions of linear regression were violated.\(^\text{124}\) Additionally, transmission \( \left(\frac{I}{I_0}\right) \), which was the dependent variable of the measured data, appears in both the independent and dependent variable terms in Eq. (2.9). Therefore, while it was inappropriate to interpret results of the linear regression analysis, the plots were used to visually assess the linear relationship described by Eq. (2.9) as a means of indirectly validating Eq. (2.4).
**FIGURE 2.2.** Scatter plot of \(-\ln \left( \frac{I}{I_0} \right) \) versus \( \frac{I}{I_0} \) for data collected at 80 kVp using the Farmer chamber setup. The calculated \( R^2 \) value of 0.97 indicates a close fit of the linear trendline (which has the following equation: \(-\ln \left( \frac{I}{I_0} \right) = 0.101 + 0.049 \cdot \left( \frac{I}{I_0} \right) \)) to the data, thus providing support in favor of Eq. (2.9) and indirectly justifying the validity of Eq. (2.4). Note that the \( R^2 \) value and the estimates of \( \mu_0 \) and \( \lambda \) listed here differ from those determined through nonlinear regression (for the reasons explained above) even though the same data was analyzed. Table 2.2 lists the \( R^2 \) value for this data set as 0.9999, as well as estimates of \( \mu_0 \) and \( \lambda \) determined through nonlinear regression based on the Lambert W model (Eq. [2.5]).
Since the linear scatter plots provided justification for Eq. (2.4), proper analysis of the Lambert W model (Eq. [2.5]) was warranted. Hence, Eq. (2.5) was fit to the detected transmission \( \left( \frac{I}{I_0} \right) \) versus attenuator thickness \( (x) \) data in Matlab R2008b (MathWorks Inc., Natick, MA) by nonlinear least squares regression, which used an iterative search algorithm\(^{125}\) to determine the best-fit values of unknown parameters \( \mu_0 \) and \( \lambda \). For each data set, ninety-five percent confidence intervals for the best-fit parameters were calculated, as well as the coefficient of determination \( (R^2) \) and the sum of squares error \( (SSE) \) of the best-fit curves. Additionally, the HVL and QVL were estimated by substituting the best-fit values of \( \mu_0 \) and \( \lambda \) into Eqs. (8.10 - 8.11) (shown in Appendix B); the HVL and QVL were estimated for data set comparison purposes only and were not used in the analysis of two-point Lambert W interpolation for HVL and QVL estimation. The same nonlinear regression analysis was also performed for Eqs. (2.1), (2.7), and (2.8).

Since the \( R^2 \) values for all four models were (for the most part) very similar to one another, Akaike’s information criterion (AIC) was used to provide another source of comparison by which to gauge the models’ relative goodness of fit.\(^{126}\) For every data set, an AIC value was calculated for each of the four attenuation models (Eqs. [2.1], [2.5], [2.7], and [2.8]). The AIC values by themselves are meaningless, and so evidence ratios were used to compare the AIC values calculated for the models described by Eqs. (2.1), (2.7), and (2.8) to the AIC values calculated for the Lambert W model (Eq. [2.5]). The evidence ratios described the likelihood the Lambert W model was correct (e.g., an evidence
ratio of 10 indicated that the Lambert W model was 10 times more likely to be correct than the model it was being compared to) by computing the probability that it was correct divided by the probability that another model (i.e., Eqs. [2.1], [2.7] or [2.8]) was correct. Rough rules of thumb used when interpreting the AIC values were that a difference between two AIC values of less than 2, which corresponds to an evidence ratio of less than 2.7, indicates substantial evidence for the other model, while a difference between 4 and 7 indicates considerably less support for the other model, and a difference greater than 10 (or an evidence ratio greater than 148) indicates essentially no support in favor of the other model.

### 2.2.2.2. HVL and QVL Estimation

Lambert W interpolation was compared to traditional methods (semilogarithmic and linear interpolation) for HVL and QVL estimation, as well as interpolations based on Eqs. (2.7) and (2.8). Each interpolation method was judged by the accuracy of its predictions, which was evaluated by comparing measured transmission to predicted transmission through the HVL or QVL nearest neighbor. Additionally, the methods were compared in terms of the sensitivity of their predictions to the choice of interpolated data points. Finally, although interpolation is typically used for HVL and QVL estimation, we also considered whether extrapolation could yield accurate estimates.

The first step in obtaining our predictions was to divide the measured data points (except for unattenuated exposure, $I_0$) into pairs and to calculate $\mu_0$ and $\lambda$.
for each pair using Eqs. (8.12 - 8.13) (see Appendix B [Chapter 8]). Although \( \mu_0 \) and \( \lambda \) can be calculated using more than two data points, such as in 2.2.2.1 where all of the collected data points were used to determine the best-fit values of \( \mu_0 \) and \( \lambda \), typically only two data points (i.e., one pair) are experimentally collected for the purpose of HVL or QVL estimation. The number of pairs analyzed and thus the number of \( \mu_0 \) and \( \lambda \) values calculated depended on the number of data points collected for each data set (shown in Table 2.1); for example, 41 data points were recorded for the pencil chamber setup at 80 kVp, which translated to 780 pairs being analyzed. All analyzed pairs were categorized according to whether they bracketed (interpolation) or lay on one side of (extrapolation) the HVL or QVL. Furthermore, a subcategory of interpolation was considered by analyzing those pairs that bracketed both the HVL and QVL.

The calculated values of \( \mu_0 \) and \( \lambda \) for each data pair and the HVL or QVL nearest neighbor (\( x_{\text{nearest}} \)) for the data set that the pair came from were substituted into the Lambert W model (Eq. [2.5]) to predict transmission \( \left( \frac{I_{\text{predicted}}}{I_0} \right) \) through thickness \( x_{\text{nearest}} \); transmission through \( x_{\text{nearest}} \) was also predicted by semilogarithmic and linear interpolation, and by interpolations based on the Bjärngard and Shackford (Eq. [2.7]) and Yu et al. (Eq. [2.8]) models. The measured data points nearest the HVL or QVL (i.e., \( x_{\text{nearest}} \)) were used to approximately represent these values since the HVL and QVL were not directly measured and therefore their true values were unknown.
the HVL ([Xnearest, Inearest]), (2) group all other data points, except for (0, 1), into pairs and separate the pairs according to whether they bracket (interpolation) or lay on one side of (extrapolation) the HVL; (3) calculate μ₀ and λ for each pair (from Eqs. [8.12-8.13] in Appendix B); (4) substitute Xnearest, μ₀, and λ into Eq. (2.5) to calculate \( \frac{I_{\text{predicted}}}{I_0} \) for each pair; (5) use Eq. (2.10) to calculate the RPD between each \( \frac{I_{\text{nearest}}}{I_0} \) and \( \frac{I_{\text{predicted}}}{I_0} \); (6) calculate the mean and range of RPDs and also the fraction of interpolation and extrapolation pairs that yielded RPDs ≤ 5%. Similar steps can be followed to evaluate the accuracy of QVL estimation, the other measured data sets, and the other four interpolation methods.
FIGURE 2.3. Sample calculation using data collected with the pencil chamber at 80 kVp. The following steps were performed to calculate the RPDs and related statistical analysis for this data set: (1) identify the measured data point closest to
2.3. RESULTS

2.3.1. VALIDATION OF LAMBERT W EQUATION

Plots of transmission versus attenuator thickness for all four data sets, along with the corresponding Lambert W best-fit curves, appear in Figure 2.4; results of the nonlinear regression analysis, including the $R^2$ values, are shown below each plot. Because the HVL and QVL estimates calculated from the Lambert W best-fit lines for the Farmer chamber and pencil chamber data sets agreed within 5%, both setups were considered to be acceptable for estimating the HVL or QVL; we could not determine which setup was better since the true HVL and QVL were unknown. The residuals (i.e., differences between the measured transmissions and the transmission values predicted by the best-fit lines for the models described by Eqs. [2.1], [2.5], [2.7], and [2.8]) are plotted in Figure 2.5; additional details of the nonlinear regression analysis for all of the models can be found in Table 2.2.
FIGURE 2.4. Lambert W best-fit curves for data collected on a GE QX/i LightSpeed CT scanner at 80 and 120 kVp. Two experimental setups (shown in Fig. 2.1) were used to collect exposure measurements, which were detected by either a Farmer ionization chamber or a 10-cm long pencil ionization chamber.
FIGURE 2.5. Residual plots for data collected on a GE QX/i LightSpeed CT scanner at 80 and 120 kVp; the two experimental setups used to collect exposure measurements are shown in Figure 2.1.
TABLE 2.2. Results of the nonlinear regression analysis used to validate the Lambert W model and to compare its performance to the Yu et al., Bjärnsgard and Shackford, and Lambert-Beer attenuation models.

<table>
<thead>
<tr>
<th>Data set</th>
<th>Model</th>
<th>Best-fit parameters with 95% CIs</th>
<th>HVL and QVL estimates</th>
<th>SSE</th>
<th>$R^2$</th>
<th>AIC value</th>
<th>Evidence Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pencil chamber (80 kVp)</td>
<td>Lambert W</td>
<td>$\mu_0 = 0.097 \pm 0.001$, $\lambda = 0.052 \pm 0.002$</td>
<td>$HVL = 5.64 \text{ mm Al}$, $QVL = 12.63 \text{ mm Al}$</td>
<td>0.0002</td>
<td>0.9999</td>
<td>-493</td>
<td>N/A</td>
</tr>
<tr>
<td></td>
<td>Yu et al.</td>
<td>$\mu_0 = 0.135 \pm 0.001$, $\lambda = 0.018 \pm 0.001$</td>
<td>$HVL = 5.62 \text{ mm Al}$, $QVL = 12.48 \text{ mm Al}$</td>
<td>0.0005</td>
<td>0.9997</td>
<td>-455</td>
<td>$1.61 \times 10^8$</td>
</tr>
<tr>
<td></td>
<td>Bjärnsgard and Shackford</td>
<td>$\mu_0 = 0.132 \pm 0.002$, $\lambda = 0.002 \pm 0.000$</td>
<td>$HVL = 5.64 \text{ mm Al}$, $QVL = 12.39 \text{ mm Al}$</td>
<td>0.0011</td>
<td>0.9995</td>
<td>-426</td>
<td>$3.41 \times 10^{14}$</td>
</tr>
<tr>
<td></td>
<td>Lambert-Beer</td>
<td>$\mu = 0.114 \pm 0.003$</td>
<td>$HVL = 6.07 \text{ mm Al}$, $QVL = 12.15 \text{ mm Al}$</td>
<td>0.0202</td>
<td>0.9903</td>
<td>-308</td>
<td>$1.63 \times 10^{40}$</td>
</tr>
<tr>
<td>Farmer chamber (80 kVp)</td>
<td>Lambert W</td>
<td>$\mu_0 = 0.101 \pm 0.001$, $\lambda = 0.053 \pm 0.002$</td>
<td>$HVL = 5.43 \text{ mm Al}$, $QVL = 12.14 \text{ mm Al}$</td>
<td>0.0004</td>
<td>0.9999</td>
<td>-840</td>
<td>N/A</td>
</tr>
<tr>
<td></td>
<td>Yu et al.</td>
<td>$\mu_0 = 0.135 \pm 0.002$, $\lambda = 0.013 \pm 0.001$</td>
<td>$HVL = 5.49 \text{ mm Al}$, $QVL = 11.80 \text{ mm Al}$</td>
<td>0.0022</td>
<td>0.9994</td>
<td>-720</td>
<td>$1.21 \times 10^{26}$</td>
</tr>
<tr>
<td></td>
<td>Bjärnsgard and Shackford</td>
<td>$\mu_0 = 0.131 \pm 0.002$, $\lambda = 0.001 \pm 0.000$</td>
<td>$HVL = 5.55 \text{ mm Al}$, $QVL = 11.75 \text{ mm Al}$</td>
<td>0.0038</td>
<td>0.9990</td>
<td>-681</td>
<td>$3.31 \times 10^{34}$</td>
</tr>
<tr>
<td></td>
<td>Lambert-Beer</td>
<td>$\mu = 0.116 \pm 0.002$</td>
<td>$HVL = 5.97 \text{ mm Al}$, $QVL = 11.93 \text{ mm Al}$</td>
<td>0.0290</td>
<td>0.9922</td>
<td>-541</td>
<td>$7.58 \times 10^{64}$</td>
</tr>
<tr>
<td>Data set</td>
<td>Model</td>
<td>Best-fit parameters with 95% CIs</td>
<td>HVL and QVL estimates</td>
<td>SSE</td>
<td>$R^2$</td>
<td>AIC value</td>
<td>Evidence ratio</td>
</tr>
<tr>
<td>--------------------------</td>
<td>---------------</td>
<td>---------------------------------------</td>
<td>---------------------------</td>
<td>--------</td>
<td>------</td>
<td>-----------</td>
<td>----------------</td>
</tr>
<tr>
<td>Pencil chamber (120 kVp)</td>
<td>Lambert W</td>
<td>$\mu_0 = 0.068 \pm 0.000, \lambda = 0.038 \pm 0.001$</td>
<td>HVL = 7.99 mm Al, QVL = 17.91 mm Al</td>
<td>0.0001</td>
<td>1.0000</td>
<td>-887</td>
<td>N/A</td>
</tr>
<tr>
<td></td>
<td>Yu et al.</td>
<td>$\mu_0 = 0.095 \pm 0.001, \lambda = 0.011 \pm 0.001$</td>
<td>HVL = 7.99 mm Al, QVL = 17.59 mm Al</td>
<td>0.0018</td>
<td>0.9995</td>
<td>-711</td>
<td>$2.29 \cdot 10^{38}$</td>
</tr>
<tr>
<td></td>
<td>Bjärngard and Shackford</td>
<td>$\mu_0 = 0.092 \pm 0.001, \lambda = 0.001 \pm 0.000$</td>
<td>HVL = 8.04 mm Al, QVL = 17.46 mm Al</td>
<td>0.0031</td>
<td>0.9991</td>
<td>-673</td>
<td>$2.78 \cdot 10^{46}$</td>
</tr>
<tr>
<td></td>
<td>Lambert-Beer</td>
<td>$\mu = 0.080 \pm 0.002$</td>
<td>HVL = 8.69 mm Al, QVL = 17.37 mm Al</td>
<td>0.0347</td>
<td>0.9903</td>
<td>-511</td>
<td>$4.27 \cdot 10^{81}$</td>
</tr>
<tr>
<td>Farmer chamber (120 kVp)</td>
<td>Lambert W</td>
<td>$\mu_0 = 0.072 \pm 0.000, \lambda = 0.037 \pm 0.001$</td>
<td>HVL = 7.67 mm Al, QVL = 17.08 mm Al</td>
<td>0.0001</td>
<td>1.0000</td>
<td>-915</td>
<td>N/A</td>
</tr>
<tr>
<td></td>
<td>Yu et al.</td>
<td>$\mu_0 = 0.098 \pm 0.001, \lambda = 0.011 \pm 0.001$</td>
<td>HVL = 7.68 mm Al, QVL = 16.75 mm Al</td>
<td>0.0015</td>
<td>0.9996</td>
<td>-745</td>
<td>$5.6 \cdot 10^{36}$</td>
</tr>
<tr>
<td></td>
<td>Bjärngard and Shackford</td>
<td>$\mu_0 = 0.095 \pm 0.001, \lambda = 0.001 \pm 0.000$</td>
<td>HVL = 7.73 mm Al, QVL = 16.64 mm Al</td>
<td>0.0027</td>
<td>0.9993</td>
<td>-705</td>
<td>$3.08 \cdot 10^{45}$</td>
</tr>
<tr>
<td></td>
<td>Lambert-Beer</td>
<td>$\mu = 0.083 \pm 0.002$</td>
<td>HVL = 8.33 mm Al, QVL = 16.65 mm Al</td>
<td>0.0319</td>
<td>0.9915</td>
<td>-535</td>
<td>$3.84 \cdot 10^{82}$</td>
</tr>
</tbody>
</table>
The $R^2$ values for the Lambert W model (Eq. [2.5]) ranged from 0.9999 to 1.000 across all data sets, indicating a very close fit of the model to the data. When compared to other attenuation models, the AIC evidence ratios indicated that the Lambert W model ranged (across all four data sets) from $1.63 \cdot 10^{40}$ to $3.84 \cdot 10^{62}$, $3.41 \cdot 10^{14} - 2.78 \cdot 10^{46}$, and $1.61 \cdot 10^8 - 2.29 \cdot 10^{38}$ times more likely to be the correct model compared to the Lambert-Beer (Eq. [2.1]), Bjärngard and Shackford (Eq. [2.7]), and Yu et al. (Eq. [2.8]) models, respectively. Because the Lambert W model was well over one-million times more likely to be the correct model, there was persuasive evidence in its favor. Furthermore, the Lambert W model featured higher $R^2$ values, across all data sets, than the models described by Eqs. (2.1), (2.7), and (2.8).

2.3.2. HVL AND QVL ESTIMATION

Figure 2.6 shows the percentage of acceptable pairs (i.e., pairs with RPDs ≤ 5%) for predicting transmission through either the HVL or QVL nearest neighbor by each interpolation method. Although Figure 2.6 does not specifically quantify accuracy of the predictions (beyond the criteria of 5%), it does indicate that (in general) Lambert W interpolation was the most versatile estimation method in terms of flexibility in the choice of interpolation data points. Table 2.3 also lists the fraction and corresponding percentage of acceptable interpolation pairs, as well as the mean and range of RPDs for each interpolation method.
FIGURE 2.6. These bar graphs provide a visual representation of the Lambert W, Yu et al., Bjärngard and Shackford, semilogarithmic and linear interpolation methods’ respective sensitivities to the choice of interpolation points used to predict transmission through the attenuator thickness nearest the HVL or QVL ($X_{\text{nearest}}$) for each data set. The RPDs between the predicted and measured transmissions through $X_{\text{nearest}}$ were calculated and the percentage of acceptable interpolation pairs (i.e., pairs with RPDs ≤ 5%) for estimating (a) HVL and (b) QVL is shown.
TABLE 2.3. Results of the HVL and QVL estimation analysis. For each interpolation method, the fraction (in parentheses) and percentage of acceptable interpolation pairs (i.e., pairs with RPDs ≤ 5%) for estimating the HVL (only), QVL (only), and both the HVL and QVL (jointly) is shown; each fraction’s denominator is equal to the number of interpolation pairs and RPDs calculated for that data set. The mean and range (in parentheses) of RPDs across all interpolation pairs for each data set is also shown.

<table>
<thead>
<tr>
<th>Data set</th>
<th>Interpolation method</th>
<th>HVL estimation</th>
<th>QVL estimation</th>
<th>HVL and QVL estimation</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Acceptable pairs (%)</td>
<td>RPD (%)</td>
<td>Acceptable pairs (%)</td>
</tr>
<tr>
<td>Pencil chamber (80 kVp)</td>
<td>Lambert W</td>
<td>100%</td>
<td>1.18</td>
<td>100%</td>
</tr>
<tr>
<td></td>
<td>(300/300)</td>
<td>(0.11 - 3.37)</td>
<td></td>
<td>(384/384)</td>
</tr>
<tr>
<td></td>
<td>Yu et al.</td>
<td>100%</td>
<td>0.44</td>
<td>100%</td>
</tr>
<tr>
<td></td>
<td>(300/300)</td>
<td>(0.00 - 2.18)</td>
<td></td>
<td>(384/384)</td>
</tr>
<tr>
<td></td>
<td>Bjärngard and Shackford</td>
<td>100%</td>
<td>0.76</td>
<td>92.2%</td>
</tr>
<tr>
<td></td>
<td>(300/300)</td>
<td>(0.00 - 3.49)</td>
<td></td>
<td>(354/384)</td>
</tr>
<tr>
<td></td>
<td>Semilogarithmic</td>
<td>68%</td>
<td>4.07</td>
<td>85.4%</td>
</tr>
<tr>
<td></td>
<td>(204/300)</td>
<td>(0.30 - 12.1)</td>
<td></td>
<td>(328/384)</td>
</tr>
<tr>
<td></td>
<td>Linear</td>
<td>22.3%</td>
<td>14.7</td>
<td>26.8%</td>
</tr>
<tr>
<td></td>
<td>(67/300)</td>
<td>(0.53 - 45.3)</td>
<td></td>
<td>(103/384)</td>
</tr>
<tr>
<td>Farmer chamber (80 kVp)</td>
<td>Lambert W</td>
<td>100%</td>
<td>1.22</td>
<td>99.4%</td>
</tr>
<tr>
<td></td>
<td>(300/300)</td>
<td>(0.00 - 4.05)</td>
<td></td>
<td>(1052/1058)</td>
</tr>
<tr>
<td>Data set</td>
<td>Interpolation method</td>
<td>HVL estimation</td>
<td>QVL estimation</td>
<td>HVL and QVL estimation</td>
</tr>
<tr>
<td>----------</td>
<td>----------------------</td>
<td>----------------</td>
<td>----------------</td>
<td>-----------------------</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Acceptable pairs</td>
<td>RPD (%)</td>
<td>Acceptable pairs</td>
</tr>
<tr>
<td>Farmer chamber (80 kVp)</td>
<td>Yu et al.</td>
<td>100% (590/590)</td>
<td>0.93 (0.00 - 4.28)</td>
<td>73.8% (781/1058)</td>
</tr>
<tr>
<td></td>
<td>Bjärgard and Shackford</td>
<td>99.3% (586/590)</td>
<td>1.37 (0.00 - 5.21)</td>
<td>61.0% (645/1058)</td>
</tr>
<tr>
<td></td>
<td>Semilogarithmic</td>
<td>56.3% (332/590)</td>
<td>5.13 (0.06 - 13.7)</td>
<td>67.8% (717/1058)</td>
</tr>
<tr>
<td></td>
<td>Linear</td>
<td>11.4% (67/590)</td>
<td>21.6 (0.31 - 62.1)</td>
<td>13.6% (144/1058)</td>
</tr>
<tr>
<td>Pencil chamber (120 kVp)</td>
<td>Lambert W</td>
<td>100% (780/780)</td>
<td>0.37 (0.00 - 0.86)</td>
<td>100% (1120/1120)</td>
</tr>
<tr>
<td></td>
<td>Yu et al.</td>
<td>99.1% (773/780)</td>
<td>1.84 (0.00 - 5.44)</td>
<td>87.1% (975/1120)</td>
</tr>
<tr>
<td></td>
<td>Bjärgard and Shackford</td>
<td>92.4% (721/780)</td>
<td>2.40 (0.00 - 6.96)</td>
<td>77.1% (863/1120)</td>
</tr>
<tr>
<td></td>
<td>Semilogarithmic</td>
<td>76.3% (595/780)</td>
<td>3.07 (0.01 - 11.7)</td>
<td>84.1% (942/1120)</td>
</tr>
<tr>
<td>Data set</td>
<td>Interpolation method</td>
<td>HVL estimation</td>
<td></td>
<td>QVL estimation</td>
</tr>
<tr>
<td>--------------------------</td>
<td>----------------------</td>
<td>----------------</td>
<td>-------</td>
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</tr>
<tr>
<td></td>
<td></td>
<td>Acceptable pairs</td>
<td>RPD (%)</td>
<td>Acceptable pairs</td>
</tr>
<tr>
<td>Pencil chamber (120 kVp)</td>
<td>Linear</td>
<td>29.6%</td>
<td>14.4</td>
<td>(231/780)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>22.9%</td>
<td>21.9</td>
<td>(256/1120)</td>
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<td></td>
<td></td>
<td>1.67%</td>
<td>28.2</td>
<td>(8/480)</td>
</tr>
<tr>
<td>Farmer chamber (120 kVp)</td>
<td>Lambert W</td>
<td>100%</td>
<td>0.52</td>
<td>(770/770)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>100%</td>
<td>0.46</td>
<td>(1188/1188)</td>
</tr>
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<td></td>
<td></td>
<td>100%</td>
<td>0.67</td>
<td>(504/504)</td>
</tr>
<tr>
<td></td>
<td>Yu et al.</td>
<td>100%</td>
<td>1.04</td>
<td>(770/770)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>88.0%</td>
<td>2.35</td>
<td>(1046/1188)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>71.8%</td>
<td>2.61</td>
<td>(362/504)</td>
</tr>
<tr>
<td></td>
<td>Bjärngard and Shackford</td>
<td>99.2%</td>
<td>1.50</td>
<td>(764/770)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>75%</td>
<td>3.26</td>
<td>(891/1188)</td>
</tr>
<tr>
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<td></td>
<td>43.8%</td>
<td>3.71</td>
<td>(221/504)</td>
</tr>
<tr>
<td></td>
<td>Semilogarithmic</td>
<td>69.6%</td>
<td>3.75</td>
<td>(536/770)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>81.9%</td>
<td>2.66</td>
<td>(973/1188)</td>
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<td>43.8%</td>
<td>4.69</td>
<td>(221/504)</td>
</tr>
<tr>
<td></td>
<td>Linear</td>
<td>24.8%</td>
<td>16.0</td>
<td>(191/770)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>21.3%</td>
<td>24.5</td>
<td>(253/1188)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>1.19%</td>
<td>31.3</td>
<td>(253/1188)</td>
</tr>
</tbody>
</table>

*Pairs containing points that bracketed both the HVL and QVL were interpolated to estimate transmission through both the HVL and QVL nearest neighbors; thus, two RPDs were calculated for each interpolation pair. The mean and range of RPDs shown includes RPDs for both the HVL and QVL estimates.
Lambert W interpolation was shown to be successful in all of the HVL cases analyzed (2440 out of 2440 pairs), regardless of the choice of data points interpolated, and in over 99.8% of the QVL cases analyzed (3744 out of 3750 pairs). The mean and range of RPDs between the measured and predicted (by Lambert W interpolation) transmissions through the HVL nearest neighbor across all four data sets was 0.72% (0.00 – 4.05%); similarly, the mean and range of RPDs for the QVL nearest neighbor across the data sets was 0.64% (0.00 – 5.17%). Across all data sets and all interpolation pairs (HVL and QVL), the Lambert W interpolated predictions yielded RPDs ranging from 0.00% to 5.17% (99.9% of the 6190 pairs analyzed had RPDs ≤ 5%) with a mean RPD of 0.68%. These results can be compared to the mean and range of RPDs (across all 6190 pairs) of 1.96% (0.00 – 13.1%) for interpolation based on the Yu et al. model (90.8% of pairs had RPDs ≤ 5%), 2.74% (0.00 – 16.2%) for interpolation based on the Bjarngard and Shackford model (82.8% of pairs had RPDs ≤ 5%), 3.37% (0.00 – 15.4%) for semilogarithmic interpolation (74.7% of pairs had RPDs ≤ 5%), and 22.6% (0.01 – 147%) for linear interpolation (21.2% of pairs had RPDs ≤ 5%).

