A Thesis

entitled

An Investigation of Dose Reduction in Head and Neck CT with the Use of Organ Exposure Modulation

by

Kyle Luttrell

Submitted to the Graduate Faculty as partial fulfillment of the requirements for the Master of Science in Biomedical Science Degree in Medical Physics

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August 2016
An Abstract of

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Purpose: To investigate the use of organ exposure modulation (OEM) in head and neck CT to reduce the dose to radiosensitive organs such as the lens of the eye and thyroid.

Methods: The Toshiba Aquilion ONE CT scanner was used to perform multiple scans of the head and neck on a RANDO phantom using the clinically-used technique without OEM and an OEM technique. For the brain protocol, the scanner reported CTDI\text{VOL} measured in the head phantom, for the fixed mA and OEM techniques was 42.7 mGy and 37.3 mGy respectively. For the head/neck protocol, the scanner reported CTDI\text{VOL} measured in the body phantom, for the current technique and OEM technique was 5.4 mGy and 4.8 mGy respectively. Dose was measured with optically stimulated luminescent dosimeters (OSLDs) affixed to the surface of the RANDO phantom at position of the eyes, both sides of the head, and posterior of the head for the brain scan and the eyes and thyroid for the head/neck scan. The OSLDs had been calibrated free-in-air against a pencil ion chamber at the CT scanner isocenter. Additionally, the fixed mA and OEM techniques were used to scan the ACR phantom for the purpose of assessing image noise and contrast-to-noise ratio (CNR).
Results: On average dose to the lens of the eyes for the head scan using the fixed mA protocol was measured to be 42.0±4.9 mGy while the OEM technique was 29.5±4.6 mGy, representing a 29.8% dose reduction. Radiation dose to the sides and posterior of the head was on average reduced by 10.2±3.5%. Image noise evaluated in the ACR phantom was increased on average by 6.3%. The CNR using the fixed mA technique was found to be 1.82 while the OEM technique was 1.55. For the head/neck scan, the average dose to the lens of the eyes and the thyroid for the current protocol was measured to be 59.1±0.84 mGy and 17.4±1.2 mGy respectively. The dose to the eye lens and thyroid using an OEM protocol was measured to be 45.0±0.69 mGy and 21.8±0.9 mGy respectively, representing a dose savings of 23.8% to the eyes and an increase in dose of 25.6% to the thyroid. Image noise was increased on average by 16.9%. The CNR using the current protocol was found to be 1.123 while the OEM technique was 0.922.

Conclusions: For the head scan, dose to the lens of the eye was reduced, by using an OEM protocol, by 29.8%. The dose reduction corresponded to a slight increase in image noise and decrease in CNR. However, the CNR remains well above the acceptable criteria of 1.0 set forth by ACR for adult head scans. For the head/neck scan, dose to the lens of the eye was reduced by 23.8%, however dose to the thyroid increased by 25.6%. This dose reduction resulted in an increase in image noise and once again a decrease in CNR. CNR for the head and neck protocol was reduced below the minimum of 1.0 set forth by the ACR. All-in-all OEM protocols reduced dose to the eye lens with an increase in dose to the thyroid. Image quality remained relatively adequate with the exception of the CNR for the head and neck protocol. OEM would be a viable option for head protocols. With further development of the head and neck technique, CNR can be
increase above the minimum limit and the OEM protocol could become a viable option for head and neck protocols as well.
Acknowledgements

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I would also like to thank the Technologists in the Radiology department. Specifically those technologists in CT. Without them, I would not have been able to get the time to use the scanner in order to obtain the measurements presented in this paper.

I also thank the physicists at Toshiba for providing additional information pertaining to the Toshiba Aquilion ONE ViSION CT scanner.
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List of Abbreviations

ACR ...............................American College of Radiology
ALARA..........................As Low As Reasonably Achievable

CNR .............................Contrast-to-Noise Ratio
CT ...............................Computed Tomography
CTA .............................Computed Tomographic Angiography
CTDI ............................Computed Tomography Dose Index
CTDI_{vol} ........................Volume Computed Tomography Dose Index
CTDI_{w} ........................Weighted Computed Tomography Dose Index
CTDI_{100} .......................Computed Tomography Dose Index from 100mm pencil chamber

FOV ..............................Field of View

Gy ...............................Gray

ICRP .............................International Commission of Radiation Protection and Measurement
ICU ...............................Intensive Care Unit

kVp .............................Tube Voltage in kilovolts

LED .............................Light Emitting Diode

mA ...............................Tube Current in milliAmperes
mR ...............................Exposure in milliRoentgens

ODM .............................Organ Dose Modulation
OEM .............................Organ Exposure Modulation
OSLD ............................Optically Stimulated Luminescent Dosimeter

PMMA ............................Polymethylmethacrylic

ROI ...............................Region of Interest

SEMAR ..........................Single Energy Metal Artifact Reduction
SNR ...............................Signal-to-Noise Ratio

UTMC ...........................University of Toledo Medical Center
Chapter One

Introduction

Introduction to this Study

On average approximately 70 to 80 million Computer Tomography (CT) studies are performed each year in the United States of America [7]. At the University of Toledo Medical Center (UTMC), approximately 21,000 CT scans are performed in the average year. Of those 21,000 scans, nearly 5,700 are head scans, half of which are brain scans. Advances over the years have led to an increase in the number of CT scans, mostly due to acquisition times on the order of a couple of seconds. While advances in technology make CT a highly popular choice in the radiology department, attention must be paid to the absorbed dose to the lens of the eye. In 2007, the International Commission of Radiation Protection and Measurement (ICRP) published publication 103 which suggested a threshold dose of 5 Gray (Gy) to the lens of the eye for acute exposures and 8Gy for fractionated exposures before the formation of lens opacification (cataracts) [10]. According to Radimetrics, a dose analyzing software, at UTMC the average dose to the lens of the eye from a Head CT scan is 50mGy. Along with the dose threshold, this suggests that a patient may receive approximately 100 CT scans before developing cataracts of the eye. However, in 2012, ICRP publication 118 suggests a dose threshold of only 0.5Gy for acute and fractionated exposures [11]. This reduces the number of scans a patient may receive before cataract formation to approximately 10. Many patients can easily exceed 10 CT scans in a short period of time, for example stroke or neuro ICU patients. While the dose thresholds for cataract formations have been reduced, the number of CT scans performed should not be expected to be affected. With exceptional
image quality, a decline in the number of CT scans performed at a given clinic is unlikely.

Absorbed dose to the lens of the eye can vary from patient to patient based on individual anatomy. The current form of dose reduction to the lens of the eye is gantry tilt. The gantry is physically tilted, typically on the order of 10-20 degrees, in order to prevent the scanning of the orbitals of the patient. Many scanners have a gantry tilt range of up to +/- 30 degrees. However, with the continuing increase in the number of detectors in CT, the scan range continues to get wider. A wider scan range leads to a narrower tilt range. For example; Toshiba’s Aquilion PRIME, an 80 row scanner, is capable of +/- 30 degrees of gantry tilt. On the other hand, Toshiba’s Aquilion ONE, a 320 row scanner, is only capable of +/- 22 degrees of tilt [20]. While the concept of gantry tilt may reduce the lens dose to a number of patients, many patients still receive a full or partial dose of radiation to the eye lens. Out of 100 CT scans performed at UTMC, 83 (83%) utilized CT gantry tilt. Of those CT scans that utilized gantry tilt, 81 (97.6%) still had partial or full orbital irradiation. While gantry tilt works in theory, the clinical advantages are minimal, leading to the need for another form of eye lens dose reduction.

Dose modulation, or tube current modulation, is another form of dose reduction, however, is not being utilized at UTMC for head scans. Dose modulation reduces the tube current based on the size, shape, and density of the patients as determined by the Hounsfield units of the scout image. With the information from the scout images, the user must specify image quality values, such as the standard deviation (a measure of image noise) and max mA, in order for the tube current (mA) modulation to be calculated. In addition to tube current modulation, organ-based tube current modulation is a way to
further reduce the dose. CT manufactures have developed their own form of organ-baser tube current modulation; Organ Exposure Modulation - OEM (Toshiba Medical Systems), Organ Dose Modulation – ODM (GE Healthcare), X-CARE (Siemens Healthcare). Organ-based tube current modulation reduces the tube current even further to the anterior of the patient, reducing absorbed dose to radiosensitive organs in the anterior of the patient such as breast tissue and eye lens [5, 8, 12, 14, 22]. Of course, with a reduction of dose, there is also a reduction and potential degradation of image quality. According to physicists at Toshiba, OEM did not reduce breast tissue dose as much as they had hoped due to breast tissue lying outside the region of reduced mA. However, to their knowledge, no published literature used OEM to reduce dose to the lens of the eye.

**Objectives of this Study**

This thesis is intended to develop a clinically relevant technique in order to reduce dose to the lens of the eye. Dose reductions were measured by comparing the dose to the eye lens without OEM and with OEM. This was done by placing dosimeters on the eye lenses of an anthropomorphic RANDO phantom. A clinically-used CT scanner at UTMC was used to perform clinical scans of the RANDO phantom in order to determine the proper OEM technique. Percent dose reduction of the OEM technique was calculated and image quality was evaluated by scanning the ACR CT Accreditation phantom.
Chapter Two

Computed Tomography

History

The first clinically-used CT scanner, the EMI Mark 1, was developed in 1972 by Hounsfield and Ambrose [7, 19]. The EMI scanner was only used for head scans with an 80 x 80 array of pixels. Those pixels had a width of 2.4 mm and were only depicted in 8 shade of gray. CT scanners are classified into generations.

![Figure 1. First Generation CT](image)

First generations CT scanners were known as translate-rotate scanners. In other words, the x-ray tube was translated across the field of view (FOV) and then rotated 1 degree until 180 degrees of data were acquired. Second generation CT was similar to first generation CT only with a larger number of detectors. Also known as translate-rotate CT, second generation scanners utilized a narrow fan beam geometry instead of a pencil beam. This led to a shorter scan time and larger FOV [7].
Third generation CT (Figure 1), also known as rotate-rotate CT, is the current generation for many of the CT scanners used in clinics today. Third generation scanners utilize a rotating x-ray tube, fixed in conjunction with a rotating array of detectors. This type of scanner significantly decreased the acquisition time (on the order of a couple of seconds) [7].

Finally, fourth generation CT (Figure 2) utilizes a rotate-stationary scanner. In fourth generation CT, the x-ray tube rotates about a 360 degree array of stationary detectors. Fourth generation scanners were developed with the hope to reduce ring artifacts as seen in many third generation CT scanners. However, with the improvements in third generation CT algorithms and artifact reduction algorithms, as well as the increased cost of a fourth generation CT scanner, fourth generation CT has not taken the place of third generation CT [7].
Throughout the last couple of years, advancements in the mode of CT acquisition have been made. Early CT scanners used an axial/sequential mode of acquisition. In an axial/sequential scan, a step-and-shoot method is used. This means that with the table stationary, a scan is made with a single rotation. The table is then moved slightly and another scan is made. This process is repeated until the entire scan range has been covered. This type of acquisition could lead to longer scan times due to the x-ray not being turned on all the time. In an attempt to reduce the scan time, helical or spiral acquisition was introduced. In helical scans, the x-ray beam is on for the entire scan range. The table moves through the beam at a predetermined speed also known as the pitch. Pitch is defined as:

\[
pitch = \frac{F_{table}}{nT}
\]

Equation 1. Definition of the Pitch Factor [7]

where \(F_{table}\) is the table movement per x-ray tube rotation and \(nT\) is the collimated width of the x-ray beam. The determination of pitch also effects the acquisition time, however, acquisition time must be weighed with the desired absorbed dose and image quality.