The special case of using Lambert W interpolation to estimate both the HVL and QVL from a single pair of two measured data points, as opposed to using two pairs (i.e., four data points) to interpolate HVL and QVL separately, had a 99.6% acceptance rate (across all data sets). This rate can be compared to 67.5%, 44.1%, 38.7%, and 1.0% acceptance when using Yu et al., Bjarngard
and Shackford, semilogarithmic, and linear interpolations, respectively, to jointly estimate HVL and QVL.

Lambert W extrapolation was also evaluated and RPDs of less than or equal to 5% were achieved by approximately 91.4% of pairs (4842 out of 5243 pairs and 3549 out of 3933 pairs used to predict transmission through the HVL and QVL nearest neighbors, respectively). Lambert W extrapolation was more versatile than Yu et al., Bjärngard and Shackford, semilogarithmic, and linear extrapolations, which had acceptance rates of 81.3%, 75.0%, 62.8%, and 15.4%, respectively. However, the diminished success rate of Lambert W extrapolation compared to Lambert W interpolation (91.4% versus 99.9% acceptance) indicates that, based on our results, interpolation rather than extrapolation should be used to estimate the HVL or QVL.

2.4. DISCUSSION

In this study, we proposed an empirical model, which uses the Lambert W function, to describe narrow-beam attenuation of diagnostic x-ray beams through homogenous materials, and our analyses showed that the proposed model accurately describes such attenuation. Furthermore, accuracy of the Lambert W attenuation model (Eq. [2.5]) over the broad range of testing conditions considered (two kVps and two experimental setups) demonstrates the versatility of this model.
Although there is an established method for assessing attenuation under broad-beam geometry, this method does not necessarily hold true when describing attenuation of the primary beam (and vice versa) because the x-ray spectra are very different under these two geometries. Therefore, the focus of this study was on developing a narrow-beam attenuation model. While several such models have been developed for high-energy beams, to the best of the authors’ knowledge, no such model has been verified for determining narrow-beam attenuation of diagnostic energy beams.

The proposed Lambert W model appears to closely model narrow-beam attenuation of kilovoltage x-ray beams in light of the high $R^2$ values computed in this study, which ranged from 0.9999 to 1.000. These promising results may motivate future development of the empirical Lambert W model which presently relies on experimental data and could be enhanced by generation of look-up tables for constants $\mu_0$ and $\lambda$, similar to the National Institute of Standards and Technology (NIST) mass attenuation coefficient tables. If look-up tables were developed, parameters $\mu_0$ and $\lambda$ would be specified according to the kVp, attenuating material, inherent filtration, and possibly other factors that affect the polyenergetic beam spectra (e.g., the bowtie filter). Our data sets demonstrate the need for specification according to these factors. For example, the effect of kVp is apparent when comparing the $\mu_0$ and $\lambda$ values across data sets collected at 80 kVp versus 120 kVp. Besides the expected differences in $\mu_0$ and $\lambda$ that we observed across data sets, unexpected (but possibility justified) inconsistencies in the best-fit parameters were also observed. Differences across the two setups
may explain the statistically-significant differences noted in $\mu_0$ at the $\alpha = 0.05$ level (as indicated by the lack of overlap in the 95% confidence intervals). In order to develop generic (non-manufacturer specific) look-up tables, inconsistencies in $\mu_0$ and $\lambda$ for data sets with the same attenuating material, kVp, and inherent filtration would need to be resolved and the stability of $\mu_0$ and $\lambda$ across these data sets would need to be established.

Two-point Lambert W interpolation was considered to be a successful means of predicting transmission through the HVL nearest neighbor, yielding RPDs of less than or equal to 5% across all data sets regardless of the choice of interpolation data points. In addition to showing less sensitivity to the choice of interpolation points compared to the other interpolation methods analyzed, Lambert W interpolation displayed a higher degree of accuracy in its predictions. Of the four other interpolation methods evaluated, linear interpolation had the weakest performance, while interpolations based on the Yu et al. model most closely matched the performance of the Lambert W model. Despite its diminished success compared to Lambert W based interpolation, linear interpolation was observed to yield accurate predictions (within 5%) as long as the interpolated data points were within $\pm30\%$ of the HVL or $\pm15\%$ of the QVL; similarly, semilogarithmic interpolation was observed to yield accurate predictions when the interpolated data points were within $\pm80\%$ of the HVL or $\pm60\%$ of the QVL. The Bjärngard and Shackford model was generally less versatile and yielded less accurate estimates of transmission through the HVL and QVL nearest neighbors compared to the other therapy-based model evaluated in this study (i.e., the Yu
et al. model); furthermore, semilogarithmic interpolation was superior to interpolations based on the Bjärngard and Shackford model for some data sets. Although the Yu et al. model had a higher overall RPD across all data sets (1.96% versus 0.68%), higher average and maximum RPDs (across most data sets), and was less versatile in its percentage of acceptable pairs than the Lambert W model, the Lambert W model's principal advantage was its usefulness for estimating both the HVL and QVL simultaneously. While the accuracy of HVL and QVL estimation was not directly evaluated in this study, we expect it to be roughly equivalent to the level of accuracy observed in the predicted transmissions through the HVL and/or QVL nearest neighbors. Therefore, we can assume that Lambert W interpolation allows for accurate estimation of both the HVL and QVL from the unattenuated exposure \( \left( I_0 \right) \) and two measured data points (e.g., exposure measured through 0.5 mm and 33 mm of Al 1100). Thus, it is feasible that two-point Lambert W interpolation could be implemented into QA practice. To estimate the HVL and/or QVL, the unattenuated exposure and two measured exposure versus attenuator thickness data points could be substituted into Eqs. (8.10 – 8.13) (in Appendix B), which could be incorporated in a spreadsheet as part of QA practice. Equations (8.10 – 8.13) are the appropriate forms to use when estimating the HVL and/or QVL because they describe attenuator thickness in terms of intensity (transmission) rather than Eq. (2.5), which expresses intensity as a function of attenuator thickness.
For all HVL or QVL estimation practices, the authors recommend that the attenuator thicknesses be measured with a digital caliper. Our findings indicated that the nominal thicknesses of Al 1100 differed from the measured thicknesses by up to 10%. Therefore, by not measuring the attenuator thicknesses, the HVL and QVL estimates could be misrepresented, for instance, by as much ±0.8 mm of aluminum and ±1.79 mm of aluminum, respectively, at 120 kVp. Additionally, constants $\mu_0$ and $\lambda$ would change if the nominal rather than the measured attenuator thicknesses were used to determine these values, potentially leading to further misrepresentation of the HVL and QVL estimates. Since the nominal attenuator thicknesses were only precise to ±10%, it is reasonable that the original acceptance criterion of 5% be adjusted accordingly. We can apply this adjustment to our data, despite the fact that we measured the attenuator thicknesses, because we assessed performance of the Lambert W model by comparing measured and predicted transmissions rather than by comparing the HVL and QVL estimates to their true values. Since all RPDs observed in this study (0.00 - 5.17%) met the adjusted acceptance criterion, we can conclude that Lambert W interpolation accurately estimated transmission through both the HVL and QVL regardless of the choice of interpolation data points.

Besides using the proposed Lambert W model to estimate the HVL and QVL, Eq. (2.5) could also be applied for other purposes, such as empirically determining attenuation through a given thickness of radiation protective clothing (i.e., in-plane bismuth breast shields, lead aprons, etc.); the international standard IEC 61331-3 specifies that the lead-equivalent thickness of a material
used for radiation-shielding garments be measured under narrow-beam geometry.\textsuperscript{130} Additionally, Eq. (2.5) can be used in MC modeling, as well as in other computer simulations of attenuation of an x-ray beam through inherent, bowtie, or other filtration. However, in order to apply Eq. (2.5) in either of these circumstances, additional analysis would need to be performed to validate the Lambert W model under the specific conditions being tested or modeled.

2.5. CONCLUSION

The empirical model proposed in this study represents a generalized Lambert-Beer law that describes attenuation of both monoenergetic and polyenergetic radiation. In the case of monoenergetic radiation, the proposed Lambert W model mathematically reduces to the Lambert-Beer law. Applications of the Lambert W model relevant to polyenergetic radiation include HVL estimation and narrow-beam patient shielding (e.g., lead apron) calculations. HVL estimation is simplified by the Lambert W model because only two attenuator thicknesses need to be supplied (e.g., 2 mm and 30 mm of Al 1100). Additionally, the Lambert W model can be utilized to jointly interpolate both the HVL and QVL from the same two measured data points. Besides yielding more accurate estimates of the HVL and/or QVL compared to traditional methods (i.e., semilogarithmic and linear interpolation), other advantages of using Lambert W interpolation are that it is less time-consuming (as it requires less guess-work.
and fewer measurements) and less vulnerable to experimenter error (since it is relatively insensitive to the choice of interpolation points).
CHAPTER 3

PRECISION OF DOSIMETRY-RELATED MEASUREMENTS OBTAINED ON CURRENT MDCT SCANNERS

PURPOSE: CT intrascanner and interscanner variability has not been well characterized. Thus, the purpose of this study was to examine the within-run, between-run, and between-scanner precision of physical dosimetry-related measurements collected over the course of one year on three different makes and models of MDCT scanners.

METHODS: Physical measurements were collected using nine CT scanners (three scanners each of: GE LightSpeed VCT, GE LightSpeed 16, and Siemens Sensation 64). Measurements were made using various combinations of technical factors, including kVp, type of bowtie filter, and x-ray beam collimation, for several dosimetry-related quantities, including: (a) free-in-air CTDI (CTDI$_{100,\text{air}}$); (b) calculated HVLs and QVLs; and (c) CTDI$_{w}$ calculated from exposure measurements collected in both a 16-cm and 32-cm CTDI phantom. Data collection was repeated at several different time intervals, ranging from seconds (for CTDI$_{100,\text{air}}$ values) to weekly for three weeks and then quarterly or triannually for one year. Precision of the data was quantified by the percent coefficient of variation (%CV).

RESULTS: The maximum relative precision error (maximum %CV) across all dosimetry metrics, time periods, and scanners included in this study was 4.33%.
The median observed %CVs for CTDI\textsubscript{100, air} ranged from 0.05% to 0.19% over several seconds, 0.12% - 0.52% over one week, and 0.58% - 2.31% over three to four months. For CTDI\textsubscript{w} for a 16-cm and 32-cm CTDI phantom, respectively, the range of median %CVs was 0.38% - 1.14% and 0.62% - 1.23% in data gathered weekly for three weeks and 1.32% - 2.79% and 0.84% - 2.47% in data gathered quarterly or tri-annually for one year.

**CONCLUSION:** From a dosimetry perspective, the MDCT scanners tested in this study demonstrated a high degree of within-run, between-run, and between-scanner precision (with relative precision errors typically well under 5%).

### 3.1. INTRODUCTION

Recent media attention regarding risk to patients receiving CT scans has prompted the need for more accurate patient radiation dose estimates.\textsuperscript{45} MC computer modeling is making fast progress in the area of CT dose estimation and is moving closer towards clinical use. One such advancement was the recent development of an equivalent source model, which uses physical data, including the HVL and QVL to construct virtual models of CT scanners.\textsuperscript{116} Equivalent source models are constructed for every combination of scanner make and model, kVp, and type of bowtie filter because a unique x-ray spectrum, which is energy-dependent and filtration-dependent, exists for each scanner. Once an equivalent source model has been constructed, free-in-air exposures are used in the calculation of normalization factors (described by DeMarco et al.) which convert output of the MC model to absolute dose (in mGy/MA\textperiodcentered s); a unique,
scanner-specific normalization factor exists for every combination of technical factors, including kVp, bowtie filter, and x-ray beam collimation. This study sought to expand the number of available equivalent source MC models by supplying the HVL, QVL, and CTDI$_{100}$,air at multiple techniques for three scanner models produced by two manufacturers.

In order for MC models to be considered reliable or trustworthy, accuracy of the dose estimates obtained from the model must be verified by benchmarking the model. Benchmarking, or validating, a MC model entails comparing the output of the simulation with the same type of physical dose metric experimentally collected on the same type of CT scanner under the same conditions as simulated. Because the HVL and QVL are used to construct equivalent source models and CTDI$_{100}$,air values are used to convert MC output to dose, these quantities cannot be used for benchmarking purposes; instead, CTDI$_w$ values can be employed. Although most MC codes are validated with CTDI$_w$ data for only one kVp, CTDI$_w$ data were collected at several kVps in this study because validating a code at multiple energies demonstrates a more robust model and thus provides a stronger validation.

Previous studies have shown that a MC model can yield dose estimates that agree within 3.5% to physical data collected in a 32-cm diameter CTDI phantom on a LightSpeed 16 CT scanner (GE Healthcare). Although 3.5% appears to indicate a high level of conformity, the data used to benchmark the model were obtained from a single scanner during a single measurement.
session; such measurements could potentially vary between scanners and over the lifetime of a scanner. Additionally, the level of agreement between MC dose estimates and physical dose measurements necessary for validation of the model has not yet been established. Therefore, in order to develop “pass-fail” benchmarking criteria, it is important to gauge measurement precision, or the closeness of repeated measurements obtained from a given scanner or “sibling” scanners of the same model. Measurement precision is most commonly quantified through the %CV, which is defined as the standard deviation divided by the mean multiplied by 100%.

According to the International Organization for Standardization (ISO), measurement precision encompasses measurement repeatability, intermediate precision, and reproducibility. Repeatability, or within-run precision, refers to precision estimates obtained when tests are performed over a very short time period by the same experimenter using the same equipment on the same subject at the same location; in this study, within-run precision was characterized using CTDI_{100, air} values calculated from repeated free-in-air exposure measurements. Intermediate precision, or within-laboratory reproducibility, refers to precision estimates obtained when measurements are made at one testing site on different days (between-run), by different experimenters (between-operator), or using different test equipment (between-scanner); between-run and between-scanner precision were evaluated in this study. Between-run precision was assessed for
two different between-run time periods by conducting both weekly and quarterly (every three months) or triannual (every four months) measurement sessions. Between-scanner variability was determined by comparing dosimetry values collected on sibling scanners of the same model. Reproducibility, which ISO refers to as between-laboratory precision, could not be assessed in this study due to lack of feasibility. Therefore, short-term and long-term intrascanner [within-run, between-run (weekly), and between-run (quarterly or triannual)] and interscanner (between-scanner) precision were characterized in this study.

The purpose of this study was two-fold: first, to collect a series of dosimetry-related values (CTDI\textsubscript{100, air}, HVL, QVL, and CTDI\textsubscript{w}) on different makes and models of MDCT scanners in order to develop and benchmark MC equivalent source models; and second, to facilitate the eventual formation of “pass-fail” benchmarking criteria by characterizing the short-term and long-term intrascanner and interscanner precision of the data collected.

3.2. MATERIALS AND METHODS

3.2.1. RADIATION DOSIMETRY MEASUREMENTS

Exposure readings were measured using a 10-cm pencil ionization chamber and an electrometer (RadCal Corporation) on three LightSpeed VCT and three LightSpeed 16 CT scanners (GE Healthcare) at site A and three Sensation 64 CT scanners (Siemens Healthcare, Forchheim, Germany) at site B.
Readings were collected on all nine scanners at various time intervals over the course of one year; the authors chose this time period because it was within the time frame of what is considered a long-term study (as specified by Bonnick and Lewis)\(^{134}\) and because it roughly reflected the average lifespan of the CT x-ray tubes at both sites. At the beginning of each measurement session, within-run precision was gauged by collecting repeated free-in-air exposure measurements using a constant technique. Subsequently, the remaining data (free-in-air exposure measurements using varying techniques, HVL, QVL, and CTDI\(_{w}\) calculated values) were acquired on the scanner. To evaluate both between-run (weekly) and between-scanner variation, the entire set of measurements was repeated on each scanner at both one-week and two-week intervals after the original measurement session. Between-run (quarterly or triannual) variation was assessed by repeating the measurement sessions every three to four months after the original session for one year; quarterly or triannual measurements were collected on five of the scanners (one GE LightSpeed VCT, one GE LightSpeed 16, and three Siemens Sensation 64 scanners). Table 3.1 summarizes the timing of all measurement sessions performed using each scanner.
Table 3.1. Schedule of weekly and quarterly (on Siemens scanners at site B) or triannual (on GE scanners at site A) data collection on all nine CT scanner units. "✓" indicates that a measurement session was performed on that unit at that relative time period, while "-" indicates that no session was performed. Some cells are grouped to illustrate the data sampling used to calculate between-run (weekly) (cells grouped with dashed oval), between-run (triannual) (cells shaded gray), and between-scanner (cells grouped with solid oval) individual %CV values.

<table>
<thead>
<tr>
<th>Scanner make and model</th>
<th>Unit ID</th>
<th>Time of measurement session</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Week 1</td>
</tr>
<tr>
<td>GE LightSpeed VCT</td>
<td>A</td>
<td>✓</td>
</tr>
<tr>
<td></td>
<td>B</td>
<td>✓</td>
</tr>
<tr>
<td></td>
<td>C</td>
<td>✓</td>
</tr>
<tr>
<td>GE LightSpeed 16</td>
<td>A</td>
<td>✓</td>
</tr>
<tr>
<td></td>
<td>B</td>
<td>✓</td>
</tr>
<tr>
<td></td>
<td>C</td>
<td>✓</td>
</tr>
<tr>
<td>Siemens Sensation 64</td>
<td>A</td>
<td>✓</td>
</tr>
<tr>
<td></td>
<td>B</td>
<td>✓</td>
</tr>
<tr>
<td></td>
<td>C</td>
<td>✓</td>
</tr>
</tbody>
</table>
3.2.1.1. **CALCULATED $\text{CTDI}_{100,\text{air}}$ VALUES**

At the beginning of each measurement session, the active detection volume of the ionization chamber was suspended free-in-air at or above the geometrical center of the gantry bore as shown in Figure 3.1; the central hole of a 16-cm CTDI phantom, which was placed in the patient head holder, was used to suspend the chamber. After aligning the chamber, within-run precision was characterized by collecting 20 consecutive free-in-air exposure measurements over several minutes on each scanner using the techniques described in Table 3.2 in axial scan mode. The delay between repeated exposures was a few seconds (the time necessary to acquire and record an individual reading). The free-in-air setup was maintained and exposures were collected using various bowties, kVps, and collimations; collimation encompasses both the number of active channels and the channel width. Measurements were collected at a fixed collimation while varying the kVp and at various collimations at a fixed kVp of 120, as described in Table 3.3. All exposure measurements were converted to $\text{CTDI}_{100,\text{air}}$ using Eq. (1.4).
FIGURE 3.1. (a) Setup for CTDI measurements as measured by a 10-cm pencil ionization chamber suspended free-in-air at isocenter from the central hole in a 16-cm CTDI phantom; (b) the chamber was aligned at isocenter using scanner's laser positioning lights.
TABLE 3.2. Techniques used on each make and model of scanner to obtain 20 repeated free-in-air CTDI measurements using a 10-cm pencil ionization chamber suspended at isocenter. All scans were performed in axial mode; nominal beam width can be calculated by multiplying the number of channels by the channel width.

<table>
<thead>
<tr>
<th>Scanner make and model</th>
<th>kVp</th>
<th>mA</th>
<th>Exposure time (s)</th>
<th>No. of channels</th>
<th>Channel width (mm)</th>
<th>Bowtie</th>
</tr>
</thead>
<tbody>
<tr>
<td>GE LightSpeed VCT</td>
<td>120</td>
<td>350</td>
<td>1</td>
<td>64</td>
<td>0.625</td>
<td>Large</td>
</tr>
<tr>
<td>GE LightSpeed 16</td>
<td>120</td>
<td>250</td>
<td>1</td>
<td>16</td>
<td>1.25</td>
<td>Large</td>
</tr>
<tr>
<td>Siemens Sensation 64</td>
<td>120</td>
<td>350</td>
<td>1</td>
<td>24</td>
<td>1.2</td>
<td>Large</td>
</tr>
</tbody>
</table>
TABLE 3.3. kVp, type of bowtie filter, and collimation variations used on each make and model of scanner to gather free-in-air CTDI measurements; axial scan mode and 1-s exposure times were used for all techniques. A total of 21, 12, and seven techniques were tested on each of the three GE LightSpeed VCT, GE LightSpeed 16, and Siemens Sensation 64 units, respectively.

<table>
<thead>
<tr>
<th>Scanner make and model</th>
<th>kVp</th>
<th>mA·s</th>
<th>No. of channels</th>
<th>Channel width (mm)</th>
<th>Bowtie</th>
</tr>
</thead>
<tbody>
<tr>
<td>GE LightSpeed VCT</td>
<td>80, 100, 140</td>
<td>350</td>
<td>64</td>
<td>0.625</td>
<td>Small, medium, large</td>
</tr>
<tr>
<td></td>
<td>120</td>
<td>350</td>
<td>8, 16, 32, 64</td>
<td>0.625</td>
<td>Small, medium, large</td>
</tr>
<tr>
<td>GE LightSpeed 16</td>
<td>80</td>
<td>350</td>
<td>16</td>
<td>1.25</td>
<td>Small, large</td>
</tr>
<tr>
<td></td>
<td>100, 140</td>
<td>250</td>
<td>16</td>
<td>1.25</td>
<td>Small, large</td>
</tr>
<tr>
<td></td>
<td>120</td>
<td>250</td>
<td>4, 8, 16</td>
<td>1.25</td>
<td>Small, large</td>
</tr>
<tr>
<td>Siemens Sensation 64</td>
<td>80, 100, 120, 140</td>
<td>350</td>
<td>24</td>
<td>1.2</td>
<td>Large</td>
</tr>
<tr>
<td></td>
<td>120</td>
<td>350</td>
<td>32</td>
<td>0.6</td>
<td>Large</td>
</tr>
<tr>
<td></td>
<td>120</td>
<td>350</td>
<td>1, 2</td>
<td>5</td>
<td>Large</td>
</tr>
</tbody>
</table>
3.2.1.2. **Calculated HVL and QVL**

The free-in-air CTDI setup was maintained and this test was performed in service mode so that the x-ray tube could be held stationary (i.e., in a nonrotational mode with the x-ray tube positioned at a fixed angle) at 180°, which corresponds to the bottom of the gantry; this setup has been shown to be an appropriate method for determining HVL. Next, sheets of type 1100 aluminum alloy were placed at the bottom of the gantry so that they directly covered the x-ray tube output port, as shown in Figure 3.2. Finally, the measured-exposure versus aluminum-thickness data for each combination of technical factors described in Table 3.4 were interpolated (using Eqs. [8.10] - [8.13] in the appendix to calculate HVL and QVL).