Typical clinical CT scans are made with a tube voltage of 120kVp with a tube current (mA) varying depending on the procedure. However, many techniques used clinically today also may be performed at 80, 100, or 135kVp. Tube voltage determines the penetrability of the beam through the patient, whereas the tube current determines the number of photons in the beam. An increase in kVp will increase exposure of the beam by approximately \(kVp^2\). An increase in tube current will increase the exposure linearly. The proper technique must be determine to avoid over/underexposure of the patient, potentially leading to the need for a second scan and an unnecessary dose of radiation.
X-ray Production

X-ray tubes used in CT are similar in construction to those used in radiography or fluoroscopy; however, CT x-ray tubes are much more powerful (5-7 megaJoule compared to 0.3-0.5 megaJoule) [7]. Production of x-ray is no different though. Diagnostic x-rays are produced in one of two ways, bremsstrahlung and characteristic x-ray.

In bremsstrahlung radiation (Figure 4), an electron passes nearby an atom. The coulombic forces attract the electron towards the atom, decelerating the electron. This deceleration causes a loss of kinetic energy. This kinetic energy is released in the form of an x-ray photon with energy equal to the loss in kinetic energy. X-ray production by bremsstrahlung has an efficiency given by:

\[
\text{Efficiency} = 9 \times 10^{-10} Z V
\]

Equation 2. Bremsstrahlung Efficiency [7]

Where Z is the atomic number of the target material and V is the tube voltage in volts. So for a typical x-ray tube consisting of a tungsten target (Z = 74) operating at 120 kVp, the efficiency of bremsstrahlung production is 0.00799 or ~0.8% meaning that the remaining ~99.2% of energy is released in the form of heat.
The second form of x-ray production comes from characteristic x-rays (Figure 5). In characteristic x-ray production the incident electron interacts with the orbital electron shells of the atom. The incident electron ejects an orbital electron from its shell, leaving a vacancy. This vacancy is filled with an electron from an outer shell electron. The transition of the outer shell electron to the inner shell is accompanied by the emission of a characteristic x-ray photon. The electron shells are designated as K, L, M, etc. with the innermost shell being the K shell. The electron binding energies are unique to each individual shell. Therefore, when an electron collides with an orbital electron the resulting x-ray is characteristic to that electron from that shell of that atom. However, the incident electron must have energy greater than the binding energy of the electron it collides with to form a characteristic x-ray. If this is the case the characteristic x-ray produced will have energy equal to:

\[ h\nu = E_{B1} - E_{B2} \]

Equation 3. Energy of Characteristic x-ray

where \( E_{B1} \) is the electron binding energy of the vacant shell and \( E_{B2} \) is the binding energy of the donor shell.

The gantry of the CT scanner houses the components necessary for x-ray production and detection. Modern CT scanners utilize a slip ring technology. In slip ring
scanners, cables between the stationary and rotating components are eliminated, removing the limitations of cable length. The rotating components are connected to the slip ring by numerous gliding contacts that are in constant contact with metal bands on the stationary component of the scanner [7].

![Diagram of x-ray tube](image)

Figure 6. Diagram of x-ray tube [7]

Many components must all work in unison for a CT scanner to function properly. The x-ray tube is one of the main components of a CT scanner. Within an x-ray tube, electrons are emitted from the cathode by the process of thermionic emission. The cathode consists of a filament(s) and a focusing cup. The filament, typically made of tungsten, is connected to the power supply. As current runs through the filament, the tungsten heats up and releases electrons at a rate proportional to the current. The x-ray output is linearly proportional to the tube current. Electrons from the cathode are focused on the anode. The anode, also typically made of tungsten, must be able to withstand the high amounts of heat given off during x-ray production. As stated above, only a small fraction of electrons which collide with the anode produce x-ray (~1%). The remaining ~99% of the energy is given off as heat. A rotating anode is typically used to help spread out some of the heat over a larger area of the anode, effectively reducing heat in any given point in the anode. The cathode/anode assembly of a CT x-ray tube is oriented parallel to the z-axis of the scanner. A collimator is used to define the beam width. Collimation width is given by
\[ w = nT, \] where \( n \) is the detector array being used and \( T \) is the thickness of the detector [7]. For example, a scanner utilizing a 320 detector array with detectors of 0.5mm thick, the beam collimation will be 160mm or 16cm of scan coverage. One of the final components used during x-ray production is the bowtie filter. The bowtie filter is a beam shaping filter used to reduce the peripheral intensity of the x-ray beam. This is due to the general decrease in attenuation near the edges of the beam, giving a near uniform x-ray fluence to the detectors. Bowtie filters come in three sizes and modern scanners have multiple filters that can be interchanged based on the size of the patient being scanned. The absence of a bowtie filter or use of the wrong size filter can lead to a decrease in image noise in the periphery of the image but can contribute to unnecessary over dosage of the patient [7].