**Figure 3.2.** To determine HVL and QVL, an ionization chamber (circle) was suspended free-in-air at isocenter and sheets of 1100 aluminum alloy (arrow) were placed at the bottom of the scanner gantry covering the x-ray beam, which was held stationary at 180°.
Table 3.4. HVLs and QVLs were found for various kVps and types of bowties on each make and model of scanner; axial scan mode and 1-s exposure times were used for all techniques. Six, five, and four techniques were tested on each of the three GE LightSpeed VCT, GE LightSpeed 16, and Siemens Sensation 64 units, respectively.

<table>
<thead>
<tr>
<th>Scanner make and model</th>
<th>kVp</th>
<th>mA·s</th>
<th>No. of channels</th>
<th>Channel width (mm)</th>
<th>Bowtie</th>
</tr>
</thead>
<tbody>
<tr>
<td>GE LightSpeed VCT</td>
<td>80, 100, 140</td>
<td>350</td>
<td>64</td>
<td>0.625</td>
<td>Large</td>
</tr>
<tr>
<td></td>
<td>120</td>
<td>350</td>
<td>64</td>
<td>0.625</td>
<td>Small, medium, large</td>
</tr>
<tr>
<td>GE LightSpeed 16</td>
<td>80</td>
<td>350</td>
<td>16</td>
<td>1.25</td>
<td>Large</td>
</tr>
<tr>
<td></td>
<td>100, 140</td>
<td>250</td>
<td>16</td>
<td>1.25</td>
<td>Large</td>
</tr>
<tr>
<td></td>
<td>120</td>
<td>250</td>
<td>16</td>
<td>1.25</td>
<td>Small, large</td>
</tr>
<tr>
<td>Siemens Sensation 64</td>
<td>80, 100, 120, 140</td>
<td>350</td>
<td>24</td>
<td>1.2</td>
<td>Large</td>
</tr>
</tbody>
</table>
3.2.1.3. **CALCULATED CTDI\(_w\)**

Exposure measurements were collected inside the two standard sizes of CTDI phantoms (i.e., 16-cm and 32-cm); the 16-cm phantom was placed in the patient head holder, while the 32-cm phantom was placed directly on the patient table top. Both phantoms were positioned such that their central and 12:00 peripheral chamber locations were aligned with the scanner's sagittal and coronal laser lights (see Fig. 3.3). Following standard methods, exposure readings were recorded with the ionization chamber placed inside both the central and 12:00 peripheral chamber holes; PMMA filler rods were placed in the four empty chamber holes.\(^{135,136}\) All measurements were collected for a single axial rotation using the techniques described in Table 3.5. CTDI\(_w\) was calculated for both CTDI phantoms from the peripheral and central exposure measurements using Eq. (1.5).

**FIGURE 3.3.** The central chamber holes of both a (a) 16-cm and (b) 32-cm CTDI phantom were aligned to gantry isocenter using the scanner's laser positioning lights. Subsequently, measurements were collected at the central and 12:00 peripheral chamber holes in both phantoms.
TABLE 3.5. Techniques employed by each make and model of scanner for CDTI phantom measurements; axial scan mode and 1-s exposure times were used for all techniques. All technical factors, except for the bowtie filter used by the GE scanners, were consistent between the 16-cm and 32-cm phantoms. Four techniques were tested per CTDI phantom using each scanner make and model, with the exception of the GE LightSpeed VCT, where two bowties and thus a total of eight techniques were tested for the 16-cm CTDI phantom.

<table>
<thead>
<tr>
<th>Scanner make and model</th>
<th>kVp</th>
<th>mA-s</th>
<th>No. of channels</th>
<th>Channel width (mm)</th>
<th>Bowtie (16-cm CTDI phantom)</th>
<th>Bowtie (32-cm CTDI phantom)</th>
</tr>
</thead>
<tbody>
<tr>
<td>GE LightSpeed VCT</td>
<td>80, 100, 120, 140</td>
<td>350</td>
<td>64</td>
<td>0.625</td>
<td>Small, medium</td>
<td>Large</td>
</tr>
<tr>
<td></td>
<td>80</td>
<td>350</td>
<td>16</td>
<td>1.25</td>
<td>Small</td>
<td>Large</td>
</tr>
<tr>
<td>GE LightSpeed 16</td>
<td>80</td>
<td>350</td>
<td>16</td>
<td>1.25</td>
<td>Small</td>
<td>Large</td>
</tr>
<tr>
<td></td>
<td>100, 120, 140</td>
<td>250</td>
<td>16</td>
<td>1.25</td>
<td>Small</td>
<td>Large</td>
</tr>
<tr>
<td>Siemens Sensation 64</td>
<td>80, 100, 120, 140</td>
<td>350</td>
<td>24</td>
<td>1.2</td>
<td>Large</td>
<td>Large</td>
</tr>
</tbody>
</table>
3.2.2. Statistical Analysis

Precision of the dosimetry data was characterized by %CVs, which were calculated in a two-tier process. In the lower tier, data were grouped according to the type of precision being characterized [i.e., between-run (weekly) precision] and individual %CV values were calculated from this data. Table 3.1 shows the data grouping used to calculate individual %CV values for each type of precision; within-run precision is not shown in the table because individual %CV values were calculated from 20 repeated CTDI$_{100, \text{air}}$ values, which were collected on a single scanner during a single session. Because scanner model and technical factors have a known effect on absolute exposure, individual %CV values were always calculated from measurements collected at the same technical factors (bowtie filter, kVp, and collimation) on the same make and model of scanner.\textsuperscript{71}

In the upper tier, individual %CVs were pooled within each precision type [i.e., between-run (weekly) precision], scanner model, and dosimetry metric (i.e., CTDI$_{100, \text{air}}$) to calculate the median and range. The median and range of %CV values were reported rather than an average and confidence interval because the presence of potential outliers was detected within the pooled data sets.\textsuperscript{138,139}

The following example illustrates the process that was employed to calculate the %CV values reported in this study. Between-run (weekly) precision of CTDI$_{100, \text{air}}$ values calculated from exposure measurements made on the GE LightSpeed 16 scanners was determined by computing individual %CV values from three CTDI$_{100, \text{air}}$ data points collected at a given technique and each
gathered one week apart on a given scanner unit. Because data were collected for 12 techniques per scanner on three scanners, a total of 36 individual between-run (weekly) %CV values were calculated. These 36 %CV values were then pooled to determine the median and range of between run (weekly) precision of free-in-air CTDI values collected on GE LightSpeed 16 scanners. Similarly, between-run (quarterly and triannual) individual %CVs were calculated using five quarterly or four triannual dosimetry values and between-scanner individual %CVs were calculated from dosimetry values collected on three “sibling” scanners of the same model; individual %CVs were then pooled across technique and “sibling” scanners (for between-run precision) or measurement sessions (for between-scanner precision).

Pooling data across techniques and either “sibling” scanners or measurement sessions better represents overall scanner performance than assessing the precision of data collected at one technique on one scanner during one session, which could overestimate or underestimate true performance. Additionally, pooling across these factors increases the degrees of freedom and thus the statistical validity of the results. Although precision error could potentially vary from technique-to-technique, scanner-to-scanner, and session-to-session, the purpose of this study was to characterize the general performance of the scanners over a broad range of conditions rather than to develop a precision profile.
3.3. RESULTS

Over 850 individual %CV values were calculated, which ranged from 0.00% to 4.33% across all dosimetry metrics, precision types, and scanners included in this study; over 95% of the calculated %CVs were less than 2.75%.

Free-in-air CTDI calculated values and the corresponding precision results, respectively, appear in Table 3.6 and 3.7 for each scanner make and model; because different current-exposure time products were used across scanner types, all CTDI\textsubscript{100, air} values were normalized to 100 mA·s. Table 3.6 shows a large spread of CTDI\textsubscript{100, air} values (5.8 - 46.7 mGy/100 mA·s), which were influenced by the scanner model and technical parameters selected. Table 3.7 shows intrascanner and interscanner variation in CTDI\textsubscript{100, air} values to be extremely low, with all %CVs less than 5%. One subtle trend emerged from these results, which was that in most cases, the scanners displayed progressively worse intrascanner precision as the between-run time increased from a few seconds to a week to several months.
**TABLE 3.6.** Mean and standard deviation, in parentheses, of CTDI_{100, air} calculated values; "-" indicates the scanner either does not offer that bowtie option or that beam width. CTDI_{100, air} values were normalized to 100 mA·s to account for differences in scanning protocols used on the different scanner makes and models.

<table>
<thead>
<tr>
<th>kVp</th>
<th>Beam width (mm)</th>
<th>Bowtie</th>
<th>CTDI_{100, air} (mGy/100 mA·s)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td>GE LightSpeed VCT</td>
</tr>
<tr>
<td>80</td>
<td>20, 40</td>
<td>Small</td>
<td>12.4 (0.3)</td>
</tr>
<tr>
<td>80</td>
<td>40</td>
<td>Medium</td>
<td>12.4 (0.3)</td>
</tr>
<tr>
<td>80</td>
<td>20, 28.8, 40</td>
<td>Large</td>
<td>8.9 (0.2)</td>
</tr>
<tr>
<td>100</td>
<td>20, 40</td>
<td>Small</td>
<td>20.4 (0.4)</td>
</tr>
<tr>
<td>100</td>
<td>40</td>
<td>Medium</td>
<td>20.3 (0.4)</td>
</tr>
<tr>
<td>100</td>
<td>20, 28.8, 40</td>
<td>Large</td>
<td>15.6 (0.2)</td>
</tr>
<tr>
<td>120</td>
<td>5</td>
<td>Small</td>
<td>46.7 (1.4)</td>
</tr>
<tr>
<td>120</td>
<td>10</td>
<td>Small</td>
<td>37.3 (0.7)</td>
</tr>
<tr>
<td>120</td>
<td>20</td>
<td>Small</td>
<td>31.4 (0.4)</td>
</tr>
<tr>
<td>120</td>
<td>40</td>
<td>Small</td>
<td>29.5 (0.5)</td>
</tr>
<tr>
<td>120</td>
<td>5</td>
<td>Medium</td>
<td>46.7 (1.5)</td>
</tr>
<tr>
<td>120</td>
<td>10</td>
<td>Medium</td>
<td>37.3 (0.8)</td>
</tr>
<tr>
<td>120</td>
<td>20</td>
<td>Medium</td>
<td>31.4 (0.5)</td>
</tr>
<tr>
<td>120</td>
<td>40</td>
<td>Medium</td>
<td>29.4 (0.5)</td>
</tr>
<tr>
<td>120</td>
<td>5</td>
<td>Large</td>
<td>37.7 (1.2)</td>
</tr>
<tr>
<td>120</td>
<td>10</td>
<td>Large</td>
<td>30.1 (0.5)</td>
</tr>
<tr>
<td>120</td>
<td>19.2</td>
<td>Large</td>
<td>-</td>
</tr>
<tr>
<td>120</td>
<td>20</td>
<td>Large</td>
<td>25.3 (0.3)</td>
</tr>
<tr>
<td>120</td>
<td>28.8</td>
<td>Large</td>
<td>-</td>
</tr>
<tr>
<td>120</td>
<td>40</td>
<td>Large</td>
<td>23.8 (0.3)</td>
</tr>
<tr>
<td>140</td>
<td>20, 40</td>
<td>Small</td>
<td>39.7 (0.6)</td>
</tr>
<tr>
<td>140</td>
<td>40</td>
<td>Medium</td>
<td>39.6 (0.6)</td>
</tr>
<tr>
<td>140</td>
<td>20, 28.8, 40</td>
<td>Large</td>
<td>33.1 (0.4)</td>
</tr>
</tbody>
</table>

*At 80 kVp, 100 kVp, and 140 kVp, nominal beam widths of 20 mm, 28.8 mm, and 40 mm were tested exclusively on the GE LightSpeed 16, Siemens Sensation 64, and GE LightSpeed VCT scanners, respectively.*
TABLE 3.7. Median and range, in parentheses, of measurement precision of CTDI$_{100}$, HVL, QVL, and CTDI$_w$ (for a 16-
cm and 32-cm CTDI phantom) calculated values. Within-run precision was only evaluated for CTDI$_{100}$, air data (collected
using the techniques that appear in Table 3.2), thus “- ” is shown for all other dosimetry metrics. Between-run and
between-scanner individual %CVs were calculated from data collected at a given technique (shown in Table 3.3, 3.4, and
3.5 for CTDI$_{100}$, air, HVL and QVL, and CTDI$_w$, respectively) on the same make and model of scanner.

<table>
<thead>
<tr>
<th>Dosimetry metric</th>
<th>Scanner make and model</th>
<th>Intrascanner precision, CV (%)</th>
<th>Interscanner precision, CV (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Within-run (week)</td>
<td>Between-run (quarterly/triannual)* &lt;/p&gt;</td>
</tr>
<tr>
<td>CTDI$_{100}$, air</td>
<td>GE LightSpeed VCT</td>
<td>0.05 (0.04 - 0.11)</td>
<td>0.52 (0.10 - 0.76)</td>
</tr>
<tr>
<td></td>
<td>GE LightSpeed 16</td>
<td>0.08 (0.06 - 0.10)</td>
<td>0.12 (0.01 - 1.13)</td>
</tr>
<tr>
<td></td>
<td>Siemens Sensation 64</td>
<td>0.19 (0.00 - 0.86)</td>
<td>0.47 (0.25 - 0.63)</td>
</tr>
<tr>
<td>HVL</td>
<td>GE LightSpeed VCT</td>
<td>-</td>
<td>0.19 (0.02 - 1.86)</td>
</tr>
<tr>
<td></td>
<td>GE LightSpeed 16</td>
<td>-</td>
<td>0.22 (0.08 - 1.46)</td>
</tr>
<tr>
<td></td>
<td>Siemens Sensation 64</td>
<td>-</td>
<td>0.24 (0.06 - 0.75)</td>
</tr>
<tr>
<td>QVL</td>
<td>GE LightSpeed VCT</td>
<td>-</td>
<td>0.37 (0.05 - 0.76)</td>
</tr>
</tbody>
</table>
### TABLE 3.7 continued

<table>
<thead>
<tr>
<th>Dosimetry metric</th>
<th>Scanner make and model</th>
<th>Intrascanner precision, CV (%)</th>
<th>Interscanner precision, CV (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Within-run</td>
<td>Between-run (weekly)</td>
</tr>
<tr>
<td>QVL</td>
<td>GE LightSpeed 16</td>
<td>-</td>
<td>0.30</td>
</tr>
<tr>
<td></td>
<td>Siemens Sensation 64</td>
<td>-</td>
<td>(0.08 - 0.72)</td>
</tr>
<tr>
<td>CTDIw (16-cm CTDI phantom)</td>
<td>GE LightSpeed VCT</td>
<td>-</td>
<td>0.19</td>
</tr>
<tr>
<td></td>
<td>GE LightSpeed 16</td>
<td>-</td>
<td>(0.02 - 0.83)</td>
</tr>
<tr>
<td></td>
<td>Siemens Sensation 64</td>
<td>-</td>
<td>0.70</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>(0.11 - 1.70)</td>
</tr>
<tr>
<td>CTDIw (32-cm CTDI phantom)</td>
<td>GE LightSpeed VCT</td>
<td>-</td>
<td>0.38</td>
</tr>
<tr>
<td></td>
<td>GE LightSpeed 16</td>
<td>-</td>
<td>(0.08 - 1.06)</td>
</tr>
<tr>
<td></td>
<td>Siemens Sensation 64</td>
<td>-</td>
<td>1.14</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>(0.88 - 3.77)</td>
</tr>
<tr>
<td></td>
<td>GE LightSpeed VCT</td>
<td>-</td>
<td>1.19</td>
</tr>
<tr>
<td></td>
<td>GE LightSpeed 16</td>
<td>-</td>
<td>(0.39 - 2.53)</td>
</tr>
<tr>
<td></td>
<td>Siemens Sensation 64</td>
<td>-</td>
<td>1.23</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>(0.34 - 2.29)</td>
</tr>
<tr>
<td></td>
<td>GE LightSpeed 16</td>
<td>-</td>
<td>0.62</td>
</tr>
<tr>
<td></td>
<td>Siemens Sensation 64</td>
<td>-</td>
<td>(0.24 - 1.43)</td>
</tr>
</tbody>
</table>

*Between-run (quarterly) precision was assessed for the Siemens scanners, which were tested at site B; between-run (triannual) precision was assessed for both models of GE scanners tested at site A.
The calculated HVLs and QVLs and their precisions appear in Table 3.8 and 3.7, respectively. As shown in Table 3.8, the mean HVLs ranged from 5.4 mm to 9.7 mm of aluminum and the mean QVLs ranged from 12.0 mm to 21.1 mm of aluminum, depending on the scanner model and technical factors tested. Table 3.7 shows extremely low short- and long-term intrascanner and interscanner variability in the HVLs and QVLs, with all median and individual %CVs less than or equal to 1.04% and 2.74%, respectively.

**Table 3.8.** Mean and standard deviation, in parentheses, of the calculated HVLs and QVLs (in mm of aluminum); "-" indicates that the scanner does not offer that bowtie option.

<table>
<thead>
<tr>
<th>kVp</th>
<th>Bowtie</th>
<th>HVL (mm AI)</th>
<th>QVL (mm AI)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>GE LightSpeed VCT</td>
<td>GE LightSpeed 16</td>
</tr>
<tr>
<td>80</td>
<td>Large</td>
<td>5.4 (0.09)</td>
<td>5.9 (0.07)</td>
</tr>
<tr>
<td>100</td>
<td>Large</td>
<td>6.6 (0.05)</td>
<td>7.2 (0.05)</td>
</tr>
<tr>
<td>120</td>
<td>Small</td>
<td>6.6 (0.03)</td>
<td>7.2 (0.06)</td>
</tr>
<tr>
<td>120</td>
<td>Medium</td>
<td>6.6 (0.01)</td>
<td>-</td>
</tr>
<tr>
<td>120</td>
<td>Large</td>
<td>7.7 (0.05)</td>
<td>8.3 (0.09)</td>
</tr>
<tr>
<td>140</td>
<td>Large</td>
<td>8.6 (0.14)</td>
<td>9.2 (0.04)</td>
</tr>
<tr>
<td>80</td>
<td>Large</td>
<td>12.0 (0.10)</td>
<td>13.0 (0.10)</td>
</tr>
<tr>
<td>100</td>
<td>Large</td>
<td>15.0 (0.11)</td>
<td>16.0 (0.11)</td>
</tr>
<tr>
<td>120</td>
<td>Small</td>
<td>15.4 (0.11)</td>
<td>16.6 (0.09)</td>
</tr>
<tr>
<td>120</td>
<td>Medium</td>
<td>15.5 (0.08)</td>
<td>-</td>
</tr>
<tr>
<td>120</td>
<td>Large</td>
<td>17.5 (0.10)</td>
<td>18.5 (0.10)</td>
</tr>
<tr>
<td>140</td>
<td>Large</td>
<td>19.8 (0.13)</td>
<td>20.7 (0.08)</td>
</tr>
</tbody>
</table>
The calculated CTDI\textsubscript{w} values (normalized to 100 mAs) for the 16-cm and 32-cm CTDI phantoms and their relative precision errors appear in Table 3.9 and 3.7, respectively. Table 3.9 shows the range of CTDI\textsubscript{w} values obtained from both the 16-cm (3.9 – 28.2 mGy/100 mAs) and 32-cm (1.7 – 12.7 mGy/100 mAs) CTDI phantoms. Table 3.7 shows that, with a few exceptions, the median %CV values (0.38 – 2.79%) were slightly higher compared to the other dosimetry metrics (CTDI\textsubscript{100, air}, HVL, and QVL). Despite the slightly higher (in general) relative precision errors observed in this portion of the study, the results still demonstrated extremely low variability, with all %CV values below 5%.
TABLE 3.9. Mean and standard deviation, in parentheses, of calculated CTDI\(_w\) values for a 16-cm and 32-cm CTDI phantom; "-" indicates that the scanner does not offer that bowtie option. CTDI\(_w\) values were normalized to 100 mA·s to account for differences in scanning protocols used on the different scanner models.

<table>
<thead>
<tr>
<th>kVp</th>
<th>Bowtie</th>
<th>CTDI(_w) (mGy/100 mA·s) for a 16-cm CTDI phantom</th>
<th>Siemens Sensation 64</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>GE LightSpeed VCT</td>
<td>GE LightSpeed 16</td>
</tr>
<tr>
<td>80</td>
<td>Small</td>
<td>6.6 (0.1)</td>
<td>6.4 (0.1)</td>
</tr>
<tr>
<td>80</td>
<td>Medium</td>
<td>7.6 (0.2)</td>
<td>-</td>
</tr>
<tr>
<td>80</td>
<td>Large</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>100</td>
<td>Small</td>
<td>11.9 (0.2)</td>
<td>11.6 (0.2)</td>
</tr>
<tr>
<td>100</td>
<td>Medium</td>
<td>13.5 (0.3)</td>
<td>-</td>
</tr>
<tr>
<td>100</td>
<td>Large</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>120</td>
<td>Small</td>
<td>18.1 (0.3)</td>
<td>17.7 (0.3)</td>
</tr>
<tr>
<td>120</td>
<td>Medium</td>
<td>20.4 (0.4)</td>
<td>-</td>
</tr>
<tr>
<td>120</td>
<td>Large</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>140</td>
<td>Small</td>
<td>25.1 (0.3)</td>
<td>24.7 (0.4)</td>
</tr>
<tr>
<td>140</td>
<td>Medium</td>
<td>28.2 (0.4)</td>
<td>-</td>
</tr>
<tr>
<td>140</td>
<td>Large</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>CTDI(_w) (mGy/100 mA·s) for a 32-cm CTDI phantom</th>
</tr>
</thead>
<tbody>
<tr>
<td>80</td>
</tr>
<tr>
<td>100</td>
</tr>
<tr>
<td>120</td>
</tr>
<tr>
<td>140</td>
</tr>
</tbody>
</table>
3.4. DISCUSSION

The dosimetry data collected in this study yielded very high measurement precision across all conditions, time intervals, and scanners. The vast majority (over 95%) of the large amount of %CV values analyzed and reported in this study were below 2.75%, with only a few reaching 4%. Though not all existing MDCT scanners were tested, three different scanner models, which consisted of scanners from two manufacturers and two models from the same manufacturer, were evaluated. To further generalize the results of this study, two testing locations were included. In all scenarios, both intrascanner and interscanner relative precision errors were extremely low.

This study’s results indicate that dose estimates obtained from MC model simulations can be validated if they agree with physical data at a level of 5% or less; however, the level of acceptable mismatch is probably greater than 5% because measurement precision error is presumably not the only source of disagreement between the model and physical data. In order to develop more accurate validation criteria, several factors not addressed in this study should be considered. For example, inherent precision of the MC model, which can be affected by the number of photons simulated, should be taken into account during benchmarking experiments. Additionally, the dosimetry-related quantities considered in this study have different roles in MC equivalent source models which may influence error estimates. Specifically, HVL and QVL are parameters input into MC equivalent source models and therefore precision error in these
values propagates through the model. Precision error associated with the free-in-air exposures, which convert MC output to dose, does not propagate through the model itself but does lead to variability in the absolute dose estimates obtained and thus error or uncertainty bars should be included with reported dose estimates.

Dosimetry measurements are collected on CT scanners for various reasons other than for benchmarking MC dose estimates. For example, such results serve as a reference for CT facility management to show that, in light of the exceptional stability of modern CT scanners, frequent dosimetry testing is unnecessary for avoiding overexposure incidents. Therefore, the results of this study also have implications for scanner compliance testing, as well as, any other applications that utilize scanner output (dosimetry) values.

To the best of the authors’ knowledge, no prior study has addressed the precision of measuring dose output from state-of-the-art MDCT scanners. Despite the strength of this study, sources of variation or error were present, including variation in scanner output, ionization chamber and electrometer imprecision, and setup variability between sessions and scanners. Scanner workloads varied from 160 to 164, 126 to 156, and 119 to 204 patients per week (on average over the time of the study) on the GE LightSpeed VCT, GE LightSpeed 16, and Siemens Sensation 64 scanners, respectively. Furthermore, at site B, one scanner had an x-ray tube change during the course of the study. At site A, a different ionization chamber and electrometer were used during the
third triannual measurement session on the GE LightSpeed VCT scanner. Additionally, at both sites, the ionization chambers and electrometers were calibrated prior to the study and were not recalibrated during the study and so drift was possible. Although the same experimenters were maintained at sites A and B throughout the course of the study, setup variability was unavoidable. To minimize setup variability at site A, centering of the ionization chamber and CTDI phantoms was verified by generating 1.25-mm thick images and then using a caliper tool to check the images for centering within at least 3 mm along both the x and y directions in the transverse plane (in the majority of cases, centering was within 1 mm). The scanners at Site B were unable to identify gantry isocenter on acquired CT images; therefore, only the scanners’ built-in positioning laser lights were utilized to center the chamber and phantoms.

Care should be taken when directly comparing the results reported in this study across scanner makes and models. When considering comparison of the %CV values, it is important to keep in mind the differences in testing locations and experimenters between sites A and B. Similarly, direct comparisons of system output (CTDI<sub>100, air</sub>, HVL, QVL, and CTDI<sub>1w</sub> values) across scanner models may be inappropriate. Although the same combination of technical factors resulted in measurable dose differences between scanner models, image quality was not assessed in this study and instead emphasis was placed on assessing precision. In order to properly characterize differences in dose performance between scanners, dose should be quantified while maintaining the same level of image quality across scanner models.
Despite the limitations of this study, the results revealed extremely low levels of variation in scanner output and dosimetry-related quantities across time and across “sibling” scanners of the same make and model. These results point to the exceptional similarity in “sibling” scanners and stability of the x-ray source output in the state-of-the-art MDCT scanners tested in this study over both very short periods of time and in time periods tested up to one year.

3.5. CONCLUSION

This study investigated the variability of CT scanner output across a range of time periods (from as short as a few seconds to as long as several months), technical parameters (kVp, bowtie filter, and x-ray beam collimation), and “sibling” scanners of three different makes and models. The results of this study demonstrated that across all conditions, the MDCT scanners tested produced dosimetry-related data that was precise (well within 5%) for all time periods tested over the course of one year. These results can be applied in MC modeling to improve upon and verify current MC models, thus improving the accuracy of patient dose estimates.
CHAPTER 4
RADIATION DOSE PENALTY IN AXIAL MODE CT

**Purpose:** A 10-millisecond (ms) rise and stabilization of the x-ray tube output occurs immediately prior to image acquisition on GE Healthcare CT scanners and is believed to contribute additional dose to anatomy facing the tube start acquisition angle. This dose penalty would presumably have a greater effect in axial scanning because the rise and stabilization time occurs for every 360° rotation of the x-ray tube, whereas it only occurs at the beginning of each pass in helical scan mode. The purpose of this study was to characterize the dose penalty due to the rise and stabilization time in axial scan mode.

**Methods:** To quantify the dose penalty associated with rise and stabilization of the x-ray tube output, 10-cm, 16-cm, and 32-cm CTDI phantoms were scanned on a GE LightSpeed VCT scanner for a single axial rotation; a range of phantom sizes was used to account for possible variations in the dose penalty across patients of different sizes. Exposure was detected using a 10-cm pencil ionization chamber placed in the 12:00 peripheral chamber hole of each phantom. For various x-ray tube start acquisition angles, which were specified in scanner's service mode, exposure was recorded and then converted to CTDI. Additionally, scan factors, including phantom centering and rotation time, were varied to quantify their effect on the dose penalty.
RESULTS: In general, the dose penalty due to the 10-ms rise and stabilization time increased as the rotation time decreased and also as the ionization chamber was moved closer towards the top of the gantry (either by increasing the phantom diameter or by moving the phantom vertically off-center). For 1-s rotations, CTDI\textsubscript{100, 12:00} was 1.0%, 1.6%, and 4.4% higher in the 10-cm, 16-cm, and 32-cm CTDI phantom, respectively, when a start angle of approximately 0° was used versus a start angle of 270°. The dose penalty in the 32-cm CTDI phantom increased to 26% when using 0.4-s rotations and moving the phantom vertically off-center by 12 cm. These observations were consistent with expectations based primarily on the geometry of the scanner and the setup was.

CONCLUSION: In light of the dose penalty observed in this study, all acquisitions (especially axial acquisitions) should feature a start angle of 180° to avoid imparting this dose penalty on superficial radiosensitive organs (e.g., breast, testes, and thyroid), which are more prevalent on a patient's anterior.

4.1. INTRODUCTION

In axial scan mode, the x-ray tube rotates around one section of the patient while the patient table is stationary. To image the next section, the patient table either moves forward or backward and another rotation is performed. In helical scanning, the x-ray tube continuously rotates as the patient table simultaneously moves through the scanner. Rather than tracing consecutive
rings around the patient (as in axial mode), the x-ray beam traces a spiral path in helical mode. In order to reconstruct the spiral data into planar images up to the specified image length, additional helical rotations are acquired at the beginning and end of a pass for interpolation. This is called z-axis overranging and refers to the difference between the user-specified image length (as entered on the scanner console) and the actual scan length including these additional rotations. The additional rotations performed during helical scanning result in a dose penalty compared to axial acquisitions. MC simulations of an MDCT scanner have estimated that overranging produces up to 35.8% higher doses in helical chest scans (pitch = 1) of a mathematical adult anthropomorphic phantom than in axial scans simulated at the same technique; even greater MDCT dose penalties (up to 70%) were observed in mathematical pediatric anthropomorphic phantoms.

Many studies have explored the effect of CT scan parameters (e.g., kVp, mA·s, beam collimation, etc.) on radiation dose, but few have directly examined the dose effect of axial versus helical scan mode (outside the scope of z-axis overranging). Although contiguous axial and contiguous helical (i.e., pitch = 1) scan protocols using the same scan techniques have traditionally been considered to deliver equivalent radiation doses, several studies have revealed higher point doses in axial versus helical scanning. For instance, McDermott et al. observed organ doses an average of 13%, but as much as 25%, higher in a pediatric anthropomorphic phantom when scanned in axial versus helical mode (with the same kVp and effective mA·s [i.e., mA·s/pitch]) on
a GE LightSpeed VCT scanner.\textsuperscript{149} Even larger discrepancies, as high as 53\%, in axial versus helical doses were observed by Pitman et al. when scanning an adult anthropomorphic phantom on a SDCT scanner.\textsuperscript{151} McNitt-Gray et al. reported dose differences of between 3\% and 14\% in a 32-cm CTDI phantom scanned on a SDCT scanner using contiguous axial versus helical acquisitions.\textsuperscript{150}

Differences between axial and helical scanning may explain the observation of higher doses, despite using equivalent techniques, in axial mode compared to helical mode. Specifically, differences in the frequency of rise and fall time events across scan modes may lead to dose discrepancies. Prior to acquisition in either scan mode, the x-ray tube output rises from zero to reach maximum exposure; after acquisition is finished, the tube output falls back to zero. Since acquisition is continuous in helical mode, the rise time (at the beginning) and fall time (at the end) only happen once during each helical pass. In contrast, rising of the x-ray tube output occurs before each axial rotation; the same is true when the x-ray tube output falls to its off state. The exposure delivered during the rise and fall time contributes to patient dose but is not typically used for image reconstruction. This phenomenon has been referred to as overscanning, which means that the beam-on time extends beyond the data acquisition time.\textsuperscript{152} Overscanning due to the rise and fall, for example, would occur 20 more times when scanning 40-cm of a patient on a scanner with a 2-cm beam width in axial mode than during a single helical pass. The increased frequency of rise and fall time events may result in an inherent dose penalty in axial compared to helical scanning. However, the specific rise and fall time
durations and their dose contributions depend on the CT scanner make and model.

The purpose of this study was to quantify the dose penalty associated with overscanning in CTDI phantoms during axial scanning. Additionally, the effect of phantom size, phantom centering, and rotation time on this/the dose penalty was also quantified.

4.2. MATERIALS AND METHODS

4.2.1. PHANTOM MEASUREMENTS

All scans in this study were performed using a 64-channel MDCT scanner (LightSpeed VCT, GE Healthcare). Before data acquisitions on the GE LightSpeed VCT scanner, there is a 10-ms rise and stabilization time (i.e., the time required for the x-ray tube output to reach and remain stable at the prescribed value) that occurs for all scans regardless of the rotation time; fall time at the end of acquisition is negligible (<1 ms) (determined via personal communication with Jiang Hsieh of GE Healthcare; 11/30/2009). Therefore, for a single axial rotation, exposure is delivered during the rotation time as well as the 10 ms rise and stabilization time that occurs prior to data acquisition.

A kVp meter (Radcal 4083; RadCal Corporation) was used measure kVp versus time on a GE LightSpeed VCT scanner to characterize the rise and fall time durations. According to the LightSpeed VCT technical reference manual,
rise time is the time elapsed before attaining 75% of the selected high voltage 
value and fall time is the time required to fall below 75% of the selected 
voltage.\textsuperscript{153} Based on these definitions, the measured kVp versus time waveforms 
(data not shown) revealed that the high voltage rise and fall time lasted less than 
1 ms each. However, the measured exposure time (i.e., the time when kVp > 0) 
lasted between 1010 ms and 1011 ms for 1-s nominal exposure times. 
Therefore, the stabilization time, rather than the rise or fall time, constitutes the 
majority of the overscanning.

The dose penalty due to overscanning is expected to increase when the 
path length between the x-ray tube and the exposure measurement location is 
minimized. Simulations performed by Nickoloff et al. showed that dose at the 
12:00 position in a CTDI phantom was higher for a stationary x-ray tube 
positioned at 0° (i.e., the top of the gantry), which has the shortest path to 12:00, 
than at any other tube angle.\textsuperscript{154} Therefore, because overscanning occurs 
immediately prior to data acquisition, any additional dose contribution from 
overscanning would most likely be detected near the start acquisition angle of the 
x-ray tube. This was evaluated by measuring dose at a fixed position (e.g., 
12:00) inside CTDI phantoms while varying the x-ray tube start angle. 
Additionally, other factors affecting the path length between the x-ray tube and 
the dose measurement location, including the phantom diameter and positioning 
of the phantom within the gantry, were also varied. Furthermore, because the 
percentage of overscanning increases as the rotation time decreases, the dose 
effect of the rotation time was also quantified.
4.2.1.1. CTDI Phantoms at Isocenter

Exposure was detected using a 10-cm pencil ionization chamber connected to an electrometer (RadCal Corporation). The ionization chamber was placed at the 12:00 peripheral chamber position inside CTDI phantoms of three different diameters: 10-cm, 16-cm and 32-cm (see Fig. 4.1). Exposure was measured at 12:00 because it is the position inside CTDI phantoms that is most representative of the location of breast tissue in patients. The phantoms were placed either on the patient table (16-cm and 32-cm phantoms) or in the patient head holder (10-cm phantom) and their central chamber holes were aligned to isocenter.

**Figure 4.1.** 10-cm, 32-cm, and 16-cm CTDI phantoms were used in this study to assess the dose effect of overscanning in axial CT.
Exposure measurements were collected using various x-ray tube start acquisition angles, which were specified in the scanner’s service mode; Figure 4.2 depicts the 32-cm phantom scanned using a 45° start angle. The scans performed in service mode corresponded to single axial rotations (without patient table movement). Helical scanning was not be evaluated because the start acquisition angle could not be controlled in helical mode. Pediatric head and chest protocols were used to scan the 10-cm and 16-cm phantom, respectively, while an adult chest protocol was used to scan the 32-cm phantom; both 0.4-s and 1-s rotation times were used to scan the 16-cm and 32-cm phantoms. The scan techniques used for each phantom appear in Table 4.1.

**FIGURE 4.2.** The 32-cm CTDI phantom was scanned using various tube start acquisition angles; a start angle of 45° is shown in this illustration.
TABLE 4.1. Each CTDI phantom was scanned (in service mode) for a single axial rotation using the following technique(s).

<table>
<thead>
<tr>
<th>Phantom diameter (cm)</th>
<th>Bowtie filter</th>
<th>mA</th>
<th>Exposure time (s)</th>
<th>kVp</th>
<th>Focal spot size</th>
<th>Nominal beam width (mm)†</th>
</tr>
</thead>
<tbody>
<tr>
<td>10</td>
<td>Small</td>
<td>200</td>
<td>1</td>
<td>120</td>
<td>Small</td>
<td>40</td>
</tr>
<tr>
<td>16</td>
<td>Medium</td>
<td>290</td>
<td>0.4, 1</td>
<td>100</td>
<td>Small</td>
<td>40</td>
</tr>
<tr>
<td>32</td>
<td>Large</td>
<td>400</td>
<td>0.4, 1</td>
<td>120</td>
<td>Large</td>
<td>40</td>
</tr>
</tbody>
</table>

* The 10-cm phantom was placed in the head holder, while the 16-cm and 32-cm phantoms were placed directly on the patient tabletop.

† The detector configuration was 64 x 0.625 mm.

Eight exposures were collected using a start acquisition angle of 270° for each phantom; these eight measurements were averaged and then converted to CTDI using Eq. (1.4). The mean CTDI_{100, 12:00} for a 270° start angle was compared to the CTDIs calculated for all other start angles through their percent differences; 270° instead of 180° was used for comparison to avoid the effect of attenuation from the patient table. The %CV for the eight exposure measurements collected using a start acquisition angle of 270° was also calculated to characterize precision of the measurements.

4.2.1.2. CTDI PHANTOMS OFF-CENTER

After performing the experiments with the phantoms properly centered within the bore of the scanner gantry, we studied the effect of moving the
phantoms off-center. Off-centering was primarily evaluated using the 32-cm CTDI phantom because, of the three phantoms, it had the highest dose penalty due to overscanning. Although the dose penalty at isocenter was less substantial for the 16-cm phantom, because there is more opportunity for off-centering in smaller patients (who occupy less space in the gantry) the effect of off-centering was evaluated to a lesser extent using the 16-cm phantom.

The effect of vertical off-centering was assessed in the 32-cm CTDI phantom by measuring exposure at 12:00 with the phantom moved vertically off-center to the limits of the patient table’s range of motion (i.e., from 6 cm below to 12 cm above the table position where the phantom’s central chamber hole was aligned at isocenter) in 3-cm increments. Vertical off-centering was also assessed in the 16-cm phantom with the patient table moved 3 cm above the position where its central chamber hole was aligned at isocenter. Horizontal off-centering was emulated because the patient table did not allow for horizontal movement and it would have been difficult to move the phantom horizontally off-center in a controlled manner. Therefore, a second ionization chamber was placed at the 9:00 peripheral chamber position, allowing both vertical and emulated horizontal off-centering to be simultaneously evaluated by adjusting only the patient table height. As Figure 4.3 illustrates, this manner of emulating horizontal off-centering (i.e., placing an ionization chamber at 9:00 and vertically moving the patient table) is geometrically equivalent, except for the location of the patient table, to placing a chamber at 12:00 and horizontally moving the phantom.
FIGURE 4.3. The position of the 32-cm CTDI phantom in the gantry after the patient table was (a) raised and (b) lowered limits (in 3-cm increments) to its limits. The effect of both vertical and emulated horizontal off-centering on the dose penalty from overscanning were simultaneously evaluated by placing two 10-cm pencil ionization chambers at the phantom's 12:00 and 9:00 peripheral chamber positions, respectively.
The phantoms were scanned using the techniques listed in Table 4.1 and 0.4-s rotations were performed, except when the 32-cm phantom was moved 12 cm above isocenter, in which case, both 0.4-s and 1-s rotations were performed. Exposure measurements were collected using various start angles, converted to CTDI (using Eq. [1.4]), and then compared (through their percent difference) to the mean CTDI calculated from eight exposure measurements collected for a start angle of 270° (for vertical off-centering) or 90° (for emulated horizontal off-centering).

4.2.2. COMPUTER SIMULATION

A simple computer model was created in Matlab R2008b (MathWorks Inc.) to better understand the results, specifically, why we observed percent differences in the CTDIs that were greater than the percent contributions of overscanning to the total exposure times (i.e., 1% for a 1-s rotation or 2.5% for a 0.4-s rotation). In the computer model, the CTDI phantoms were represented as solid PMMA (i.e., the chamber holes were not specifically included) and the ionization chamber was represented as a point located 1 cm below the phantoms’ surfaces. The x-ray fan beam was modeled as originating from points located around the circumference of the scanner bore (which had a 70-cm diameter), with the beam’s central ray passing through gantry isocenter. “Exposure” was calculated for every central ray angle from 0° – 359° (in 1° increments) as the intensity, after accounting for attenuation through the phantom and the inverse square law, of the primary beam ray passing directly
through 12:00 (i.e., the point modeled as the ionization chamber). If a ray within the 50-cm SFOV did not pass through 12:00, as was the case for some central ray angles when the phantom was very far off-center, the intensity for that central ray angle was set to 0. The total "exposure" over a full rotation was calculated for every start acquisition angle as the area under the curve (using trapezoidal integration) from 0° – 360° plus the area under the curve from the angle where the rise and stabilization time began (e.g., for a 0.4-s rotation, the rise and stabilization time begins 9° before the start acquisition angle) to the start acquisition angle (i.e., the overscan region). The "exposures" by themselves were meaningless, and so their percent differences relative to the "exposure" for a simulated start angle of 270° were calculated and used to assess the dose penalties from overscanning in the same manner as for the measured data.

4.3. RESULTS

4.3.1. CTDI PHANTOMS AT ISOCENTER

The CTDIs calculated from exposures detected at the 12:00 peripheral chamber positions in a 10-cm, 16-cm, and 32-cm CTDI phantom are shown in Figure 4.4 for various x-ray tube start acquisition angles. The dose penalty for the start angle with the highest CTDI_{100, 12:00} relative to the mean CTDI_{100, 12:00} for a 270° start angle, increased from 1.0% to 1.6% to 4.4% as the phantom diameter increased from 10-cm to 16-cm to 32-cm, respectively. Although minor dose penalties were observed in the 10-cm and 16-cm phantom, these penalties were
considered to be meaningful given that the maximum %CV for exposures using a start angle of 270° was 0.11%. For all three phantoms, the peak CTDI\textsubscript{100, 12:00} was observed when using a start angle of approximately 0°; this observation was consistent with our expectations since the path lengths through the phantoms were shortest for 0° start angles. However, because overscanning occurs immediately prior to acquisition, the peak CTDI\textsubscript{100, 12:00} values were observed for start angles that were slightly shifted from 0°. The peak CTDI\textsubscript{100, 12:00} occurs when the overscan region is centered at 0° rather than when using a start acquisition angle of 0°; thus, for 1-s rotations, the 10-ms of overscanning begins approximately 4° before acquisition, and so the peak CTDI\textsubscript{100, 12:00} values were observed when using start acquisition angles of approximately 2°. As follows, when 0.4-s rotations were used, the peak CTDI\textsubscript{100, 12:00} was observed when the start acquisition angle was further shifted away from 0°, as shown in Figure 4.4(b). Furthermore, when the rotation time decreased from 1 s to 0.4 s, the dose penalty due to overscanning increased and peak CTDI\textsubscript{100, 12:00} values, relative to the mean CTDI\textsubscript{100, 12:00} for a start angle of 270°, of 4.1% and 10.5% were observed in the 16-cm and 32-cm phantom, respectively.
FIGURE 4.4. CTDI$_{100, 12:00}$ versus the x-ray tube start acquisition angle in a 10-cm, 16-cm, and 32-cm CTDI phantom. The phantoms were scanned using the techniques listed in Table 4.1 for both (a) 1-s and (b) 0.4-s rotation times; however, the 10-cm phantom was scanned using 1-s rotations only.
4.3.2. CTDI Phantoms Off-center

The CTDIs calculated from exposures detected at the 12:00 and 9:00 peripheral chamber positions in a 32-cm CTDI phantom are shown in Figure 4.5 for various start acquisition angles and amounts of vertical and emulated horizontal off-centering. Overscanning had a greater effect on dose as the phantom moved "left," "right," and above the position where its central chamber hole was aligned at isocenter; the opposite was observed when lowering the phantom. As Figure 4.5 shows, the peak CTDI$_{100, 12:00}$ values, relative to the mean CTDI$_{100, 12:00}$ values for 270° start angles, ranged from 6.0% to 26% when the phantom was moved from 6 cm below isocenter to 12 cm above isocenter, respectively. When using 1-s rotations, the peak CTDI$_{100, 12:00}$ was 11% when the phantom was 12 cm above isocenter (graph not shown). In the emulated horizontal off-centering cases, the peak CTDI$_{100, 9:00}$ values, relative to the mean CTDI$_{100, 9:00}$ values for 90° start angles, ranged from 11% when the phantom was 3 cm "left" or "right" of isocenter to 16% when the phantom was 12 cm "right" of isocenter. For the same degree of emulated left and right off-centering, the CTDI$_{100, 9:00}$ versus start acquisition angle curves were roughly symmetric; asymmetries that was observed may be attributable to the location of the patient table relative to the dosimeter and may not have existed if dose had been measured at 12:00 while physically moving the phantom left and right.
FIGURE 4.5. (a) CTDI\textsubscript{100, 12:00} and (b) CTDI\textsubscript{100, 9:00} versus the x-ray tube start acquisition angle in a 32-cm CTDI phantom scanned using the technique listed in Table 4.1 (for 0.4-s rotations) with various amounts of vertical off-centering (ranging from 6 cm below isocenter to 12 cm above isocenter) and emulated horizontal off-centering (from 6 cm “left” of isocenter to 12 cm “right” of isocenter), respectively.
In the 16-cm CTDI phantom, the peak CTDI\textsubscript{100, 12:00} when the phantom was 3 cm above isocenter was observed when using a start angle of 6° and was 6.0% than the mean CTDI\textsubscript{100, 12:00} for a 270° start angle. For emulated horizontal off-centering in the 16-cm phantom when moved 3 cm to the "right" of isocenter, the peak CTDI\textsubscript{100, 9:00} was observed for a start angle of 295° and was 5.6% higher compared to the CTDI\textsubscript{100, 9:00} when using a 90° start angle.

In addition to changes in dose due to overscanning, the CTDIs were also affected by off-centering. We compared our results to those reported by Li et al.\textsuperscript{155} to characterize the dose effect of off-centering outside the scope of overscanning. In the Li et al. study, a 32-cm CTDI phantom was moved below isocenter and dose was measured at several peripheral chamber locations, including the top, side, and bottom of the phantom (i.e., 12:00, 9:00, and 6:00, respectively). While we did not measure dose at the bottom of the phantom (6:00), we did measure dose for a geometrically-equivalent scenario (besides the location of the patient table) by measuring dose at 12:00 with the phantom moved above isocenter. To compare our results to those reported by Li et al., we calculated the ratio of CTDIs with the phantom moved off-center to CTDIs with phantom at isocenter. Since the Li et al. study was not related to overscanning, we used the CTDI\textsubscript{100, 12:00} and CTDI\textsubscript{100, 9:00} values for 270° and 90° start acquisition angles, respectively, as these CTDIs were not affected by overscanning. Table 4.2 shows that our results were consistent with those observed by Li et al., confirming that overall dose and not just the dose penalty from overscanning is affected by off-centering.
TABLE 4.2. The ratio of CTDIs in a 32-cm CTDI phantom when the phantom was positioned at isocenter to when the phantom was moved below isocenter (for a given peripheral dose measurement location). To provide a source of comparison for our results, the results of a study by Li et al.\textsuperscript{155} are also shown.