**Detection of X-ray**

![Figure 7. Multidetector CT](image)

Most modern scanners are known as multidetector CT (MDCT) scanners. Multidetector scanners utilize an array of scintillating solid-state detectors typically composed of sintered crystals. The process of sintering creates a ceramic phosphor, which is laser cut into individual detector elements. The space between each individual detector is filled with an opaque filler with the intention to limit the amount of “cross-talk” between adjacent detectors. In order to detect the x-ray incident on a detector, each detector is coupled to a photodiode. Detector arrays for some scanners are engineered to
64x64 detectors with some being manufactured even larger. The detector/photodiode arrays are connected to an electronics module. This electronics module provides the power to the detectors as well as receives the electronic signal used to form an image. Detectors can be arranged in one of two ways; a single array or, more commonly, multiple detectors. Single array detectors have all but been eliminated due to the significant decrease in scan time capable with multiple detector systems. In a multiple detector system, slice thickness is determined by the detector thickness. As mentioned before, the width of the beam is determined by the collimation of the beam.

**Image Reconstruction**

X-rays produced and transmitted through the patient are detected by the detectors within the gantry of the scanner. However, images don’t just appear, they must be reconstructed from the data acquired by the detectors. There are multiple reconstruction algorithms, each with their own pros and cons. A brief description of those reconstruction algorithms are as followed.

**Simple Backprojection**

![Figure 8. Simple Backprojection](image.png)

Forward projection takes multiple projections through an object and sums the voxel values for each projection. Simple back projection is the reverse process of forward projection. The image matrix is computed from the projection values. Figure 8 above
shows the concept of backprojection. Each projection is backprojected in the matrix and CT numbers are calculated. With the calculation of CT numbers, the image can be formed. However, in a simple backprojection image, a characteristic 1/r blur will be visible (the star pattern shadow seen in Figure 8). This blur is corrected by applying a mathematical filter, known as filtered backprojection.

**Filtered Backprojection**

Filtered backprojection is a similar process to simple backprojection with the correction for the characteristic 1/r blur. The blur is corrected using image processing. Convolution is one of the processing procedures used to correct for the blur. Convolution is known as a shift, multiply, add procedure used to smoothen a set of values. Below is an example of a 1D row of values with a convolution kernel applied.

![Figure 9. Convolution](image)

The values seen in column H are less smooth than those with the convolution applied in column G. This is due to the fact that the convolution kernel is designed to smooth the raw data. The deconvolution kernel is the exact opposite of the convolution kernel and is designed to eliminate the 1/r blur of simple backprojection. The name filtered backprojection comes from the mathematical filter of the convolution process [7].

The deconvolution process is mathematically simpler when performed in the frequency domain with the use of a Fourier transformation. A Fourier transform converts
the spatial signal detected by the detectors and converts them into the frequency domain. This is done by converting the spatial signal into a series of sine waves. When summed together, the sign waves nearly replicate the spatial signal. When the $1/r$ blurring effect is corrected in the frequency domain, the $1/r$ effect becomes a $1/f$ effect. This correction of the $1/f$ is done with the use of a ramp filter. A ramp filter alone will increase noise levels, but a ramp filter with a roll-off function will eliminate the high frequency component of the image, leading to lower noise levels. There are a number of roll-off functions, each with a different application. The inverse Fourier transform converts the signal from the frequency domain back to the spatial domain. Once back in the spatial domain, the data is backprojected to obtain an image.

**Iterative-based Reconstruction**

In past years, iterative reconstruction was seldom used in CT due to the numerical intensive algorithm. However, with recent advances in computing speed, iterative reconstruction has been gaining popularity in CT image reconstruction [7]. Iterative reconstruction algorithms can produce images with equivalent signal-to-noise ratio (SNR) with a lower dose to the patient. Iterative reconstruction, as the name suggests, is a series of iterations or updates made to the initial image. The first image is low quality and usually unrecognizable. The iterative reconstruction algorithm updates the image a number of times, by applying a forward projection until an accurate image is achieved. Between each update, the forward projection is compared to the measured data that was acquired by the detectors. This difference between projections becomes the error matrix. The error matrix is used to reduce the next iteration. This is done by creating a forward
projection with a reduced error matrix. When the error matrix is reduced further, the image becomes a near perfect depiction of the patient.

**Radiation Dose in CT**

Radiation dose in computed tomography cannot explicitly be calculated; however there has been a way to estimate the dose from a CT scan. The computed tomography dose index (CTDI) was developed initially as an index to estimate the dose from a CT scan. CTDI\textsubscript{100} uses measurements obtained from 100mm pencil chamber inserted into a polymethylmethacrylate (PMMA) phantom. CTDI\textsubscript{100} is given by:

\[
CTDI_{100} = \frac{1}{nT} \int_{L=-50mm}^{50mm} D(z)dz
\]

Equation 4. Definition of CTDI\textsubscript{100} \[7\]

Where \(nT\) is the x-ray beam width as shown on the scanner. Using a single rotation about the phantom, measurements are made with the chamber in the center and peripheral locations of the phantom. A combination of the center measurements and peripheral measurements along with a weight distribution, leads to a weighted CTDI:

\[
CTDI_W = \frac{1}{3} CTDI_{100,center} + \frac{2}{3} CTDI_{100,periphery}
\]

Equation 5. Definition of the Weighted CTDI \[7\]

The weighted CTDI gives an estimate of the average dose to the PMMA phantom along the central slice parallel to the \(z\)-axis of the scanner. In most, if not all cases, helical mode is used as the primary mode of acquisition in CT. Helical scanning is used not only to reduce the acquisition time, but also to adjust the dose from the CT scan. This is achieved by adjusting the pitch of the scan. Pitch is defined as the table movement per rotation of the CT gantry,

\[
Pitch = \frac{F_{table}}{nT}
\]
Where $F_{table}$ is the table movement (in mm) per gantry rotation and $nT$ is the collimated x-ray beam width (in mm). Dose from a CT scan in helical mode is inversely proportional to pitch [7]:

$$dose \propto \frac{1}{pitch}$$

Due to this relationship of dose to pitch, CTDIw is thus converted to CTDI$_{vol}$, the volume CTDI.

$$CTDI_{vol} = \frac{CTDI_w}{pitch}$$

Equation 6. Definition of CTDI$_{vol}$ [7]

CTDI$_{vol}$ can be calculated based on measurements from either a head phantom (16cm diameter) or abdomen phantom (32cm diameter). ACR Accreditation provides a limit on what a CTDI$_{vol}$ must be, and those values can be seen in Table 1 below:

<table>
<thead>
<tr>
<th></th>
<th>Pass/ Fail Limit</th>
<th>Reference Level</th>
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<tbody>
<tr>
<td></td>
<td>$CTDI_{vol}$ (mGy)</td>
<td>$CTDI_{vol}$ (mGy)</td>
</tr>
<tr>
<td>Adult Head</td>
<td>80</td>
<td>75</td>
</tr>
<tr>
<td>Adult Abdomen</td>
<td>30</td>
<td>25</td>
</tr>
<tr>
<td>Pediatric Head</td>
<td>40</td>
<td>35</td>
</tr>
<tr>
<td>Pediatric Abdomen</td>
<td>20</td>
<td>15</td>
</tr>
</tbody>
</table>

Table 1. ACR CTDI$_{vol}$ limits [1]

Pass/Fail limits are the limits set forth by the ACR for Accreditation in CT. Reference levels are intended to be used as a reference to identify abnormal CTDI$_{vol}$ values. Most scanners display CTDI$_{vol}$ values before the CT scan is acquired. Once again, CTDI is limited to being a dose index. It cannot be relied upon to give exact dose measurements to a patient.

**Dose Reduction Techniques**

Most CT manufactures have developed techniques that when applied, reduce dose to the patient. The most basic of these dose reduction techniques is gantry tilt. As
mentioned before, gantry tilt involves the physical tilting of the CT gantry. In most cases, the gantry can be tilted up to ±30 degrees. As CT scanners continue to grow larger, tilt angles become smaller due to the length of the CT bore in the z-axis. Gantry tilt is typically used in head scans to avoid the scanning of the orbitals.

Figure 10. Gantry Tilt

Dose modulation involves the modulation of the tube current based on the Hounsfield units acquired in the scout image. After the scout image is acquired, the computer calculates the modulation of the tube current based on the size, shape and density of the patient. Thinner and/or less dense tissue receives lower mA than thicker and/or denser tissue during the diagnostic CT scan as seen in Figures 11 and 12 below:

Figure 11. SureExposure 3D
Organ-based tube current modulation is a further advance of dose modulation. Organ-based tube current modulation reduces the mA even further in the anterior of the patient. Studies have shown that dose to the anterior of the patient can be reduced by 16-40% [8, 12, 14, 22]. These results were seen on scans of an anthropomorphic phantom in head and thoracic CT scans. Organ-based tube current modulation is intended to reduce dose to radiosensitive organs such as the eye lens, thyroid, and breast tissue.

The final form of dose reduction in CT is iterative-based image reconstruction. As mentioned before, iterative-based image reconstruction provides the same image quality with lower doses to the patient. When used with other forms of dose reduction, dose can further be reduced.

**Toshiba Aquilion One Scanner**

![Toshiba Aquilion One Scanner](image13)
The Aquilion One scanner, developed by Toshiba Medical Systems, is a state of the art scanner that is capable of up to 640 slices for rotation. For each rotation, 320 slices of data are acquired and reconstructed to 640 slices. The scanner utilizes a 0.5mm x 320 detector array that is capable of scanning up to 16cm of patient coverage per rotation. Rotation times as low as 0.275 seconds reduce the acquisition time and allow for fewer motion artifacts that can be seen with longer rotation times. Features such as SureExposure 3D, SEMAR metal artifact reduction, and Organ Exposure Modulation are also available on this scanner [15, 20].

SureExposure 3D is Toshiba’s form of tube current modulation. Tube current modulation is the reduction of tube current depending on the size, shape and density of the patient. Size, shape, and density of the patient are determined by acquiring a scout image prior to the prescribed scan. From the scout image (Figure 12 above), the computer software is able to determine a tube current modulation based on the Hounsfield units. SureExposure 3D allows users to conform to the ALARA principle by reducing dose to the patient. Tube current modulation reduces dose to the patient while maintaining adequate image quality [6].

SEMAR or single energy metal artifact reduction is artifact reduction software with the intentions to reduce artifacts cause by metal implant within the patient. SEMAR is a raw-data based iterative reconstruction technique which has been shown to reduce or remove streaking artifacts cause by metal within the patient [21].

Organ Exposure Modulation (OEM) builds upon SureExposure 3D. OEM further reduces the tube current to the anterior portion 120 degrees of the patient (Figure 14), reducing radiation dose to radiosensitive organs such as breast tissue and eye lens.
When turned on, OEM is on for the entire scan range. Therefore the entire scan range is subject to a reduce mA to the anterior side. However, tube current to the lateral and posterior of the patient is not changed, potentially leading to an increase in image noise due to a decrease in photons striking the detectors.