<table>
<thead>
<tr>
<th>Distance below isocenter (mm)</th>
<th>Ratio to isocenter dose*</th>
<th>Li et al. study\textsuperscript{155}</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>This study</td>
<td>&quot;Bottom&quot;\textsuperscript{†}</td>
</tr>
<tr>
<td>0</td>
<td>1.00</td>
<td>1.00</td>
</tr>
<tr>
<td>30</td>
<td>1.19</td>
<td>0.97</td>
</tr>
<tr>
<td>60</td>
<td>1.41</td>
<td>0.91</td>
</tr>
</tbody>
</table>

\textsuperscript{*}CTDI\textsubscript{100, 12:00} values, which were calculated from the mean of eight exposures measured using 270° start acquisition angles, were used to determine the ratios for the top and "bottom" positions; for the side position, CTDI\textsubscript{100, 9:00} values, which were calculated from the mean of eight exposures measured using 90° start acquisition angles, were used to determine the ratios.

\textsuperscript{†}For the current study, the phantom was moved above isocenter and measurements were collected at 12:00, which is geometrically equivalent to the phantom being moved below isocenter with the chamber placed at 6:00; the only difference is the location of the patient table.

### 4.3.3. COMPUTER SIMULATION

Figure 4.6 shows the results of the computer simulations for the dose penalties versus start acquisition angles. The corresponding measured data is also plotted in Figure 4.6 in terms of the percent differences in the CTDI\textsubscript{100, 12:00}
values across start acquisition angles, compared the CTDI_{100,1200} values using a 270° start acquisition angle. Figures 4.6(a) and 4.6(b) show the dose penalties in the 10-cm, 16-cm, and 32-cm CTDI phantoms when placed at isocenter and scanned using 1-s rotations; Figures 4.6(c) and 4.6(d) show the dose penalties in the 32-cm CTDI phantom when moved vertically off-center for 0.4-s rotation times. The data shown in Figure 4.4(a) and Figure 4.5(a) was used to construct Figures 4.6(a) and 4.6(c), respectively, while the computer model generated the data shown in Figures 4.6(b) and 4.6(d). Despite a gross simplification of the measurement conditions, such as not accounting for scatter, not modeling the shape of the bowtie filter, and not modeling the patient table, the computer model was able to closely predict the dose penalties due to overscanning. This indicates that the observed dose penalties were consistent with what would be expected based primarily on the geometry of the experimental setup used (i.e., phantom diameter, gantry diameter, scan field of view, etc.).
Figure 4.6. Results of the computer model simulation alongside the corresponding measured data. Graphs (a) and (b) show dose penalties in the 10-cm, 16-cm, and 32-cm CTDI phantoms when placed at isocenter and scanned using 1-s rotations; (c) and (d) show dose penalties in the 32-cm CTDI phantom when moved vertically off-center and scanned using 0.4-s rotations. The measured data shown in Figure 4.4(a) and Figure 4.5(a) was used to construct plots (a) and (c), respectively, while the computer model produced plots (b) and (d).
4.4. DISCUSSION

In this study, we characterized the dose effect of overscanning, in the form of a 10-ms rise and stabilization time which occurs prior to data acquisition on GE scanners. Dose penalties of up to 26% were observed on a GE LightSpeed VCT scanner due to overscanning. Furthermore, overscanning was shown to have a greater impact on dose for larger phantoms, shorter rotation times, and for regions in closer proximity to the tube start angle. To the best of the authors' knowledge, no prior study has quantified the dose penalty due to overscanning.

Based on the results of the off-centering portion of this study, moving patients above isocenter could be considered to be a breast dose-reduction strategy. However, moving a phantom above isocenter can affect TCM by magnifying the patient's size in an anteroposterior localizer, thus resulting in higher tube currents than necessary being used. Additionally, off-centering affects image quality because the bowtie filter is designed to match a patient's cylindrical shape, assuming the patient is properly centered within the gantry. For these reasons, patient off-centering is not recommended for dose reduction.

One of the objectives of this study was to pinpoint the 13% average decrease in point doses observed between axial and helical scan modes in the prior study by McDermott et al. The McDermott et al. study featured a small phantom, 1-s rotations, and appropriate centering; thus, based on the results of this study, the dose penalty due to overscanning should have been less than 2% and does not explain the results of the McDermott et al. study. Additionally,
overranging was not considered to be a factor since the scan setup was such that any overranging would have occurred well beyond the location of the TLD point dosimeters. Instead, it is likely that the results observed in the McDermott et al. study can be explained by the use of TLDs to estimate point doses; issues with comparing point dose estimates obtained across scan modes due to differences in alignment of the z-axis radiation dose profiles in relation to the TLDs have been reported.\textsuperscript{158} On the other hand, overscanning does explain reported inconsistencies in CTDI around the periphery of a CTDI phantom. For a single axial rotation, dose measured by a 10-cm pencil ionization chamber at the 12:00 chamber position in a 32-cm CTDI phantom was reported to be 5% larger than dose at the 3:00 and 9:00 positions, which is consistent with the results of this study.\textsuperscript{159}

Helical dose estimates are commonly calculated from data collected in axial mode by scaling according to pitch. However, this conversion does not take into account the increased frequency of overscanning events that occurs in axial compared to helical scan mode. Although contiguous axial and helical (pitch = 1) scan protocols with the same scan parameters have traditionally been considered to deliver equivalent doses,\textsuperscript{150} the results of this study indicate that helical dose could be overestimated by as much as 26%. Because the dose penalty due to overscanning is not uniform within the scan plane and is largest near the tube start acquisition angle, it would not be easy to incorporate a correction for overscanning into standard dose estimates.
Because overscanning has been given little attention, to the best of the author's knowledge, efforts have not been made to reduce this penalty. We suggest that both axial and helical acquisitions feature a default start acquisition angle of 180° (i.e., bottom of the gantry) because the dose penalty due to overscanning is largest near the x-ray tube start acquisition angle and superficial radiosensitive organs (e.g., breast, testes, and thyroid) are more prevalent on the anterior side of patients. Furthermore, the patient table would attenuate some of dose penalty due to overscanning if a 180° start acquisition angle was used.

While these results indicate that contiguous helical scanning (i.e., pitch = 1) may provide a dose advantage over axial scanning (all other technique factors being equal) within the scan extent (not including helical overranging), helical scanning is also associated with a dose penalty. In some cases, helical scanning is expected to have increased dose compared to axial scanning because of overranging. Organs that are outside the scan extent and would not have been directly exposed in axial scan mode could potentially be exposed due to overranging in helical scan mode. Although the dose penalties reported from overscanning are generally less than from overranging, one scan mode does not display a clear advantage over the other. For instance, if the organs beyond the acquisition length are not sensitive, while organs within the scan length are, then helical scanning could provide an advantage over axial scanning. On the other hand, in cases where single images are acquired (e.g., delayed kidney and/or bladder in pediatric CT exams), axial mode would be preferred to avoid overranging. Thus, a general recommendation on which mode is preferable for
dose savings cannot be made; instead, the specific nature of the examination must be considered to determine which mode provides a dose advantage.

4.5. CONCLUSION

In light of the dose penalty observed in this study, all acquisitions (especially axial acquisitions) should feature a default start angle of 180° to avoid imparting the dose penalty from overscanning on superficial radiosensitive organs (e.g., the breast, testes, and thyroid), which are more prevalent on the anterior side of patients. Furthermore, factors like patient centering are important across both acquisition modes.
CHAPTER 5

VARYING KVP AS A MEANS OF REDUCING CT BREAST DOSE TO PEDIATRIC PATIENTS

PURPOSE: The radiosensitivity of breast tissue is relatively high (compared to other tissues) and inversely related to age at exposure. Therefore, young girls represent a high-risk patient population from radiation exposure during CT scanning. We investigated the possibility of reducing radiation dose to the breast tissue of pediatric female patients by using multiple tube voltages during a single CT examination.

METHODS: The kVp was raised or lowered at angular positions (during the x-ray tube’s 360° rotation) that corresponded to direct exposure of breast tissue; this strategy was called kVp splitting. kVp splitting is a novel strategy and thus has not been implemented in CT scanners. To evaluate the potential of kVp splitting, it was emulated when scanning anthropomorphic phantoms representative of a 5-year-old, 10-year-old, and an adult female by using a different kVp over the anterior and posterior tube angles (for every pair of the available kVp options). Dose savings from kVp splitting were calculated relative to using a static kVp over all tube angles. MC simulations with and without kVp splitting (using the optimal anterior and posterior kVps determined in the anthropomorphic phantoms) were performed to estimate breast dose-savings in voxelized patient models constructed from the CT images of a small, medium, and large pediatric female.
RESULTS: Scans emulating kVp splitting in anthropomorphic phantoms revealed dose savings, relative to static kVp scans, when using 80 kVp over the posterior tube angles in all three phantoms, regardless of the anterior kVp. MC simulations of kVp splitting, which used 80 kVp over the posterior tube angles and 100 kVp (for the small and medium voxelized models) or 120 kVp (for the large voxelized model) over the anterior angles, revealed breast dose-savings of between 9.8% and 34%, compared to scans using the anterior kVp for the entire scan.

CONCLUSION: kVp splitting was estimated to reduce absorbed dose to the breast tissue by between 9.8% and 34%. However, image reconstruction algorithms and the image quality of scans with kVp splitting must be further investigated before this strategy can be implemented clinically.

5.1. INTRODUCTION

An inverse relationship between a patient’s age at the time of exposure to ionizing radiation and the patient’s relative radiation sensitivity has been reported. Therefore, pediatric patients undergoing CT scanning are at higher risk for adverse health effects (e.g., cancer) than adult patients. Furthermore, according to the ICRP’s most recent tissue weighting factors, female breast tissue is among the most radiosensitive tissues. For these reasons, the risk of breast cancer is considered to be especially high for pediatric female patients undergoing CT procedures.
Many user-input scan parameters, including the kVp, tube current, and exposure time, influence the amount of radiation patients are exposed to during CT scanning.\textsuperscript{146} The x-ray tube current and exposure time affect the number (quantity) of photons generated by the x-ray tube and are linearly proportional to radiation dose.\textsuperscript{6} The kVp affects both the quantity and energy (quality) of photons in the beam and dose is directly proportional to the 2.5 or 2.8 power of kVp.\textsuperscript{160} Although both the tube current and kVp can be adjusted to reduce radiation dose, reducing the quantity of x-rays affects both the spatial resolution and image noise.\textsuperscript{146, 161}

Adjusting the x-ray tube current is the simplest and most predictable means of reducing dose.\textsuperscript{95} Thus, TCM is often employed, which enables CT scanners to vary the tube current throughout scanning according to a patient’s anatomy.\textsuperscript{99} TCM allows for dose reduction because only the photons needed to maintain a desired level of image quality are delivered to each anatomical region; TCM also makes it possible to maintain consistent image quality throughout the entire examination. In MC simulations of TCM on 30 voxelized female patient models, glandular breast dose was reduced by an average of 17\% compared to scans using a static mA; however, TCM was also shown to increase breast dose in some larger patients.\textsuperscript{162}

A form of TCM specifically designed to reduce breast dose has recently been developed (X-Care; Siemens Healthcare). X-Care reduces direct exposure of the breast tissue by either greatly reducing or completely turning off the x-ray
tube current (i.e., 0 mA) over the breasts. A similar technology (HandCare; Siemens Healthcare) was tested in CT fluoroscopy and was shown to reduce breast dose by 47%, as measured using TLDs in an adult female anthropomorphic phantom, when the x-ray beam was turned off over the top 120° of the rotation. Computer simulations have estimated breast dose reductions of nearly 50% using X-Care. However, because posterior tube current was increased to maintain image quality (in the absence of anterior projection data), higher doses to the spine and bone marrow were also estimated in the simulations.

Dose is roughly quadratically proportional to kVp, but only linearly proportional to tube current; therefore, varying the kVp during scanning may result in greater dose-savings than from either TCM or X-Care. However, the kVp affects both beam quality and quantity, and so the relationships between kVp, dose, and image quality are much complex than with tube current. kVp affects the image contrast and in order to achieve acceptable image quality, kVp reduction may require a compensatory increase in tube current. In pediatric patients, the compensatory tube current increase may be less important because reducing the kVp (for a fixed tube current) has been reported to have a lesser impact on image noise in smaller versus larger phantoms. Likewise, in a study by Kim and Newman, diagnostic image quality was maintained when simultaneously reducing both the kVp and tube current across pediatric protocols. Because increasing the kVp increases the beam’s penetrability, which means that photons are less likely to be attenuated at the patient’s surface.
(breast tissue is a relatively superficial organ), breast dose reduction may be possible even with a compensatory increase in the tube current.

We evaluated the possibility of reducing CT breast dose by changing the kVp at angular positions of the x-ray tube that would correspond to direct exposure of patients' breast tissue. Therefore, the kVp was either raised or lowered when the x-ray tube was over the breast; both options were tested in light of the complex relationship between surface dose and kVp. Because the kVp effectively “splits” into two values during rotations over the breasts (illustrated in Fig. 5.1) and then converges back to a single kVp once past the breasts, we will refer to this technique as “kVp splitting.” Although kVp splitting involves the use two kVps within a single scan, it is not to be confused with dual-energy CT, which acquires data for a full rotation at each kVp.

**Figure 5.1.** kVp splitting is a novel strategy for reducing CT breast dose by varying the kVp over the breast tissue.
The purpose of this study was to quantify the potential of kVp splitting for reducing CT breast dose. Because CT scanners currently do not allow for kVp splitting, the physical data that could be collected to assess this technique’s effectiveness was limited. Therefore, MC computer modeling was employed to simulate kVp splitting on voxelized models generated from the CT images of female pediatric patients of different size and age.

5.2. MATERIALS AND METHODS

5.2.1. MEASURED DATA

Before MC modeling was used, physical data was collected to guide the MC simulations using CTDI and anthropomorphic phantoms. CTDI_{100, air} and CTDI_{100, 12:00} (for a 16-cm and 32-cm CTDI phantom) were measured at all of the four available kVp options to determine how dose varies with kVp and thus how the tube current should be adjusted in scans with kVp splitting. The anterior and posterior kVps used in the MC simulations of kVp splitting were selected based on measured data collected in anthropomorphic phantoms.

5.2.1.1. CTDI

Adjustment of the tube current is typically needed to maintain constant CT image quality when varying the kVp. However, because CT images could not be acquired with kVp splitting, image quality could not be directly evaluated. Therefore, we instead determined the tube currents that would keep CTDI
constant across kVps. Because of the reported inconsistencies in the powers of kVp which CTDI is proportional to,\(^6\),\(^{160}\),\(^{166}\) these powers were determined within the context of this study.

To determine the powers of kVp that dose varies as, exposures were detected at 80 kVp, 100 kVp, 120 kVp, and 140 kVp using a 10-cm pencil ionization chamber (RadCal Corporation) connected to an electrometer. The ionization chamber was initially suspended free-in-air at the gantry isocenter, which allowed us to measure exposure without the effect of scatter from a phantom or the patient table. Next, the ionization chamber was placed at the 12:00 peripheral chamber position inside a 16-cm and 32-cm CTDI phantom placed on the patient tabletop; the chamber was positioned at 12:00 because this position most closely represents the location of breast tissue in a patient. For both setups, data was collected on a LightSpeed VCT scanner (GE Healthcare) in axial scan mode using single 0.5-s rotations. Additionally, the small focal spot size, which was controlled by the tube current, and the “large body” SFOV, which employed the large bowtie filter option, were used to match the protocols for scanning pediatric patients at our facility. A LightSpeed VCT scanner was used for data collection because it is a state-of-the art CT scanner and thus was considered to be a good representative of a scanner that could incorporate kVp splitting in the future.
Because kVp splitting is a novel technique and thus has not been implemented in CT scanners, it was emulated by performing a series of stationary scans over the equivalent of a full rotation and changing the kVp at angular positions of the x-ray tube that would correspond to direct exposure of patients' breast tissue (shown in Figure 5.4). This approach was validated by first performing 0.5-s axial scans in service mode on a GE LightSpeed VCT scanner. Subsequently, a series of 0.5-s exposures were delivered with the x-ray tube held stationary at evenly-spaced angular positions over the equivalent of a 360° rotation. Each series consisted of 18 stationary scans performed at tube angles from 0 – 340° (in 20° angular increments). The same scan techniques used to determine CTDI (i.e., 80, 100, 120, and 140 kVp, a small focal spot size, a 40-mm nominal beam width, and a large bowtie filter) were also used in this portion of the study. Entrance exposures were measured using a Farmer ionization chamber (in conjunction with an electrometer; RadCal Corporation) placed on the surfaces of the anthropomorphic phantoms at the location representing the sternum. Exposures delivered by the stationary scans, which were calculated as the sum of the 18 individual exposure measurements in each series, and the rotating scans were normalized by exposure time and then compared. Across all three phantoms and all four kVps, the difference in the exposures delivered by the rotating scans and the series of stationary scans ranged from 0.07% to 1.44%.

Exposures delivered by both the stationary scans, which were calculated as the sum of the 18 individual exposure measurements in each series, and the
anterior ("kVp1" in Figure 5.1) and posterior ("kVp2" in Figure 5.1) tube angles (e.g., 80 kVp for the anterior angles and 100 kVp for the posterior angles). A single tube current was used for each phantom across all scans (regardless of kVp), and later the detected exposures were scaled according to the exponents of kVp determined free-in-air and in the CTDI phantoms. For example, if the exponent of kVp was 2.5 (for 80 kVp relative to 100 kVp), the current for the 80-kVp exposures would be scaled so that it was 1.75 times the current used for the 100-kVp exposures. When estimating dose delivered by kVp splitting, the stationary exposures measured for the posterior tube angles (angles between 80° and 280° [inclusive], where 0° is at the top of the gantry) and anterior tube angles (300°, 340°, 0°, 40°, 60°) were summed separately. To determine cumulative exposure from the equivalent of an entire rotation, we combined exposures measured for the posterior and anterior regions. To combine data collected at different kVps and maintain image quality comparable to that of a static kVp scan, we scaled the cumulative exposure for each region using the exponents of kVp; however, image quality could not be directly evaluated because images cannot be reconstructed from stationary scans. The cumulative posterior and anterior exposures were summed and compared to the cumulative exposure for a scan that used a static kVp across all 18 stationary scans. The static kVp scan, which the kVp splitting scans were compared to, was selected for each phantom based on the kVp used to scan the corresponding patient (i.e., the examinations for both the small and medium patients were performed at 100 kVp, so 100 kVp was used in the comparison scan). Dose savings were
determined by calculating the percent difference in exposure between each kVp splitting scan (for all combinations of anterior and posterior kVps) and the relevant fixed kVp scan.

5.2.2. **MONTE CARLO DOSE ESTIMATES**

Three pediatric female patients, who were considered to be representative of small, medium, and large pediatric patients, were identified and used to construct voxelized models for MC modeling. MC modeling was employed because it provided a more accurate means of estimating breast dose to individual patients and also because it allowed for scans to be simulated that were unable to be physically performed because kVp splitting is not available on CT scanners (e.g., helical scanning with kVp splitting). MC simulations with and without kVp splitting were performed to estimate the amount of radiation dose deposited to the glandular breast tissue of three pediatric female voxelized models so that the breast dose-savings potential of kVp splitting could be evaluated. The anthropomorphic phantom results were used to determine the anterior and posterior kVps used in the simulations because acquiring physical measurements for all kVp combinations was much more time efficient than performing an equivalent number of MC simulations.
5.2.2.1. **Voxelized Models**

After receiving institution review board approval, patient data used to construct the voxelized models was obtained from three pediatric female patients who had undergone positron emission tomography with CT (PET/CT) using a Discovery STE scanner (GE Healthcare) at our institution. PET/CT rather than diagnostic CT was used because it acquires a single, continuous set of images without intravenous or oral contrast, which could interfere with tissue identification and characterization. A potential disadvantage of using PET/CT images to construct the voxelized models is that they have a larger voxel size than diagnostic CT images; however, because the models generated from the images are typically somewhat coarse (even when detailed CT image data is available), the large voxel size was not considered to be an issue.

Voxelized models of the three patients were developed by identifying those voxels in the patients’ CT images that contained glandular breast or lung tissue; a pediatric radiologist was consulted to ensure proper identification of the patients' breast tissue. Lung tissue was also contoured because it is among the most radiosensitive organs (according to the ICRP tissue weighting factors) and is in the same scan plane as breast tissue. Figure 5.2 shows contouring of both the lung and breast tissue in one CT image for each of the three patients; all images containing breast tissue were contoured individually.
<table>
<thead>
<tr>
<th>Patient model</th>
<th>kVp</th>
<th>Current (mA) *</th>
<th>Rotation time (s)</th>
<th>Pitch</th>
<th>Effective mA·s †</th>
<th>Beam width (mm)</th>
<th>SFOV (^{\dagger} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Small (3-year-old)</td>
<td>100</td>
<td>10 – 70</td>
<td>0.5</td>
<td>1.375</td>
<td>15</td>
<td>20</td>
<td>Large</td>
</tr>
<tr>
<td>Medium (10-year-old)</td>
<td>100</td>
<td>10 – 120</td>
<td>0.5</td>
<td>1.375</td>
<td>24</td>
<td>20</td>
<td>Large</td>
</tr>
<tr>
<td>Large (17-year-old)</td>
<td>120</td>
<td>10 – 110</td>
<td>0.5</td>
<td>1.375</td>
<td>33</td>
<td>20</td>
<td>Large</td>
</tr>
</tbody>
</table>

* The tube currents listed refer to the minimum and maximum values used in the tube current modulation scheme; for all three patients, a noise index of 20 was used.

† The effective mA·s was calculated from pitch, rotation time, and the average tube current over the entire scan length (as determined from each patient's TCM scheme)

\(^{\dagger} \) The large SFOV corresponded to a 50-cm DFOV; this SFOV used the large bowtie filter.
5.2.2.2. SIMULATIONS

Anterior and posterior kVps determined to reduce dose to the anthropomorphic phantoms were used in MC simulations of kVp splitting to estimate the amount of radiation dose deposited to the breast tissue of the three voxelized patient models. A Monte Carlo computer model that uses the MC N-Particle code eXtended version 2.6d (MCNPX) was employed to estimate the amount of radiation dose deposited to the breast tissue for a given voxelized patient model (generated from the CT data of the small, medium, or large pediatric patient).\textsuperscript{116, 167} Breast dose has been previously quantified in patient-based models (GSF; Neuherberg, Germany) using the MCNPX model.\textsuperscript{168} To simulate kVp splitting, we performed two separate MC simulations and the results were summed in a manner similar to that used to sum the posterior and anterior exposures in the anthropomorphic phantom measurements. Because the normalization factors used to convert output of the MC model to dose are kVp-dependent, output of a single simulation that has varying kVps cannot be converted to absolute dose. Therefore, angular regions of the scan performed at each kVp were simulated separately. The MCNPX model was validated for modeling kVp splitting within 6.1\% of the physical data collected on a GE LightSpeed VCT scanner in a 32-cm CTDI phantom using 120 kVp for posterior angles and 80 kVp for anterior angles (data not shown). Validation was done in a CTDI phantom rather than an anthropomorphic phantom based on prior validation.
To assess the effectiveness of kVp splitting in a manner that was clinically representative of how it would be employed, we followed the protocols used when scanning the patients as closely as possible when performing the simulations. However, several factors were changed between the patient scans and the simulations including the scanner type and model, the scan extent, the x-ray tube start angle, and the use of TCM. Although a Discovery STE PET/CT scanner, which contains a 16-slice CT scanner, was used to collect the patient data, a GE LightSpeed VCT was used in the simulations. This change was made in part because an equivalent-source MC model has not yet been built for the GE 16-slice scanner used in the Discovery STE PET/CT scanner. To define where kVp splitting would be applied, we determined the anatomical location of breast tissue in the angular and z-directions from the CT images. The posterior and anterior regions were defined according to which x-ray tube angles directly exposed the breast tissue with the patient positioned “as is” within the 50-cm DFOV. Although the angular coverage for off-center patients was different than if the patient had been centered (as can be seen in Figure 5.5), the simulations were performed without adjusting the patients’ position within the field of view. The scan extent used in the simulations was based on the z-axis locations where breast tissue could be visualized in the CT images plus one beam width (i.e., 40 mm) before and one beam width after the region containing breast tissue; the additional beam widths were included to ensure complete coverage of the breast tissue. Figure 5.6 shows the z-axis coverage of the breast tissue as well as the one-beam-width border (40 mm), which make up the scan extent simulated for
the small and large patients. While performing the simulation over the breasts only is not clinically realistic in terms of the scan extent, it can be assumed that dose to anatomy beyond the breast region would be unaffected by kVp splitting since this technique would only be used over the breasts and thus dose elsewhere should be relatively constant. This assumption allowed for simulation times to be reduced by only simulating kVp splitting over the region encompassing the breast tissue. The angular position of the x-ray tube was set to equal 0° (i.e., the top of the gantry) at the z-axis location which corresponded to the center of the simulated scan extent; the start angle was then calculated based on the beam width, pitch, and the distance (along the z-axis) between the start and center of the simulated scan length. Because simulating TCM would add another layer of difficulty and was unnecessary for demonstrating the potential of kVp splitting, a single mA was used for each portion of the scan. The actual patient TCM schemes (taken from the raw data) were used to calculate the average current values over the simulated scan extent (as seen in Figure 5.7) and this value was used to scale the output of the MC simulations (the MC simulations did not directly account for current). For the small, medium, and large patients, the average currents over the scan extent simulated (i.e., images showing breast tissue with a border of one beam width) were calculated to be 65 mA, 118 mA, and 108 mA, respectively. Table 5.2 lists the technical factors and other information relevant to the simulations.
TABLE 5.2. Techniques used in the MC simulations to assess the effectiveness of kVp splitting for the small, medium, and large patients; all simulations were performed using a GE LightSpeed VCT scanner.

<table>
<thead>
<tr>
<th>Patient model</th>
<th>kVp</th>
<th>mA</th>
<th>Helical pitch</th>
<th>Rotation time (s)</th>
<th>Beam width (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Small†</td>
<td>80 &amp; 100</td>
<td>65 &amp; 116 or 114</td>
<td>1.375</td>
<td>0.5</td>
<td>40</td>
</tr>
<tr>
<td>Medium†</td>
<td>80 &amp; 100</td>
<td>118 &amp; 209 or 206</td>
<td>1.375</td>
<td>0.5</td>
<td>40</td>
</tr>
<tr>
<td>Large‡</td>
<td>80 &amp; 120</td>
<td>108 &amp; 318 or 286</td>
<td>1.375</td>
<td>0.5</td>
<td>40</td>
</tr>
</tbody>
</table>

* For the small patient, 100 kVp and 65 mA were used for the static kVp simulation. For the kVp splitting simulation, 80 kVp and 116 mA or 114 mA (depending on whether mA was scaled by CTDI$_{100, 12:00}$ or CTDI$_{100, \text{air}}$, respectively) was used over the posterior tube angle (from 115° to 245°, clockwise); 100 kVp and 65 mA were used over the anterior tube angles (from 245° to 115°, clockwise).

† For the medium patient, 100 kVp and 118 mA were used for the static kVp simulation. For the kVp splitting simulation, 80 kVp and 209 mA or 206 mA (depending on whether mA was scaled by CTDI$_{100, 12:00}$ or CTDI$_{100, \text{air}}$, respectively) was used over the posterior tube angle (from 90° to 270°, clockwise); 100 kVp and 118 mA were used over the anterior tube angles (from 270° to 90°, clockwise).

‡ For the large patient, 120 kVp and 108 mA were used for the static kVp simulation. For the kVp splitting simulation, 80 kVp and 318 mA or 286 mA (depending on whether mA was scaled by CTDI$_{100, 12:00}$ or CTDI$_{100, \text{air}}$, respectively) was used over the posterior tube angle (from 105° to 255°, clockwise); 120 kVp and 108 mA were used over the anterior tube angles (from 255° to 105°, clockwise).
Outputs from each of the two simulated scans (i.e., the scans over the anterior and posterior regions) were combined into a single dose by summing the dose obtained over each region (in mGy). Before the doses could be added, they were multiplied by the appropriate kVp normalization factors, weighted by their fraction of a 360° rotation (e.g., the output would be multiplied by 0.5 if the region covered 180°), and then multiplied by the current-exposure time product; the in-air exposure data were used to determine the normalization factors for each kVp. After determining dose for the kVp splitting scans, breast tissue dose estimates were also obtained for a static kVp protocol. To quantify the breast dose-savings potential of kVp splitting, we compared dose (through percent difference) for a full scan at a fixed kVp and tube current to the dose for scans with kVp splitting.

5.3. RESULTS
5.3.1. MEASURED DATA
5.3.1.1. CTDI

Because both the small and medium patients were scanned at 100 kVp and thus 100 kVp was used in the static kVp simulations, the exponents of kVp for the 16-cm CTDI phantom were determined relative to 100 kVp only. Similarly, the kVp exponents for the 32-cm phantom were determined relative to 120 kVp only. Table 5.3 lists the kVp exponents determined at the 12:00 peripheral chamber position in CTDI phantom and free-in-air.
Table 5.3. Exponents of kVp calculated using Eq. (5.1) and the measured exposures collected free-in-air or at the 12:00 chamber position in the CTDI phantoms; exponents were only determined relative to 100 kVp for the 16-cm CTDI phantom or 120 kVp for the 32-cm phantom. The ratio of kVps can be raised to these powers to determine how to scale the tube current across different kVps for scans with kVp splitting.

<table>
<thead>
<tr>
<th>kVp</th>
<th>CTDI_{100, air}</th>
<th>CTDI_{100, 12:00} (16-cm phantom)</th>
<th>CTDI_{100, 12:00} (32-cm phantom)</th>
</tr>
</thead>
<tbody>
<tr>
<td>80 kVp (relative to 100 kVp)</td>
<td>2.50</td>
<td>2.57</td>
<td>-</td>
</tr>
<tr>
<td>80 kVp (relative to 120 kVp)</td>
<td>2.40</td>
<td>-</td>
<td>2.66</td>
</tr>
<tr>
<td>100 kVp (relative to 120 kVp)</td>
<td>2.27</td>
<td>-</td>
<td>2.46</td>
</tr>
<tr>
<td>120 kVp (relative to 100 kVp)</td>
<td>2.27</td>
<td>2.28</td>
<td>-</td>
</tr>
<tr>
<td>140 kVp (relative to 100 kVp)</td>
<td>2.21</td>
<td>2.25</td>
<td>-</td>
</tr>
<tr>
<td>140 kVp (relative to 120 kVp)</td>
<td>2.13</td>
<td>-</td>
<td>2.22</td>
</tr>
</tbody>
</table>
5.3.1.2. ANTHROPOMORPHIC PHANTOMS

The exponents of kVp determined free-in-air and in the CTDI phantoms were used to scale the current (by the ratio of the kVps raised to the appropriate power). Data collected in the 16-cm CTDI phantom were used to calculate the exponents for anthropomorphic phantoms representing a 5-year-old and a 10-year-old (because they were more similar in size to the 16-cm phantom than the 32-cm CTDI phantom); likewise, data collected in the 32-cm CTDI phantom were used to determine the scaling factors for the adult phantom. Table 5.4 lists the dose savings as a percentage for every kVp combination over the anterior and posterior tube angles using tube currents scaled either by CTDI100, 12:00 or CTDI100, air.
All exposure measurements were converted to CTDI using Eq. (1.4) and then the powers (exponents) of kVp that CTDI was proportional to were calculated according to the following equation:

$$\text{Exponent} = \log_{kVp_1/kVp_2}(\frac{CTDI_{kVp_1}}{CTDI_{kVp_2}}),$$  \hspace{1cm} (5.1)

where $kVp_1/kVp_2$ is the base of the logarithm and $CTDI_{kVp_1}/CTDI_{kVp_2}$ is the ratio of CTDIs calculated from the exposures measured at $kVp_1$ and $kVp_2$.

5.2.1.2. ANTHROPOMORPHIC PHANTOMS

Once the relationships for describing how dose varies with kVp were determined, kVp splitting was evaluated using anthropomorphic ATOM family dosimetry phantoms representative of a 5-year-old, 10-year-old, and an adult upper torso (with breast attachments) (CIRS, Norfolk, VA), which are shown in Figure 5.3.

**Figure 5.3.** Anthropomorphic phantoms representing a 5-year-old, 10-year-old, and a (female) adult were used in this study.
rotating scans, were normalized by exposure time and then compared. Across all three phantoms and all four kVps, the difference in the exposures delivered by the rotating scans and the series of stationary scans ranged from 0.07% to 1.44%.

**Figure 5.4.** A series of 18 stationary scans were performed in 20° angular increments from 0° to 340° to emulate a 360° rotation with kVp splitting. The central ray of the x-ray fan beam at each angular increment is shown as red or blue arrows; the blue arrows represent scans at the anterior kVp (kVp₁) while the red arrows represent scans at the posterior kVp (kVp₂).

Because exposures measured using the stationary scan method were consistent with those measured for the rotating scans, the stationary scan method was considered valid for emulating kVp splitting. Therefore, every combination of the four available kVps (i.e., 12 pairs) was tested over both the
FIGURE 5.2. Lung (teal) and glandular breast tissue (purple) were contoured for the (a) small, (b) medium, and (c) large pediatric patients. All of the images are shown with a 50-cm display FOV (DFOV), which allows for direct comparison and visualization of the difference in the patient sizes.

The patients chosen to represent the small, medium, and large pediatric patients were 3 years old (height: 102.3 cm, weight: 14.8 kg), 10 years old (height: 144.0 cm, weight: 36.7 kg), and 17 years old (height: 174.5 cm, weight: 72.3 kg), respectively, at the time of scanning. Patients with surgical defects (e.g., lung removal) or large tumors causing noticeable surface or anatomical deformity were excluded from the study. The 10-year-old patient had one leg amputated (at the thigh), but this did not affect the results of the MC simulations because breast dose was the focus of this investigation. The CT image data and raw data were collected for these three patients; from the raw CT data, the specific TCM schemes used on each patient were extracted. The techniques used to scan the patients are listed in Table 5.1.
FIGURE 5.5. The angular coverage of breast tissue for the small patient was 230° when centered “as is” within the 50-cm DFOV. However, if the patient had been properly centered, her breast tissue would have only covered a 130° angular region.
FIGURE 5.6. Highlighting shows the z-axis coverage of the breast tissue (blue highlighted region) and the scan extent used in the simulations (lighter highlighted region) for the (a) small and (b) large pediatric patient. The dashed line in the middle of the scan extent is the z-axis position at which the x-ray tube angle was set to 0°.
FIGURE 5.7. Plot showing the TCM scheme used to scan the medium-sized pediatric patient superimposed on the patient’s scout projection image (i.e., localizer image); the values used to construct this graph were obtained from the raw data. Because the MC simulation was performed over the breast region only, the average current over this region (enclosed between the dashed lines) was used in the simulation; for this patient, the average current was determined to be 118 mA.
TABLE 5.4. Percentage of dose savings from using kVp splitting in anthropomorphic phantoms representing a 5-year-old, 10-year-old, and adult relative to a full scan at a fixed 100 kVp (for the pediatric phantoms) or 120 kVp (for the adult phantom); negative values (bold-faced) indicate dose savings. The currents were scaled across kVps using the exponents listed in Table 5.3, which were calculated from CTDI$_{100, \text{air}}$ or CTDI at the 12:00 chamber position in a 16-cm CTDI phantom (for the pediatric phantoms) or 32-cm CTDI phantom (for the adult phantom).

<table>
<thead>
<tr>
<th>kVp</th>
<th>5-year-old</th>
<th>10-year-old</th>
<th>Adult</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>CTDI$_{100, \text{air}}$</td>
<td>CTDI$_{100, 12:00}$</td>
<td>CTDI$_{100, \text{air}}$</td>
</tr>
<tr>
<td>80 (anterior), 100 (posterior)</td>
<td>-1.3%</td>
<td>-0.2%</td>
<td>-1.3%</td>
</tr>
<tr>
<td>80 (anterior), 120 (posterior)</td>
<td>0.7%</td>
<td>1.7%</td>
<td>0.7%</td>
</tr>
<tr>
<td>80 (anterior), 140 (posterior)</td>
<td>2.1%</td>
<td>2.8%</td>
<td>2.0%</td>
</tr>
<tr>
<td>100 (anterior), 80 (posterior)</td>
<td>-4.2%</td>
<td>-5.3%</td>
<td>-3.5%</td>
</tr>
<tr>
<td>100 (anterior), 120 (posterior)</td>
<td>2.0%</td>
<td>1.9%</td>
<td>2.0%</td>
</tr>
<tr>
<td>100 (anterior), 140 (posterior)</td>
<td>3.4%</td>
<td>3.0%</td>
<td>3.3%</td>
</tr>
<tr>
<td>120 (anterior), 80 (posterior)</td>
<td>-3.9%</td>
<td>-3.8%</td>
<td>-3.2%</td>
</tr>
<tr>
<td>120 (anterior), 100 (posterior)</td>
<td>0.3%</td>
<td>0.3%</td>
<td>0.3%</td>
</tr>
<tr>
<td>120 (anterior), 140 (posterior)</td>
<td>3.7%</td>
<td>3.1%</td>
<td>3.6%</td>
</tr>
<tr>
<td>140 (anterior), 80 (posterior)</td>
<td>-3.8%</td>
<td>-4.4%</td>
<td>-3.1%</td>
</tr>
<tr>
<td>140 (anterior), 100 (posterior)</td>
<td>0.4%</td>
<td>-0.5%</td>
<td>0.4%</td>
</tr>
<tr>
<td>140 (anterior), 120 (posterior)</td>
<td>2.4%</td>
<td>1.4%</td>
<td>2.4%</td>
</tr>
</tbody>
</table>
5.3.2. Monte Carlo Dose Estimates

Results from the anthropomorphic phantom study were used to dictate the MC simulations that were performed. The 5-year-old, 10-year-old, and adult anthropomorphic phantoms were intended to be matched with the small, medium, and large pediatric patients, respectively, when determining which kVps would be used in the simulations for each patient; Figure 5.8 shows the CT images of the patients and the anthropomorphic phantoms used to represent them. Rather than choosing the kVps that gave the greatest dose savings for each individual phantom, instead we followed a general rule that the kVp used during the static kVp scan was used over the anterior tube angles of the kVp splitting simulated scan and the lowest kVp available (i.e., 80 kVp) was used over the posterior tube angles. Although using the combination of kVps specified by this rule definitively resulted in the greatest dose-savings in the pediatric phantoms, the results were less clear in the adult phantom and depended on whether current was scaled by the kVp exponents determined in the CTDI phantoms or free-in-air. However, to be consistent with the kVp splitting pattern used in the pediatric phantoms and because this pattern (i.e., 120 kVp over the anterior region and 80 kVp over the posterior region) did indicate a dose savings in the adult phantom when scaled either way, this general rule was followed across all simulations.
FIGURE 5.8. CT images of anthropomorphic phantoms representing a (a) 5-year-old, (b) 10-year-old, and (c) (female) adult alongside the CT images of the (d) small, (e) medium, and (f) large pediatric patients used to construct voxelized patient models. Note that the 10-year-old patient was noticeably bigger than the anthropomorphic phantom used to represent her.
Results of the MC simulations for all three patient models are shown in Table 5.5. Lung and breast dose are reported in the scans with and without kVp splitting along with the estimated percentage of dose savings from using kVp splitting. Scans simulated with kVp splitting had breast doses that were 9.8% to 34% (depending on how the current was scaled) lower than scans with a fixed kVp while generally not affecting the lung dose (within the scan extent simulated, which did not necessarily include all of the patients' lung tissue). Although we expected the absolute doses measured in the anthropomorphic phantoms to differ from those estimated through MC modeling, calculated absorbed dose to the phantoms was within 12% or less of the MC breast dose estimates for static kVp scans (data not shown); the phantom doses were normalized to match the mA·s and pitch used in the MC simulations. The difference in the dose estimates in phantoms and in the voxelized patient models increased to up to 39% in scans with kVp splitting.
TABLE 5.5. Breast and lung dose absorbed by three pediatric patients as estimated through MC computer modeling. For static kVp scans, a single tube current was used and thus a single dose is reported in both columns in the table. For scans with kVp splitting, different currents were used for the anterior and posterior tube angles (the posterior currents were calculated using the CTDI$_{100,\text{air}}$ or CTDI$_{100,12:00}$ scaling factors listed in Table 5.3 and thus two dose estimates were obtained; the CTDI$_{100,12:00}$ scaling factors were determined in a 16-cm CTDI phantom for the small and medium patients and in a 32-cm CTDI phantom for the large patient). In parentheses below the dose estimates, dose savings is given for the kVp splitting scans relative to a fixed kVp scan; negative values indicate dose savings.

<table>
<thead>
<tr>
<th>Patient model</th>
<th>kVp*</th>
<th>Lung dose (mGy)</th>
<th>Breast dose (mGy)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>CTDI$_{100,\text{air}}$</td>
<td>CTDI$_{100,12:00}$</td>
</tr>
<tr>
<td>Small</td>
<td>100  (full scan)</td>
<td>2.12</td>
<td>2.12</td>
</tr>
<tr>
<td></td>
<td>80 (posterior), 100 (anterior)</td>
<td>2.11, (-0.59%)</td>
<td>2.12, (-0.09%)</td>
</tr>
<tr>
<td>Medium</td>
<td>100  (full scan)</td>
<td>2.48</td>
<td>2.48</td>
</tr>
<tr>
<td></td>
<td>80 (posterior), 100 (anterior)</td>
<td>2.33, (-5.8%)</td>
<td>2.35, (-5.1%)</td>
</tr>
<tr>
<td>Large</td>
<td>120  (full scan)</td>
<td>4.40</td>
<td>4.40</td>
</tr>
<tr>
<td></td>
<td>80 (posterior), 120 (anterior)</td>
<td>4.48, (1.7%)</td>
<td>4.62, (4.9%)</td>
</tr>
</tbody>
</table>

* The tube currents used by each kVp are listed in Table 5.2.
5.4. DISCUSSION

kVp splitting was estimated in this study to lower CT breast dose in pediatric females by between 9.8% and 34% while presumably maintaining constant image noise. A trend in breast dose-savings versus patient size was not observed. Although this study related to pediatric patients, the large pediatric patient was adult-sized, so we expect that dose savings could also be achieved in adult patients. However, additional research is needed to assess this possibility and to assess dose savings across a greater sample of pediatric patients.

The discrepancy in the dose savings from kVp splitting between the phantom measurements and the simulated patient results (i.e., 4% dose savings in 5-year-old phantom versus 34% dose savings in the small patient model) is likely due to the fact that anthropomorphic phantoms do not fully represent human anatomy. However, it could also be related to differences between the scans and the simulations, including the angular position of the x-ray tube relative to the breast tissue, improper patient centering, or the added effect of helical scanning. Because the percentages of dose savings observed in the anthropomorphic phantoms were inconsistent with those observed in the MC, we cannot be sure that the optimal combinations of kVps were used in the simulations because these values were chosen based on the phantom results. Additional MC modeling could be performed to validate the anthropomorphic phantom study.
results; this would give us a higher degree of confidence in the validity of the MC
results.

In the age-based protocols that were initially used to scan the three prediatric
female patients, TCM allowed for specification according to overall patient
anatomy. The addition of kVp splitting would allow for further patient specification
by specifically accounting for differences in the amount and distribution of breast
tissue across pediatric patients (whose breast tissue may not protrude as in adult
patients and thus would be less likely to be detected by TCM). Organ-based
dose adjustment is important because cancer typically develops in a specific
organ. Although X-Care is geared specifically toward reduction of breast dose,
the x-ray beam can only be turned off for 180° minus the fan angle; on the other
hand, kVp splitting does not have such limitations and could be implemented
over any angular region according to specific anatomical location of each
patient’s breast tissue. Additionally, kVp splitting may produce better image
quality than X-Care because it does not involve turning the beam off, and so data
are sampled throughout the entire rotation. However, Vollmar and Kalender
reported that X-Care has little effect on image quality compared to a static mA
scan and because the specifics of kVp splitting’s effect on image quality are
unknown, this would need further evaluation prior to implementation.113 Dose
savings at the same level as X-Care (which has been reported to achieve 45 -
48% breast dose-savings) may be possible through kVp splitting. If the lower
tube current in the kVp splitting scheme (i.e., the current used over the anterior
region, with the higher kVp setting) was used for the entire scan (rather than
scaling current according to the powers of kVp), dose to the breast and lung in
the three voxelized models would be reduced by 42 – 60% and 15 - 26%,
respectively. Not increasing the current to compensate for reducing the kVp may
be reasonable for pediatric patients because the compensatory tube current
increase is thought to be less important for maintaining image quality in smaller
patients.164 Another drawback of X-Care is that dose to the spine can increase.
Although spine dose was not evaluated in this study, because it is expected to be
increased by kVp splitting, spine dose should be evaluated in the future when
considering implementation of the kVp splitting technique; this would require the
spine to be contoured in the patient models. The effectiveness of kVp splitting
could also be evaluated in the context of dose savings to other superficial
radiosensitive organs (e.g., the lens of the eye and thyroid), which would require
these organs to be contoured as well.

Although tube voltage modulation has been mentioned in prior studies,159
to the best of our knowledge, dose reduction has not been previously quantified
for tube voltage modulation or kVp splitting (which is a form of tube voltage
modulation). However, at least one study has addressed image reconstruction for
tube voltage modulation.169 While image quality could not be evaluated in this
study with physical data, image reconstruction would likely pose the greatest
challenge to implementation of kVp splitting. By applying an empirical cupping
correction algorithm to the raw data, Ritschl et al. showed that equivalent image
quality was achieved compared to image acquired using a single kVp. However,
this study did not specify the exact shift in the CT numbers between the
corrected image and the image acquired at a single kVp. Another potential image quality solution to implementation of kVp splitting would be to use a single kVp within each rotation (i.e., to flip from one kVp to another rather than to split the kVp into two different values); this approach would be similar to z-axis TCM. In terms of dose savings, the results in anthropomorphic phantoms were inconsistent when “flipping” to a different kVp for the entire rotation exposing the breasts (dose savings was achieved using 80 kVp for the entire rotation in the pediatric phantoms, but not in the adult phantom; data not shown); MC simulations could be used for further evaluation. While flipping the kVp would resolve issues with image reconstruction, the CT numbers would vary across images, which could lead to problems with interpretation of these values over an entire exam. If the issues with reconstruction and CT number correction were resolved, kVp splitting with more than two kVps could be explored.

In addition to addressing image reconstruction and CT number correction, if kVp splitting was eventually implemented in CT scanners, a means of identifying the breast tissue or other sensitive organs would be needed. To identify target tissues, technologists could locate and mark a specific anatomical landmark, possibly from the localizer image (as shown in Figure 5.9). Then, an appropriate age- or size-based preset could be selected to specify the kVps and region of coverage. Although only four kVp options are available on most scanners, and the minimum is 80 kVp (for the GE scanners involved in this study), in the future, additional kVp options (e.g., 60 kVp) could be added to scanners to expand the selection of kVp splitting schemes.
5.5. CONCLUSION

Breast tissue is one of the most radiosensitive tissues in the human body. The results of this study indicate that CT breast dose could be reduced by as much as 34% in pediatric patients by varying the tube voltage during those rotations that directly expose breast tissue. The development of novel techniques like kVp splitting for reducing pediatric breast dose may decrease the incidence of radiation-induced breast cancer in adult women who underwent CT scans as children.
CHAPTER 6
ORGAN-BASED BEAM FILTRATION: A NOVEL APPROACH TO REDUCING CT BREAST DOSE

**Purpose:** We sought to assess the effectiveness of a novel CT radiation dose-reduction strategy in which additional filtration was placed in between the x-ray beam and the breast area of three sizes of anthropomorphic phantoms.

**Methods:** Since dynamic organ-based x-ray beam filtration could not easily be added to the rotating x-ray tube, we emulated this technique by placing static filtration over the CT scanners' Mylar windows according to the location of the representative breast tissue of anthropomorphic phantoms. We initially scanned anthropomorphic phantoms representative of a 5-year-old, 10-year-old and an adult female without additional filtration using clinical TCM protocols with and without ASiR. We then scanned the phantoms with in-plane bismuth breast shielding placed over the phantoms' breast areas or copper foil or lead foil filtration placed over the scanners' Mylar windows. To detect entrance radiation exposure, we used a Farmer ionization chamber placed on the phantoms' surfaces at locations representing the sternum and left breast; at least 10 measurements were collected at each location, and the mean dose was calculated. Dose reduction was then determined by calculating the percent difference between the mean dose in unshielded scans compared to the mean dose in scans shielded by bismuth or copper or lead foil filtration. Additionally, we sampled CT numbers and noise at regions representing the lung and the soft
tissue near the sternum, left breast, and spine in CT images of the phantoms during unshielded scanning compared to scanning with bismuth breast shielding, copper foil filtration, or lead foil filtration in place.