**CT Dosimetry of the Eye Lens**

Radiation injury to the eye lens comes in the form of radiation induced cataracts, the leading cause of blindness. A cataract is an opacification of the eye lens. Radiation injury to the eye lens is a deterministic effect. This means that the severity of the injury increases with the absorbed dose beyond a threshold dose. A threshold dose was initially determined by multiple studies that observed survivors of atomic bombs, Chernobyl and other nuclear events [10]. The threshold dose to the lens of the eye was determined to be 5Gy for acute exposures and 8Gy for fractionated exposures, as suggested by ICRP publication 103 in 2007 [10]. However, in 2011, ICRP issued a statement suggesting that the threshold of 5Gy was too high. Based on further examination of those survivors mentioned above and radiation workers, the dose threshold to the lens of the eye was lowered [9, 11]. In 2012, ICRP publication 118 lowered the eye lens threshold dose to 0.5Gy, for both acute and fractionated exposures, before the formation of radiation...
induced cataracts [11]. This lowering of the dose threshold makes dose reduction to the lens of the eye a potentially hot topic in the realm of radiology.

**CT Dosimetry of the Thyroid**

Typically, radiation injury to the thyroid is seen in radiation therapy treatments for head and neck cancers. Because of this, there is not much data from CT observations of the thyroid. Injury to the thyroid can be thyroid dysfunction, tumors, or thyroid cancer. Studies that observed atomic bomb and Chernobyl survivors as well as cancer patients revealed damage to the thyroid at a fractionated dose of greater than 18Gy [11]. Based on the data, hyperthyroidism (an overactive thyroid) occurred in a small number of people irradiated to a fractionated dose of 35Gy and greater [11]. While damage to the thyroid has been seen in atomic bomb and Chernobyl survivors, typical latency periods for radiation thyroid damage is in the range of 25-47 years [11]. This means that childhood radiation of the head and neck has a significantly higher probability of developing an injury to the thyroid. While there is data for the formation of thyroid injury, there is insufficient data to calculate a dose threshold in order to help prevent radiation injury to the thyroid, especially in head and neck CT.

Development of thyroid cancer is also a possibility when irradiating the thyroid. Thyroid cancer is a stochastic effect of radiation. This means that the probability of thyroid cancer development is proportional to absorbed dose with no threshold dose. Therefore, it is possible, yet unlikely, that a patient may develop cancer of the thyroid when irradiated with x-ray from a CT scan.
Chapter Three

Optically Stimulated Luminescent Dosimeters

Optically stimulated luminescent dosimeters (OSLD) are typically used in a clinical setting as personnel dosimeters. However, because of their ease of readability, high sensitivity, and small size make OSLD’s an increasingly popular choice for measurement of patient dose [16].

Figure 15. Landauer nanoDot - OSLD

Optically stimulated luminescent dosimeters are typically made of carbon-doped aluminum oxide (Al₂O₃:C). Dosimeters comprised of Al₂O₃:C have little to no impact on image quality due to their small size and since Al₂O₃:C is nearly a tissue equivalent material. This quality makes OSLD a popular choice for dose measurements in diagnostic radiology. When irradiated, electrons within the OSLD’s sensitive volume are elevated to a meta-stable energy state. The energy state that the electrons are elevated to is proportional to the absorbed dose to the dosimeter. OSLDs are readout on a dedicated OSLD reader as discussed below.

OSLDs are readout by a photomultiplier tube connected to a light emitting diode (LED). When placed under a beam of x-ray, the electrons in the structure of the OSLD are elevated to a meta-stable energy level. The OSLD can then be read out with the use of an OSLD reader containing the photomultiplier tube and LED. The LED stimulates the OSLD, causing the electrons to fall from their meta-stable energy state back to their original energy state. This return to the original energy state emits a characteristic light
that is recorded in the number of counted light photons. The number of counted light photons is proportional to the dose to the OSLD.

**OSLDs in CT**

OSLDs can be used in the radiotherapy environment with ease, however, recent studies have shown that OSLDs over respond significantly at the energy ranged typically used in diagnostic radiology.

![Figure 16. OSLD response with energy](image)

As one can see from Figure 16, at the diagnostic energy range, the OSLD response is significantly higher. This over response can be corrected with the use of the proper calibration factors. Dose measured by the OSLD is given by:

\[
Dose = M_{corr}(cnts) \times C_D \left(\frac{mR}{cnts}\right) \times f \left(\frac{mGy}{mR}\right) \times k_L \times k_F \times k_G \times k_\theta \times k_E
\]

Equation 7. Equation for dose with the inclusion of OSLD correction factors

where

\[
M_{corr} = M_{raw} \times k_D
\]

Correction factors applied to the raw meter reading (\(M_{raw}\)) in order to give an accurate dose are signal depletion (\(k_D\)) to give the corrected meter reading (\(M_{corr}\)). The remaining factors are the dosimeter specific calibration factor (\(C_D\)), \(f\) factor, dose linearity (\(k_L\)), signal fading (\(k_F\)), irradiation geometry (\(k_G\)), angular dependence (\(k_\theta\)), and energy dependence (\(k_E\)). A brief description of each of the calibration factors is followed.
$k_D$ - $k_D$ is a correction factor for signal depletion. With each read-out of the dosimeter, studies have shown that approximately 1.6% of the signal is depleted [17, 18]. This signal depletion correction factor is multiplied with the raw signal counts each time the dosimeters are read-out multiple times without being bleached. (Bleaching is a procedure in which the dosimeters are placed in the open position under a bright white light, releasing any remaining energy within the dosimeter) For example; the first time a dosimeter is read-out $k_D$ is 1, the second 1.016, the third 1.032 (or 1.016^2). Each time the dosimeter is bleached, $k_D$ becomes 1 again. As seen above, $M_{corr} = M_{raw} \times k_D$.

$C_D$ - $C_D$ is a dosimeter specific calibration factor that relates OSLD signal to dose. This calibration factor is also unique to the energy of the beam being used for dose measurement. $C_D$ is determined for each OSLD dosimeter by means of a calibration technique, to be discussed later in this paper.