**RESULTS:** In the TCM scans without ASiR, entrance dose reduction at the sternum and left breast in the three anthropomorphic phantoms ranged from 22% to 41%, from 28% to 42%, and from 54% to 60% when using in-plane bismuth breast shields, copper foil filtration, and lead foil filtration, respectively. The addition of 40% ASiR to the TCM protocols increased the dose reduction ranges to 51 - 68%, 53 - 66%, and 71 - 79% when using in-plane bismuth breast shields, copper foil filtration, and lead foil filtration, respectively. Copper foil filtration affected the CT image noise and CT numbers less than the bismuth breast shields and lead foil filtration (8.2% mean increase in noise versus 23% and 32%, respectively).

**CONCLUSION:** Breast surface dose-reductions between 28% and 66% were achieved in anthropomorphic phantoms with a minimal impact on image quality by placing copper foil filtration between the phantoms' breast areas and the x-ray beam. It is expected that similar dose reduction could be achieved in patients by incorporating a dynamic filter into CT scanners that covers the x-ray beam output port when passing over patients' breast tissue.
6.1. INTRODUCTION

The breast tissue weighting factor published in a recent report by the ICRP increased substantially relative to the previously-published value (from 0.05 to 0.12), and so breast tissue is now considered to be among the most radiosensitive organs in the body. Breast tissue is incidentally exposed to radiation during many routine CT scans, including coronary angiography and thoracic CT, but is rarely the organ of interest. Because patients undergoing CT scanning may be at higher risk for developing breast cancer than previously thought, CT breast dose must be given additional consideration. While overall dose-reduction strategies (e.g., TCM and iterative reconstruction) have been implemented across CT scanner manufacturers, few organ-based strategies exist.

X-care (Siemens Healthcare) is a strategy that was recently developed to reduce CT breast dose. In this method, the x-ray tube current is turned off (i.e., 0 mA) or greatly reduced at angular tube positions where the beam is directly exposing the breast tissue. Disadvantages of X-Care are that the x-ray tube must sample data, and thus deliver radiation, over a minimum of 180° plus the fan angle to reconstruct CT images, which limits the effectiveness of this technique in patients whose breast tissue extends beyond a 180° region, and to date X-Care has only been implemented by one manufacturer. Another strategy specifically aimed at reducing CT breast dose is to cover the breast tissue with in-plane shielding during scanning. Breast tissue is relatively superficial (i.e., there is a
lack of overlying tissue) and absorbs much of the radiation dose from the x-ray beam's low-energy photons, which are unlikely to penetrate through the patient and are instead likely to be attenuated upon entering the body. Thus, in-plane breast shields (typically bismuth-based) are used to partially attenuate the x-ray beam before it enters the body, reducing dose to the underlying tissue. Unlike X-Care, bismuth breast shielding functions independently of the CT scanner and thus can be used universally across scanners of different models and from different manufacturers. In two studies of bismuth breast shielding, diagnostic quality CT images (with no difference in image quality between shielded and unshielded lung) were obtained in pediatric and adult female patients, and dose reductions of 29% and 41%, respectively, were achieved in the corresponding phantom scans.\textsuperscript{171, 172} Another study revealed that combining bismuth breast shielding with TCM reduced dose to the representative breast tissue of a pediatric anthropomorphic phantom by 52%; however, this dose reduction was only achieved if the breast shield was placed on the patient after the localizer (also known as the scout) was acquired.\textsuperscript{173}

Although bismuth breast shields have been reported to reduce CT radiation dose to breast tissue, degraded quality of the CT images has also been reported, which has limited the acceptance of this technology by some medical physicists.\textsuperscript{174} In one study, dose reductions of almost 50% achieved using bismuth breast shielding on a semi-anthropomorphic phantom (scanned with a thoracic CT protocol) were accompanied by increased noise and streak artifacts in the CT images.\textsuperscript{175} However, such artifacts, which have been observed in
images acquired with 1-cm or smaller gaps between the bismuth breast shield and the patient surface, may be eliminated by using additional foam padding to increase the gap between the shield and the patient; equivalent noise between shielded and unshielded regions can also be achieved by increasing the gap.\textsuperscript{176, 177} Shifts in CT numbers have been associated with the use of bismuth breast shielding.\textsuperscript{177} The impact of breast shielding on CT numbers is important to consider because breast shielding could potentially be in place during cardiac CT and one use of CT numbers is for clinical interpretation of coronary calcium scoring. Another criticism of in-plane breast shields is that they attenuate the x-ray beam not only before it enters the patient (i.e., when the x-ray tube is facing the patient’s anterior), but also after the beam exits the patient (i.e., when the tube is facing the patient’s posterior); Figure 6.1 shows the entrance and exit beam. This is problematic because the exit beam “carries” information about the patient’s anatomy to the detectors, which hinders image quality by reducing the number and increasing the average energy of transmitted photons that reach the detectors. This results in more image noise and a further shift in CT numbers, and can also lead to artifacts. Furthermore, shielding the exit beam does not reduce patient dose since the beam has already been transmitted through the patient.
Incorporating organ-based beam filtration into CT scanners may resolve several of the drawbacks of in-plane breast shielding and X-Care while maintaining their dose-saving benefits. Similar to in-plane breast shielding, a physical filter made of a dense material (e.g., lead, copper, etc.) could be placed in the x-ray beam’s path when it is directly exposing a patient’s breast tissue (as shown Fig. 6.2). Filtering an x-ray beam preferentially removes low-energy photons, which are thought to contribute to patient dose but not image quality, consequently increasing the beam’s penetrability, and thus reducing absorbed dose to the breast tissue.\(^6\) On the other hand, reducing the number of photons may increase image noise, and increasing the mean energy of photons in the beam may reduce image contrast.\(^{178}\) However, the effect of organ-based beam filtration on image quality remains unclear.
FIGURE 6.2. Direct exposure of radiosensitive breast tissue occurs over a portion of the x-ray tube's 360° rotation around a patient. During this portion of the rotation, filtration could be placed over the x-ray beam to reduce CT breast dose.

The purpose of this study was to investigate the effect of organ-based beam filtration on CT dose and image quality. Since the proposed technology (i.e., organ-based beam filtration) has not yet been implemented in CT scanners, we emulated the technology by securing lead and copper foil over a portion of the scanners' Mylar windows. Pediatric and adult anthropomorphic phantoms were then scanned using clinical chest TCM protocols with and without ASiR to evaluate the potential of organ-based beam filtration in combination with existing dose-reduction strategies.
6.2. MATERIALS AND METHODS

6.2.1. PHANTOMS AND CT SCANNERS

Two state-of-the-art CT scanners and three anthropomorphic phantoms were used in this study to evaluate the proposed breast dose-reduction strategy of organ-based beam filtration across phantoms representative of patients of different sizes. A LightSpeed VCT scanner and a Discovery CT750 HD scanner (GE Healthcare) were used to scan anthropomorphic ATOM family dosimetry phantoms representative of a 5-year-old, 10-year-old, and an adult upper torso (with breast attachments) (CIRS). Each phantom was placed on the patient tabletop and centered within the bore of the scanners' gantries using the scanners' built-in laser lights, as shown in Figure 6.3; the CT images were used to verify centering.

**Figure 6.3.** Phantoms were centered within the scanners' bores using the built-in laser lights. The 10-year-old phantom is shown within the bore of the Discovery CT750 HD scanner's gantry.
6.2.2. SCANNING TECHNIQUES

After each phantom was properly centered, the TCM schemes were determined from the scout scans, which were acquired without any added filtration or shielding. A technique of 10 mA and 80 kVp was used for the pediatric phantom scouts, while 10 mA and 120 kVp were used for the adult phantom scout. All scouts were acquired using a 180° scout plane (i.e., a posteroanterior projection). The phantoms were then scanned with a helical pass from the chin to roughly the diaphragm using the clinical TCM protocols listed in Table 6.1; the scout images in Figure 6.4 show the specified image lengths for each phantom. For the GE scanners involved in this study, we specified both a noise index (NI) and the range (i.e., minimum and maximum) of x-ray tube currents within which the tube current could be modulated. Using the scout, the scanner then calculated the TCM scheme that achieved the target noise in each reconstructed image (as indicated by the NI). On the Discovery CT 750 HD scanner, 40% ASiR was also employed during the scans (the 40% level was chosen on the basis of the recommendations from Karen Procknow of GE Healthcare [via personal communication, 3/4/2011]). The TCM protocols were adjusted to account for ASiR by scaling the noise index by a factor of 1.3. Because ASiR was not installed on the LightSpeed VCT scanner, it was not used during those scans.
**Figure 6.4.** Localizer (scout) images showing the image lengths used when scanning the (a) 5-year-old, (b) 10-year-old, and (c) (female) adult anthropomorphic phantoms. The scout images shown here were acquired on the LightSpeed VCT scanner, but the same image lengths were also used for the Discovery CT750 HD scanner.

**Table 6.1.** The TCM protocols used when scanning the pediatric and adult anthropomorphic phantoms; 40-mm beam coverage, 2.5-mm helical image thickness, 0.984 pitch, and 0.4-s rotation times were used for all scans.

<table>
<thead>
<tr>
<th>Phantom</th>
<th>Noise index*</th>
<th>mA range</th>
<th>kVp</th>
<th>DFOV (cm)</th>
<th>SFOV</th>
</tr>
</thead>
<tbody>
<tr>
<td>5-year-old</td>
<td>21 or 27.3</td>
<td>35 – 230</td>
<td>80</td>
<td>25</td>
<td>Pediatric body</td>
</tr>
<tr>
<td>10-year-old</td>
<td>21.5 or 27.95</td>
<td>35 – 300</td>
<td>100</td>
<td>29</td>
<td>Pediatric body</td>
</tr>
<tr>
<td>Adult</td>
<td>22 or 28.6</td>
<td>35 – 360</td>
<td>120</td>
<td>50</td>
<td>Medium body</td>
</tr>
</tbody>
</table>

*The higher NI listed was used for scans with ASiR (performed on the Discovery CT750 HD scanner only) whereas the lower NI was used for scans without ASiR.
6.2.3. **UNSHIELDING SCANNING**

The phantoms were initially scanned and exposures were detected without any added filtration or shielding, using the protocols listed in Table 6.1.

6.2.4. **IN-PLANE BISMUTH BREAST SHIELDING**

Scanning was repeated (using the protocols in Table 6.1) and exposure measurements were collected with in-plane bismuth breast shields (AttenuRad CT breast shield system; F&L Medical Products, Vandergrift, PA) placed on the phantoms' surfaces (see Fig. 6.5). The approximate dimensions of the breast shields used on the 5-year-old, 10-year-old, and adult phantoms were 28 cm x 8 cm, 33 cm x 10 cm, and 25 cm x 20 cm (per side), respectively. The breast shields used on the pediatric phantoms and adult phantom contained 0.5 mm of bismuth (0.03-mm lead equivalence) and 1 mm of bismuth (0.06-mm lead equivalence), respectively, impregnated in synthetic rubber. Each shield was mounted to a 1-cm foam base.
FIGURE 6.5. Placement of appropriately-sized bismuth breast shields on the (a) 5-year-old, (b) 10-year-old, and (c) (female) adult phantoms.

6.2.5. ORGAN-BASED BEAM FILTRATION

Subsequently, scanning was repeated with high-purity (≥ 99.9%) copper or lead foil placed over the scanners' Mylar windows, as shown in Figure 6.6, and exposures were recorded (using the protocols listed in Table 6.1). Copper and lead are commonly used in radiological applications and could be acquired relatively inexpensively in high-purity forms with the desired dimensions. Width of the copper and lead foils (i.e., 76.2 mm) matched that of the Mylar window; the lengths of the foils were sized to match the angular region covered by the in-plane bismuth breast shields, which was approximately 140° for the 5-year-old and adult phantoms and 150° for the 10-year-old phantom (see Fig. 6.7).
Figure 6.6. The lead and copper foil tape covered a 140° arc of the CT scanners' Mylar windows when scanning the 5-year-old and adult phantoms and a 150° arc when scanning the 10-year-old phantom. As shown here, the lead foil is placed over the Mylar window of the GE LightSpeed VCT scanner.

Figure 6.7. An arc of approximately 150° was covered by the bismuth breast shield when scanning the 10-year-old phantom, therefore the top 150° of the Mylar window was covered with lead or copper foil when scanning the 10-year-old phantom.
6.2.5.1. TRANSMISSION MEASUREMENT

Radiation transmitted through the bismuth breast shields, copper foil, and lead foil was determined using the same setup as for HVL estimation; this setup is described in 2.2.1. and shown in Figure 2.1(a). However, instead of aluminum filters, the breast shields and varying thicknesses of the copper foil and lead foil were placed at the bottom of the gantry, covering the x-ray tube output port. The transmission percentages (for a 120 kVp beam) were calculated relative to the exposure measured for an unattenuated beam (i.e., nothing at the bottom of the gantry). The calculated transmission percentages were used to determine the thicknesses of copper foil that would give roughly the same transmission as through the bismuth breast shields.

6.2.5.2. COPPER FOIL FILTRATION

Based on the results of the transmission measurements, four and eight layers of 0.0356-mm thick copper foil (CFL-5A copper foil tape; J.V. Converting Company, Inc., Fairless Hills, PA) were placed over the Mylar window when scanning the pediatric phantoms and the adult phantom, respectively.

6.2.5.3. LEAD FOIL FILTRATION

Because the 0.127-mm thickness of the lead foil (LF-5A lead foil tape; J.V. Converting Company, Inc.) was greater than the lead equivalence of bismuth in the breast shields, one layer of lead foil was placed over the Mylar window for all three phantoms.
6.2.6. DOSE ANALYSIS

Radiation entrance (skin) exposures were measured without added beam filtration or shielding, with in-plane bismuth breast shielding, with copper foil beam filtration, and with lead foil beam filtration. A Farmer ionization chamber connected to an electrometer (RadCal Corporation) was used to detect exposure. The Farmer chamber was placed on the phantoms' surfaces at locations representing the sternum and left breast, as shown in Figure 6.8. For the 5-year-old and 10-year-old phantoms, the active detection region of the Farmer chamber was placed approximately 4 cm and 5 cm, respectively, to the left of the sternum; for the adult phantom, the active detection region was placed at the nipple. Ten exposure measurements were collected at each location and averaged. Exposures (in units of R) were converted to doses (in units of mGy) by using an exposure-to-dose conversion factor (f-factor) of 0.94 rad/R, which is the appropriate f-factor for tissue dose estimates, and then multiplying by 10.\textsuperscript{136}
FIGURE 6.8. Entrance exposure was measured using a Farmer chamber placed at the sternum and left breast of a (a) 5-year-old, (b) 10-year-old, and (c) (female) adult anthropomorphic phantom. As shown, the Farmer chamber is placed at the sternum for the pediatric phantoms and at the left breast for the adult phantom.

Dose reductions were calculated as the percent differences between the mean doses with bismuth breast shielding, copper foil beam filtration, or lead foil beam filtration compared to scans without shielding or filtration (for each phantom, scanner, and dose measurement location). While all scans included TCM, the scans on the Discovery CT750 HD also incorporated ASiR; for both scanners, dose reductions were calculated relative to the unshielded scan with TCM only (i.e., without ASiR). Dose reductions for the unshielded scans with TCM and ASiR were also calculated to determine the dose savings from ASiR alone.
6.2.7. IMAGE QUALITY ANALYSIS

To assess image quality, regions of interest (ROIs) were sampled in the CT images in the anterior soft tissue adjacent to the sternum and in the left breast, in the posterior soft tissue near the spine, and in the lung; Figure 6.9 shows the sizes and specific locations of these four ROIs for each phantom. We compared the CT numbers and noise level across the unshielded scans, scans with in-plane bismuth breast shielding, and scans with lead or copper foil filtration. The mean percent difference in the noise and shift in the CT numbers among the three phantoms, both scanners, and the four ROIs were determined for each shielding/filtration scenario compared to for the unshielded scan. While the relative percent differences in the noise values across shielding conditions can be directly calculated, because percent difference depends on the baseline value of the comparison, absolute differences in the CT numbers were considered. For example, the relative percent difference between the CT numbers 2 and 10 is 400%, but the difference between 200 and 208 is 4%, even though the absolute difference between them is the same. Furthermore, when considering the effect of shielding or filtration on the CT number, we are concerned with the magnitude of the shift, particularly in relation to the noise level. The CT number shift was not considered to be substantial if the CT number was within the noise (i.e., one standard deviation) of the mean CT number in an unshielded scan (e.g., if the mean CT number is 400 and the noise is 15, then CT numbers between 385 and 415 were not considered to be substantially different).
FIGURE 6.9. ROIs were drawn at the sternum (labeled “1”), the left breast (labeled “2”), in the posterior soft tissue near the spine (labeled “3”), and in the lung (labeled “4”) in the (a) 5-year-old (ROIs “1”, “2”, and “3” were 0.61 cm$^2$; ROI “4” was 4.3 cm$^2$), (b) 10 year old (ROI “1” was 0.36 cm$^2$; ROIs “2” and “3” were 0.65 cm$^2$; ROI “4” was 4.0 cm$^2$), and (c) adult phantom (ROI “1” was 0.27 cm$^2$; ROIs “2” and “4” were 4.3 cm$^2$; ROI “3” was 2.4 cm$^2$).
6.3. RESULTS

6.3.1. TRANSMISSION MEASUREMENTS

The percentages of radiation transmitted through the bismuth breast shields, the copper foil filtration, and the lead foil filtration are shown in Table 6.2. When we folded the adult breast shield in half in order to cover the x-ray beam with two layers of the shield (i.e., 0.12 mm lead equivalence), the beam attenuation was slightly more than that achieved with one layer of 0.127-mm thick lead foil.

**Table 6.2.** Percentages of exposure transmitted through bismuth breast shields and foil filtration. The transmission percentages were calculated relative to exposure measured for an unattenuated beam.

<table>
<thead>
<tr>
<th>Filtration or shielding</th>
<th>Transmitted exposure (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>5-year-old bismuth breast shield</td>
<td>71</td>
</tr>
<tr>
<td>10-year-old bismuth breast shield</td>
<td>73</td>
</tr>
<tr>
<td>Adult bismuth breast shield</td>
<td>54</td>
</tr>
<tr>
<td>Adult bismuth breast shield (folded over [i.e., 2 layers])</td>
<td>34</td>
</tr>
<tr>
<td>Lead foil tape (1 layer)</td>
<td>38</td>
</tr>
<tr>
<td>Lead foil tape (2 layers)</td>
<td>19</td>
</tr>
<tr>
<td>Copper foil tape (1 layer)</td>
<td>91</td>
</tr>
<tr>
<td>Copper foil tape (4 layers)</td>
<td>72</td>
</tr>
<tr>
<td>Copper foil tape (8 layers)</td>
<td>55</td>
</tr>
</tbody>
</table>
6.3.2. DOSE ANALYSIS

The radiation doses and percentages of dose reduction achieved for all phantoms and shielding conditions on the LightSpeed VCT and Discovery CT750 HD scanners are given in Table 6.3. The reduction in entrance dose achieved with lead foil filtration ranged from 54% to 80% across all phantoms, both scanners, and both measurement locations (sternum and left breast); in every case, lead foil filtration achieved the greatest reduction in entrance dose. The range of dose savings from the copper foil filtration (28 - 66%) was roughly the same as that of the bismuth breast shields (22 - 68%). Across all shielding scenarios and phantoms, dose reduction was higher for the scans performed on the Discovery CT750 HD than those performed on the LightSpeed VCT because of the added effect of ASiR, with dose reductions due to ASiR alone ranging from 30% to 44%; these dose reductions agreed with expectations based on the ASiR level chosen, which was expected to reduce dose by 40%. Despite several differences between the two scanners (e.g., the Discovery CT750 HD scanner uses gemstone detector technology, while the LightSpeed VCT does not), the detected radiation doses and the mA values used in scanners’ TCM schemes (without ASiR) were similar.
TABLE 6.3. Calculated dose estimates (in mGy) and the corresponding percentages of dose savings (in parentheses) for all three phantoms, across all shielding conditions, on both scanners; all scans used the clinical TCM and ASiR protocols listed in Table 6.1. Dose savings for each phantom and measurement location were calculated relative to the scan without filtration, shielding, or ASiR; therefore, the percentage of dose savings is not listed for the “No shielding (TCM)” scan.

<table>
<thead>
<tr>
<th>Dose measurement location</th>
<th>Shielding condition</th>
<th>LightSpeed VCT*</th>
<th>Discovery CT750 HD†</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>5-year-old</td>
<td>10-year-old</td>
</tr>
<tr>
<td>Sternum</td>
<td>No shielding (TCM)</td>
<td>2.59</td>
<td>2.05</td>
</tr>
<tr>
<td></td>
<td>No shielding (TCM &amp; ASiR)</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td></td>
<td>Bismuth breast shield</td>
<td>1.72 (34%)</td>
<td>1.59 (23%)</td>
</tr>
<tr>
<td></td>
<td>Copper foil</td>
<td>1.70 (34%)</td>
<td>1.47 (28%)</td>
</tr>
<tr>
<td></td>
<td>Lead foil</td>
<td>1.04 (60%)</td>
<td>0.83 (60%)</td>
</tr>
<tr>
<td>Left breast</td>
<td>No shielding (TCM)</td>
<td>2.42</td>
<td>1.93</td>
</tr>
<tr>
<td></td>
<td>No shielding (TCM &amp; ASiR)</td>
<td>N/A</td>
<td>N/A</td>
</tr>
<tr>
<td></td>
<td>Bismuth breast shield</td>
<td>1.58 (35%)</td>
<td>1.50 (22%)</td>
</tr>
</tbody>
</table>
**TABLE 6.3 continued**

<table>
<thead>
<tr>
<th>Dose measurement location</th>
<th>Shielding condition</th>
<th>LightSpeed VCT(^\dagger)</th>
<th>Discovery CT750 HD(^\dagger)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>5-year-old</td>
<td>10-year-old</td>
</tr>
<tr>
<td>Left breast</td>
<td>Copper foil</td>
<td>1.63</td>
<td>1.36</td>
</tr>
<tr>
<td></td>
<td></td>
<td>(33%)</td>
<td>(30%)</td>
</tr>
<tr>
<td></td>
<td>Lead foil</td>
<td>0.96</td>
<td>0.81</td>
</tr>
<tr>
<td></td>
<td></td>
<td>(60%)</td>
<td>(58%)</td>
</tr>
</tbody>
</table>

\(^\dagger\) For the GE LightSpeed VCT, the mA range used in the TCM scheme for the 5 year-old, 10 year-old, and adult phantom was 46 – 123 mA (mean: 96 mA; mean over chamber: 97 mA), 41 – 78 mA (mean: 64 mA; mean over chamber: 53 mA), and 37 - 259 mA (mean: 191 mA; mean over chamber: 195 mA), respectively. Mean mA was determined by averaging the mA values as they appeared on CT images. This TCM scheme was also used for the scans performed using bismuth breast shielding and with added filtration.

\(^\dagger\) For the GE Discovery CT750 HD, the mA range used in the TCM scheme (without ASiR) for the 5 year-old, 10 year-old, and adult phantom was 47 – 111 mA (mean: 88 mA; mean over chamber: 86 mA), 36 – 77 mA (mean: 63 mA; mean over chamber: 51 mA), and 37 - 245 mA (mean: 182 mA; mean over chamber: 182 mA), respectively. This TCM scheme was also used for the scans performed using bismuth breast shielding and with added filtration. The mA range used in the TCM scheme (with ASiR) for the 5 year-old, 10 year-old, and adult phantom was 35 – 66 mA (mean: 52 mA; mean over chamber: 51 mA), 35 – 46 mA (mean: 38 mA; mean over chamber: 35 mA), and 38 - 145 mA (mean: 109 mA; mean over chamber: 107 mA), respectively. Mean mA was determined by averaging mA values as they appeared on CT images.
6.3.3. Image Quality Analysis

Table 6.4 lists the CT numbers and noise in the CT images. Although the lead foil filtration resulted in the highest dose reduction, it also caused the greatest shift in CT numbers compared to unshielded scans. The mean absolute shifts in CT numbers (relative to unshielded scans) across the four ROIs, three phantoms, and two scanners were 18.3, 10.3, and 32.4 HU for scans with bismuth breast shields, copper foil filtration, and lead foil filtration, respectively. For the most part, the shift in the CT numbers produced by the presence of copper foil filtration was within one standard deviation of the CT number of the unshielded scans. The mean percentages of increase in the noise (across ROIs, phantoms, and scanners) compared to the noise in the unshielded scans were 23%, 8.2%, and 32% for the breast shields, copper foil, and lead foil, respectively. Across only the two anterior ROIs (left breast and sternum), the mean absolute shifts in CT numbers were 31.4, 11.6, and 38.5 HU for scans with bismuth breast shields, copper foil filtration, and lead foil filtration, respectively; the corresponding mean percentages of increase in noise were 36.4%, 7.7%, and 39.4%, respectively. The mean shift in the CT numbers in scans with ASiR compared to scans without ASiR was less than 1 HU, and the images reconstructed with ASiR had 6% less noise on average than the images reconstructed without ASiR.
**TABLE 6.4.** CT numbers (HU) and noise (shown in parentheses) measured in the CT images of the three anthropomorphic phantoms for ROIs drawn at the sternum, left breast, and in the posterior soft tissue near the spine. The specific location of each ROI can be seen in Figure 6.9.