$f$ factor – the $f$ factor is an exposure to dose conversion factor based on the medium compared to air. For this study the $f$ factor is taken to be 0.0094 mGy/mR.

$k_L$ - $k_L$ is a dose response linearity correction factor. Studies have shown that OSLDs show a linear response with dose, therefore the dose response linearity correction factor can be taken to be unity [17, 18].

$k_F$ - $k_F$ is a correction factor for signal fading. Over time, the signal within a dosimeter can fade. This is particularly high if read within 10 minutes after irradiation. However, studies have shown that if the dosimeter is read-out between 10 minutes and 2 weeks, the signal of the dosimeter will remain relatively stable. Therefore, if the dosimeter is readout within the time period mentioned above, the correction factor for signal fading can be taken to be unity [17, 18].
$k_G - k_G$ is a correction factor for irradiation geometry. When the OSLD is placed in a beam of x-ray, the dosimeter may lose signal when irradiating the sides (edges) of the dosimeter. Studies have shown that if the OSLD is used in a planar fashion, with the beam irradiating towards the edges of the dosimeter, (radiography, planar CT, etc.) a correction factor of 1.03 should be used. However, if the OSLDs are used in rotational CT the irradiation geometry correction factor can be taken to be unity [17, 18].

$k_\theta - k_\theta$ is an angular dependence correction factor. Studies have shown that a correction factor is needed if the OSLD dosimeter is placed in a manner in which the dosimeter is at an angle with respect to the x-ray tube. For example if the dosimeter is placed such that the x-ray beam is irradiating the OSLD at an angle. The same studies have also shown that if the dosimeter is placed at less than a 90 degree angle, the angular dependence correction factor can be taken to be unity [17, 18].

$k_E - k_E$ is a correction factor for energy dependence. This calibration factor is only used if the beam energy being used for dose measurement is different than the beam energy used for OSLD calibration. If the beam energy is the same the energy dependence correction factor can be taken to be unity. However, if the beam energy differs, studies have shown that the correction factor varies with energy and dosimeter position on the phantom. The same studies have shown that correction factors, relative to a 120kVp beam in air, ranged from 0.82 to 1.09 [17, 18].

OSLD calibration is needed to determine $C_D$, the dosimeter specific calibration factor. The OSLDs can be calibrated with one of three different procedures: a vendor specific calibration, a free-in-air CT calibration, or calibration using a $^{60}Co$ beam. In all three calibration protocols, the calibration factor is given by:
\[ C_D = \frac{\text{Delivered Dose (mGy)}}{\text{OSLD Signal (counts)}} \]

Equation 8. OSLD Specific Calibration Protocol [18]

The vendor specific calibration protocol is based on dosimeters being pre-irradiated by the manufacturer to known dose levels. The dosimeters are read out in counts and the calibration factor can be determined by the equation above. The free-in-air CT calibration protocol uses dose measured from a pencil ionization chamber at isocenter of the CT bore compared to the OSLD signal for the dosimeters irradiated at isocenter using the same CT protocol. The free-in-air protocol requires a calibration factor for each CT beam energy to avoid the determination of the energy dependence correction factor mentioned above. The Co-60 beam calibration protocol, as the name suggests, utilizes a Co-60 beam of 1.25MeV to deliver a known dose to the OSLDs. Exact dose to the dosimeters can be calculated based on a decay calculation of the dose rate of the source. A study [18] investigated each of these three protocols and calculated the uncertainties in OSLD signal, calibration factor, and dose. Those results are seen in Table 2 below:

<table>
<thead>
<tr>
<th></th>
<th>Uncertainty in</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>OSLD Signal</td>
<td>Calibration Factor</td>
</tr>
<tr>
<td>Vendor Specific</td>
<td>± 1.3%</td>
<td>± 5.2%</td>
</tr>
<tr>
<td>Free-in-Air</td>
<td>± 1.3%</td>
<td>± 5.3%</td>
</tr>
<tr>
<td>Co-60</td>
<td>± 1.3%</td>
<td>± 1.6%</td>
</tr>
</tbody>
</table>

Table 2. Uncertainties in OSLD Signal, Calibration Factor, and Calculated Dose [18]

The Co-60 calibration protocol showed the lowest uncertainty in calibration factor and calculated dose. However, the Co-60 protocol is typically recommended for use in the radiotherapy field rather than for diagnostic measurements. The free-in-air calibration protocol showed the next lowest uncertainty in calculated dose. This procedure however,
as mentioned before, must be carried out for each beam energy. The vendor specific calibration protocol proved to give the most uncertainty in calculated dose.

In this study, the free-in-air calibration protocol was used. This calibration protocol was chosen to avoid the determination of other calibration factors such as energy dependence. The free-in-air offered relatively equivalent dose uncertainty to the Co-60 calibration protocol without the calculation of dose delivered. Delivered dose is directly measured with the use of a pencil ionization chamber. Of the three calibration protocols, the free-in-air protocol was the most attractive protocol for this investigation.
Chapter Four

Materials and Methods

The scanner used to obtain all of the measurements in this study is a Toshiba Aquilion One Vision by Toshiba Medical Systems. The CT scan protocols used are current clinical scans of the head and neck with and without OEM. The parameters of those protocols are depicted in Tables 3 and 4 shown below. The OSLD dosimeters used to measure the dose are Landauer nanoDot dosimeter with a thickness of 0.3mm and a diameter of 4mm. The sensitive material of the dosimeter is housed within a 10 x 10mm plastic casing of thickness 1mm. The dosimeters were read-out on a Landauer InLight Microstar OSL reader. The phantoms used for measurements are a RANDO phantom for dose measurements and the CT ACR Accreditation phantom for image quality analysis. The RANDO phantom is an anthropomorphic phantom with different density materials to simulate anatomy and attenuation/scatter properties of an actual patient.