<table>
<thead>
<tr>
<th>ROI location</th>
<th>LightSpeed VCT</th>
<th>Discovery CT750 HD</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Unshielded</td>
<td>Breast shield</td>
</tr>
<tr>
<td></td>
<td>(no ASiR)</td>
<td>(ASiR)</td>
</tr>
<tr>
<td>Sternum</td>
<td>2.13 (17.5)</td>
<td>24.0 (23.6)</td>
</tr>
<tr>
<td>Left breast</td>
<td>-5.02 (14.0)</td>
<td>13.2 (22.9)</td>
</tr>
<tr>
<td>Posterior soft tissue</td>
<td>2.21 (20.5)</td>
<td>2.87 (24.9)</td>
</tr>
<tr>
<td>Lung</td>
<td>-798 (13.1)</td>
<td>-786 (14.3)</td>
</tr>
</tbody>
</table>

**5-year-old phantom**

<table>
<thead>
<tr>
<th>ROI location</th>
<th>LightSpeed VCT</th>
<th>Discovery CT750 HD</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Unshielded</td>
<td>Breast shield</td>
</tr>
<tr>
<td></td>
<td>(no ASiR)</td>
<td>(ASiR)</td>
</tr>
<tr>
<td>Sternum</td>
<td>13.5 (17.4)</td>
<td>30.7 (21.4)</td>
</tr>
<tr>
<td>Left breast</td>
<td>-15.4 (15.7)</td>
<td>17.7 (24.2)</td>
</tr>
<tr>
<td>ROI location</td>
<td>10-year-old phantom</td>
<td></td>
</tr>
<tr>
<td>----------------------</td>
<td>---------------------</td>
<td>----------</td>
</tr>
<tr>
<td></td>
<td>LightSpeed VCT</td>
<td>Unshielded</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Unshielded</td>
</tr>
<tr>
<td>Posterior soft tissue</td>
<td>10.9</td>
<td>11.7</td>
</tr>
<tr>
<td></td>
<td>(21.2)</td>
<td>(26.2)</td>
</tr>
<tr>
<td></td>
<td>(13.3)</td>
<td>(16.3)</td>
</tr>
<tr>
<td>Adult phantom</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Sternum</td>
<td>32.4</td>
<td>58.1</td>
</tr>
<tr>
<td></td>
<td>(19.4)</td>
<td>(15.1)</td>
</tr>
<tr>
<td>Left breast</td>
<td>-42.2</td>
<td>27.1</td>
</tr>
<tr>
<td></td>
<td>(16.0)</td>
<td>(26.2)</td>
</tr>
<tr>
<td>Posterior soft tissue</td>
<td>-2.26</td>
<td>-0.44</td>
</tr>
<tr>
<td></td>
<td>(20.5)</td>
<td>(20.1)</td>
</tr>
<tr>
<td>Lung</td>
<td>-791.5</td>
<td>-786</td>
</tr>
<tr>
<td></td>
<td>(10.0)</td>
<td>(12.4)</td>
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</table>
The CT images reconstructed for the 5-year-old, 10-year-old, and adult phantoms appear in Figures 6.10 – 6.12 for the GE LightSpeed VCT scanner and Figures 6.13 – 6.15 for the GE Discovery CT750 HD scanner. Streak artifacts appeared in the CT images of the adult phantom when in-plane bismuth breast shielding was used. Although streak artifacts also appeared in the CT images of the adult phantom when copper or lead foil filtration was used, the artifacts were at the edge of the DFOV in the images, beyond the anatomy of the phantom.
FIGURE 6.10. CT images (mediastinum window level) of the 5-year-old phantom scanned on a GE LightSpeed VCT scanner with (a) no shielding, (b) bismuth breast shielding, (c) copper foil beam filtration, and (d) lead foil filtration; the phantom was scanned using the clinical pediatric chest protocol listed in Table 6.1.
FIGURE 6.11. CT images (mediastinum window level) of the 10-year-old phantom scanned on a GE LightSpeed VCT scanner with (a) no shielding, (b) bismuth breast shielding, (c) copper foil beam filtration, and (d) lead foil filtration; the phantom was scanned using the clinical pediatric chest protocol listed in Table 6.1.
Figure 6.12. CT images (mediastinum window level) of the (female) adult phantom scanned on a GE LightSpeed VCT scanner with (a) no shielding, (b) bismuth breast shielding, (c) copper foil beam filtration, and (d) lead foil filtration; the phantom was scanned using the clinical adult chest protocol listed in Table 6.1. Notice the artifact caused by the breast shield in image (b) and by the lead foil in image (d).
FIGURE 6.13. CT images (mediastinum window level) of the 5-year-old phantom scanned on a GE Discovery CT750 HD scanner with (a) no shielding (no ASiR), (b) no shielding (ASiR), (c) bismuth breast shielding, (d) copper foil beam filtration, and (e) lead foil filtration; the phantom was scanned using the clinical pediatric chest protocol listed in Table 6.1. As shown in (c), getting breast shield to lay flat on top of the Farmer chamber was difficult due to the rigidity of the shield, but this did not seem to affect the results in terms of dose savings.
FIGURE 6.14. CT images (mediastinum window level) of the 10-year-old phantom scanned on a GE Discovery CT750 HD scanner with (a) no shielding (no ASiR), (b) no shielding (ASiR), (c) bismuth breast shielding, (d) copper foil beam filtration, and (e) lead foil filtration; the phantom was scanned using the clinical pediatric chest protocol listed in Table 6.1.
FIGURE 6.15. CT images (mediastinum window level) of the (female) adult phantom scanned on a GE Discovery CT750 HD scanner with (a) no shielding (without ASiR), (b) no shielding (with ASiR), (c) bismuth breast shielding, (d) copper foil beam filtration, and (e) lead foil filtration; the phantom was scanned using the clinical adult chest protocol listed in Table 6.1. Notice the artifact caused by the breast shield in image (c) and by the copper and lead foil in images (d) and (e), respectively.
6.4. DISCUSSION

Traditionally, x-ray beam filtration has been constant within a given CT examination; however, if the technology proposed in this study (i.e., organ-based beam filtration) was implemented into scanners, it would enable beam filtration to be dynamically adjusted during scanning. To the best of our knowledge, no prior study has evaluated the effect of organ-based filtration of the primary x-ray beam. We found that copper foil filtration reduced the radiation dose by 28 - 66% while increasing the CT image noise by 8.2% on average. Therefore, copper foil beam filtration yields roughly the same dose reduction as that achieved with in-plane bismuth breast shielding but has a lesser impact on image quality (particularly the anterior image quality).

In our study, in-plane breast shielding resulted in a mean shift in CT number of 18.3 HU, with a maximum observed shift of 69.3 HU (across all measurement conditions), and a mean increase in noise of 26.5%. Our results can be compared to those obtained in a study by Kalra et al.\textsuperscript{177} Kalra et al. reported that bismuth breast shields with a 1-cm foam offset resulted in a shift in CT number of almost 100 HU and a 53.5% increase in the noise at the surface of a 30-cm anthropomorphic chest phantom. Inconsistency in the noise and CT number shift reported in our study and the Kalra et al. study could be attributed to the fact that we used GE scanners while Kalra et al. used a Siemens scanner. In our study, copper filtration yielded a dose reduction that was roughly the same as for the bismuth breast shielding; this is consistent with Kalra et al.'s finding that
dose did not significantly change when the distance between the shield and phantom was increased. Although both bismuth breast shields and copper foil beam filtration produced roughly the same dose reductions, the improved image quality (i.e., less shift in the CT numbers and less noise) from using copper foil compared with bismuth breast shields may also be explained by Kalra et al., who reported less shift in the CT numbers and less image noise when increasing the offset between the shield and phantom surface. Although both the current study and that of Kalra et al. noted that using breast shielding with a 1-cm foam offset resulted in streak artifacts in CT images, in the current study, neither the lead foil or copper foil filtration caused streak artifacts within the phantom anatomy.

The current study had several potential limitations. For example, both the lead foil and copper foil used in the study had an acrylic adhesive and a paper liner. Although that did not affect demonstration of proof of concept, the specific dose reduction might have been different if pure lead and copper (without adhesive) had been used. Presence of the adhesive and paper may explain why the bismuth shielding, which had a lead equivalence of 0.12 mm, had a lower percentage of transmission than 0.127 mm of lead; however, the fact that attenuation properties vary across the energy spectrum (and thus limit the meaning of the term “lead equivalent”) could also account for this inconsistency. Furthermore, differences in the amount of precision specified for the lead equivalence of the bismuth shield and the thickness of the lead foil may provide another explanation.
Another potential limitation of the current study was that we did not compare dose reductions from TCM with those of a static mA scan. Although TCM was used, we did not specifically quantify the effect of TCM on dose reduction because the same TCM schemes were used across scans and thus the same reductions would have been observed if static mA protocols had been used. The best attempt at comparison would be to use the maximum mA specified by the TCM protocol (to ensure dose reduction, the GE LightSpeed VCT scanner manual recommends setting the maximum mA in TCM protocols to the mA used in the static protocol) or the maximum observed mA (from the actual TCM scheme for each phantom). Because the mean mA over the active detection region of the Farmer chamber was at least 20% less than maximum mA used in the actual TCM scheme and 45% less than maximum mA specified in protocol (across all phantoms and both scanners), dose from using TCM would be at least 20% lower than from a static mA scan. When applying this additional dose reduction, a total breast dose reduction of at least 65% could be achieved by combining TCM, ASiR, and copper beam filtration. The dose reduction from TCM compared to a static mA scan would likely be even higher if the scout was performed using a plane of 0° (i.e., with the x-ray tube at the top of the gantry, which corresponds to an anteroposterior projection) rather than 180°. After data collection, we discovered (via personal communication with Karen Procknow of GE Healthcare [7/27/2010]) and confirmed with measured data that the mA values in the TCM scheme were generally lower when a scout plane of 0° was used, unless the mA values were close to the maximum or minimum of the TCM
range, in which case the scout plane did not matter. Although using a 180° scout plane did not specifically affect the conclusions of this study because all scans used the same TCM scheme within a phantom and scanner type, and we did not determine dose reduction relative to a static mA scan, the absolute doses would have likely been lower if a 0° scout plane had been used.

Despite the limitations of this study, our findings indicate that organ-based beam filtration (if incorporated into CT scanner design) would likely improve upon a number of the drawbacks typically associated with in-plane bismuth breast shields, including issues with timing of shield placement relative to scout acquisition and streak artifacts within the anatomy. While organ-based beam filtration may cause a small shift in CT numbers, this shift could be addressed by implementing correction algorithms or by using less attenuating beam filtration (although this would also result in less dose reduction). In addition, the shifts in CT numbers due to the copper foil filtrations were generally less than those caused by the bismuth breast shields, and radiologists are accustomed to reading images that have been affected by breast shields. Furthermore, if organ-based beam filtration were incorporated into CT scanners, the CT number shift would likely be smaller than that reported in the study because only the entrance beam would be filtered, whereas both the entrance and exit beam were filtered in the current study.

Organ-based beam filtration could be incorporated into CT scanners such that a physical filter made of a dense material (e.g., lead, copper, etc.) could be used to filter the primary x-ray beam when the beam is directly exposing a patient’s breast tissue. To implement this technology, a filter that can be quickly
placed and removed during scanning would have to be developed. One approach to filter design would be to use an aperture that opens and closes like the shutter on a camera at the output port of the x-ray beam; the aperture would close at a given angular position of the x-ray tube and then re-open at a second angular position based on the location of breast tissue. To allow for patient specificity, two aperture layers could be included in the design, thus enabling the use of two filter thicknesses. Because transmission depends on the beam energy, and because pediatric patients are typically scanned for 80 kVp or 100 kVp while adult patients are scanned for 120 kVp or 140 kVp, adding two thicknesses would allow for roughly the same percentage of dose reduction for all patients. When choosing the filter material, several factors need to be considered, including the photoelectric attenuation coefficient. While some studies have recommended the use of copper filters, the criteria for organ-based filtration would be different than the criteria used to determine inherent or bowtie filter materials since organ-based filtration would only be used to filter the beam while it is directed at breast tissue. In addition, since the filter would need to be placed and removed quickly while it rotates with the x-ray tube, a material that provides the desired level of dose reduction without being too thick (heavy) or thin (flimsy) should be selected to prevent strain on the scanner. Therefore, although copper was the better of the two filtration options evaluated in the current study in terms of balancing dose reduction and image quality and overcame many of the issues associated with bismuth breast shields, copper is
not necessarily the optimal filtration material to use when implementing dynamic filtration into scanners.

Another issue that would need to be addressed when considering implementation is the method of specifically targeting beam filtration to the breast tissue. Using the patient’s posteroanterior or anteroposterior scout, a technologist could outline the breast tissue (Fig. 6.16); from the outline, the start and stop z-axis locations, and thus the number of rotations over which to apply the filter, could be determined. This would enable the use of patient-specific organ shielding (in contrast to in-plane organ shielding, which is available in standard sizes). The angular region over which to apply the beam filter (within the xy-plane) could be calculated by the technologist outlining the breast on the patient’s lateral scout, or preset angular coverage regions could be employed. However, if the lateral scout were used for this purpose, the order of the scout acquisitions could be an issue since the TCM scheme is determined from the last scout performed.

**Figure 6.16.** The anteroposterior scout of the adult anthropomorphic phantom, which has the breast tissue outlined; outlining the breast tissue on the scout could be used to identify the location of breast tissue (along the z-axis) as the target for selective application of x-ray beam filtration.
Before dynamic organ-based beam filtration can be incorporated into CT scanner design, further analysis is needed to evaluate the effect of this technology on image quality in patients. The application of selective beam filtration for other superficial radiosensitive organs such as the eyes, thyroid, and testes should also be considered.

6.5. CONCLUSION

In this study, we found that copper foil beam filtration reduced entrance breast dose by 28 – 66% across three sizes of anthropomorphic phantoms, which was roughly the same dose reduction as that achieved with in-plane bismuth breast shielding. Furthermore, copper foil beam filtration was shown to have a lesser impact on image quality than in-plane bismuth breast shielding compared to scans without any added filtration or shielding (10.3 mean HU shift and 8.2% mean increase in noise for copper foil beam filtration versus 18.3 mean HU shift and 23% mean increase in noise for in-plane bismuth breast shielding); this was particularly true when considering image quality in the phantoms' anterior soft tissue and no streak artifacts were observed within the phantom in the scans with copper foil beam filtration. This study supports the development of new strategies to reduce CT radiation dose to breast and other radiosensitive tissues. Additionally, these findings support the combination of multiple strategies (including TCM and ASiR) to maximize dose reduction without impairing diagnostic image quality.
7.1. SUMMARY

The primary contributions of this thesis are as follows:

- An empirical model for describing CT attenuation was developed and validated ($R^2 > 0.99$). This model can be used to estimate the HVL and QVL, which are the most common metrics used to describe the penetrating ability of an x-ray beam.

- The precision of dosimetry-related data, including the HVL, CTDI$_{100}$, air, and CTDI$_w$, was evaluated over the course of one year using three different models of CT scanners and was found to be well within 5%. 

- The dose penalty from overscanning, due to stabilization of the x-ray tube output prior to scanning, was determined for three sizes of CTDI phantoms. Dose penalties ranging from 1% to 26% were observed when dose was measured near the angular position of the x-ray tube where acquisition began. Therefore, it is recommended that CT acquisitions begin with the x-ray tube at the bottom of the gantry to avoid imparting this penalty on radiosensitive tissues near patients’ anterior surfaces (e.g., breast tissue). 

- Monte Carlo modeling estimated that CT dose absorbed by the glandular breast tissue of pediatric females was reduced by up to 34% (while presumably maintaining constant image noise) from using kVp splitting
compared to scans using a static kVp. kVp splitting is a novel strategy in which kVp is varied during rotations when the x-ray beam is directly exposing the breast tissue.

- Placing copper foil filtration over CT scanners' Mylar windows, in between the x-ray beam and the representative breast tissue of pediatric and adult anthropomorphic phantoms, resulted in entrance (skin) dose reductions (at the left breast and sternum) ranging from 28% to 66%. These dose reductions were accompanied by an average increase of 8.2% in the CT image noise.

7.2. FUTURE WORK

The novel methods evaluated in this thesis for reducing CT breast dose showed promise; however, more research is needed to assess such strategies before implementation in CT scanners could be recommended. Additional dose quantification of both kVp splitting and organ-based beam filtration could be obtained through MC modeling. Specifically, the dose-saving potential of these strategies across scanners of different makes and models, which have different geometries and filtration, and thus which emit different levels of radiation (even for the same scan parameters) could be quantified through MC modeling. Furthermore, patient studies could be performed to evaluate the effect of organ-based beam filtration on image quality.
Besides the three voxelized patient models used to simulate kVp splitting, 22 additional models were developed from female pediatric PET/CT patient data (with institutional review board approval) during the course of this thesis. MC simulations of both kVp splitting and organ-based beam filtration could be performed using these 22 models to evaluate breast dose-savings. Additionally, simulations could be performed when combining the two strategies and also with the added effect of TCM or bismuth breast shielding. The TCM schemes of all 22 patients were obtained from the raw data, and thus could be included in future simulations. Most of the patients had bismuth breast shields in place during scanning; the breasts shields were contoured in the patient images (see Fig. 7.1) so that they could either be modeled as bismuth or “removed” from simulations by labeling them as air. Two of the patients had bean bag toys and these were also contoured (as shown in Fig. 7.2) so that they could be labeled as air if necessary (i.e., if they affect the dose estimates).
**Figure 7.1.** A CT image of one of the 25 pediatric female patients used to construct voxelized patient models, which shows the contoured breast shield (light green), glandular breast tissue (purple), and lung tissue (teal).

**Figure 7.2.** (a) The localizer image of one of the pediatric patients, whose CT images were used to construct voxelized patient models, shows the patient holding a bean bag toy during scanning; (b) the toy was contoured so that its effect could be "removed" from future MC simulations if necessary.
Performing additional MC analysis is important because of the limited physical data that can be collected to assess kVp splitting. Furthermore, the kVp splitting results obtained up to this point showed inconsistent dose reductions across patients, and so more data is needed to assess this technology's potential. MC modeling of organ-based beam filtration is also needed because the doses that we measured at the breast surface of anthropomorphic phantoms will likely differ from patient breast dose estimates obtained through MC modeling (as was the case in the kVp splitting study). This is because anthropomorphic phantoms cannot account for the broad range in the amount and distribution of breast tissue across patients (even patients of similar size and age). Figure 7.3 shows such differences in the breast tissue across two of the 25 pediatric female patients whose CT images were used to construct the voxelized patient models. Additionally, anthropomorphic phantoms may not accurately represent modern patient sizes in light of the increasing rates of obesity and the multicultural composition of the U.S. population.\textsuperscript{183}
Breast tissue and lung tissue were the only organs contoured in the voxelized patient models as part of this thesis; however, full-body scans were performed on many of the patients and all scans at least covered the anatomy from mid-skull and to the bottom of the knee. Therefore, the voxelized patient models could be expanded in the future to include other superficial radiosensitive organs, like the lens of the eye and the thyroid. Furthermore, organs within the same scan plane as these radiosensitive tissues could also be contoured. For instance, dose to the spine could be checked when implementing strategies over the breasts to ensure that the technology designed to decrease breast dose does not have the opposite effect on spine dose. Finally, because patients' arms are raised above the head in routine CT examinations, the arms of the voxelized
patient models, which were positioned next to the trunk, could be contoured and set to air to remove their effect on dose (due to attenuation and scatter).

New strategies for CT dose reduction are continually emerging, including the novel breast dose-reduction strategies of kVp splitting and organ-based beam filtration proposed in this thesis. While optimization of CT dose is ongoing, such advances in technology move us closer towards full realization of the ALARA principle.
8.1. APPENDIX A: DERIVATION OF EQ. (2.5)

Initially, Eq. (2.4) was substituted into Eq. (2.1) for \( \mu \):

\[
I = I_0 e^{-(\mu_0 + \lambda \frac{j}{I_0}) x} \tag{8.1}
\]

Eq. (8.1) was rearranged to achieve separation of variables:

\[
\left( \frac{l}{I_0} \right) e^{\lambda x \left( \frac{l}{I_0} \right)} = e^{-\mu_0 x} \tag{8.2}
\]

Both sides of Eq. (8.2) were multiplied by \( \lambda x \):

\[
\lambda x \left( \frac{l}{I_0} \right) e^{\lambda x \left( \frac{l}{I_0} \right)} = \lambda x e^{-\mu_0 x} \tag{8.3}
\]

Term \( u \) was created such that:

\[
u = \lambda x \left( \frac{l}{I_0} \right) \tag{8.4}
\]

Substituting \( u \) into Eq. (8.3):

\[
ue^u = \lambda x e^{-\mu_0 x} \tag{8.5}
\]
By definition of $W$, Eq. (8.5) becomes:

$$u = W(\lambda x e^{-\mu_0 x}) \quad (8.6)$$

Setting Eq. (8.4) equal to Eq. (8.6) and rearranging:

$$I = I_0 \frac{W(\lambda x e^{-\mu_0 x})}{\lambda x} \quad (2.5)$$

8.2. APPENDIX B: EQUATIONS FOR CALCULATING HVL AND QVL

Rearranging Eq. (8.1):

$$\left(\frac{I}{I_0}\right) = e^{-\left(\mu_0 + \lambda \left(\frac{1}{I_0}\right)\right)x} \quad (8.7)$$

Taking the natural logarithm of both sides of Eq. (8.7):

$$ln \left(\frac{I}{I_0}\right) = -\left(\mu_0 + \lambda \frac{1}{I_0}\right)x \quad (8.8)$$

Rearranging Eq. (8.8):

$$x = \frac{-ln \left(\frac{I}{I_0}\right)}{\mu_0 + \lambda \frac{1}{I_0}} \quad (8.9)$$
Eq. (8.9) can be written as the HVL or QVL by substituting $\frac{1}{2}$ or $\frac{1}{4}$, respectively, for $\left(\frac{1}{l_0}\right)$:

$$\text{HVL} = \frac{\ln(2)}{\mu_0 + \frac{\lambda}{2}} \quad (8.10)$$

$$\text{QVL} = \frac{\ln(4)}{\mu_0 + \frac{\lambda}{4}} \quad (8.11)$$

where $\mu_0$ and $\lambda$ are calculated as follows for a two-point Lambert W interpolation of measured data points $(x_1, \frac{l_1}{l_0})$ and $(x_2, \frac{l_2}{l_0})$:

$$\mu_0 = \frac{x_1 \left(\frac{l_1}{l_0}\right) \ln\left(\frac{l_2}{l_0}\right) - x_2 \left(\frac{l_2}{l_0}\right) \ln\left(\frac{l_1}{l_0}\right)}{x_1 x_2 \left[\left(\frac{l_2}{l_0}\right) - \left(\frac{l_1}{l_0}\right)\right]} \quad (8.12)$$

$$\lambda = \frac{x_2 \ln\left(\frac{l_1}{l_0}\right) - x_1 \ln\left(\frac{l_2}{l_0}\right)}{x_1 x_2 \left[\left(\frac{l_2}{l_0}\right) - \left(\frac{l_1}{l_0}\right)\right]} \quad (8.13)$$
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