<table>
<thead>
<tr>
<th>Current Protocols</th>
<th>kVp</th>
<th>mA</th>
<th>min mA</th>
<th>max mA</th>
<th>SD</th>
<th>Rot. Time</th>
<th>Bowtie</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head (BRAIN)</td>
<td>120</td>
<td>Fixed 180</td>
<td>n/a</td>
<td>n/a</td>
<td>n/a</td>
<td>0.75s</td>
<td>Small</td>
</tr>
<tr>
<td>Head &amp; Neck CTA</td>
<td>100</td>
<td>Modulated</td>
<td>100</td>
<td>680</td>
<td>10.00</td>
<td>0.4s</td>
<td>Small</td>
</tr>
</tbody>
</table>

Table 3. Clinically-used Protocols

<table>
<thead>
<tr>
<th>OEM Protocols</th>
<th>kVp</th>
<th>mA</th>
<th>min mA</th>
<th>max mA</th>
<th>SD</th>
<th>Rot. Time</th>
<th>Bowtie</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head (BRAIN)</td>
<td>120</td>
<td>Modulated</td>
<td>50</td>
<td>200</td>
<td>2.40</td>
<td>0.75s</td>
<td>Small</td>
</tr>
<tr>
<td>Head &amp; Neck CTA</td>
<td>100</td>
<td>Modulated</td>
<td>100</td>
<td>680</td>
<td>7.50</td>
<td>0.4s</td>
<td>Small</td>
</tr>
</tbody>
</table>

Table 4. OEM Protocol
**OSLD Calibration**

The calibration procedure used in this study is the free-in-air CT calibration. The calibration was performed at both 100 and 120kVp to avoid the need for an energy dependence correction factor. The free-in-air CT calibration protocol relates exposure measured with an ionization chamber to OSLD signal using identical scan parameters and beam energy of the CT scanner. The calibration measurements were performed at 200mA using a single rotation volume scan of 0.5 seconds, not allowing any table movement. The scan was performed with the medium bowtie filter and a 0.5mm x 240 detector configuration giving a scan range of 12cm. Three nanoDot dosimeters were placed at the isocenter of the scanner via a line of tape. The tape was positioned through isocenter by alignment with the laser system of the scanner.

![Figure 17. OSLD Calibration - OSLD](image)

The nanoDots were scanned using the parameters mentioned above. The same procedure was performed for the remaining three nanoDot dosimeters. Each of the nanoDots was read-out on the Microstar reader, giving a signal in number of counts. This procedure was repeated a total of three times such that each nanoDot was irradiated and read out a total of three times. An average, depletion corrected, signal was calculated.

In order to determine a calibration factor, the OSLD signal must be compared to a measured dose from an ionization chamber under the same conditions. The ion chamber
used was a CT pencil beam chamber coupled to a RaySafe X2 meter. The chamber was suspended at isocenter using the laser light system.

![Figure 18. OSLD Calibration – Ion Chamber](image)

Using the same scan parameters above, a total of four exposures were taken in unit of Roentgen (mR) and an average exposure was calculated. The dosimeter specific calibration factor is given by:

\[
C_D = \frac{\text{Delivered Dose (mR)}}{\text{OSLD Signal (counts)}}
\]

**Dose Measurement**

As mentioned above, dose measurements were performed using clinically-used protocols on a RANDO phantom. The 120kVp scans were performed with the clinical brain protocol, and the 100kVp scans were performed with the carotid head and neck CTA protocol. The phantom was placed on the couch of the CT scanner and aligned using the laser light system. With the phantom aligned, a scout image was performed without the nanoDot dosimeters.
After the scout image was obtained, the nanoDots were placed on the phantom (Figure 20 below) at the following positions for the brain protocol: left eye, right eye, left side of head, right side of head, and the posterior of the head to verify the functionality of the organ exposure modulation feature. The nanoDots were placed on the phantom for the head and neck CTA protocol in the following positions: left eye, right eye, and the thyroid. The thyroid dosimeter placement was determined by examining CT images and determining an average depth of the thyroid. The depth of the thyroid used in this study was taken to be 1cm. This depth was achieved by the use of a tissue equivalent bolus of 1cm placed over the dosimeter on the phantom.
With the dosimeters in place, a clinical scan was made with the current clinically-used protocol. The current clinically-used brain (head) protocol is shown in Table 3 above. Once the scan was completed, the dosimeters were removed and read-out on the Microstar reader. The exposure to each location of the phantom, leaving out factors of unity, is given by:

\[ \text{Exposure (mR)} = M_{\text{raw}} \times k_D \times C_D \]

Equation 9. Calculation of exposure from OSLD

The dose to each location of the phantom, leaving out factors of unity, is given by:

\[ \text{Dose (mGy)} = \text{Exposure(mR)} \times f \]

Equation 10. Calculation of Dose from the exposure of an OSLD

Where a \( f \)-factor for tissue to air of 0.0094 mGy/mR was used to convert exposure to dose. This procedure was repeated three times and an average dose to each location of the phantom was calculated. This set of dose measurements was for a clinical scan without OEM. The same procedure was repeated using the same scan protocol, this time, with OEM turned ON. The OEM brain protocol is shown in Table 4 above, with a maximum mA of 200 and a minimum mA of 50. This protocol was determined by scanning the RANDO phantom a number of times with the current protocol and several slightly different OEM protocols. Noise was measured in a region of interest (ROI) at the same location (ie. same slice) on each set of images. The protocol was determined once the noise levels in the ROI of the OEM scan were relatively equal to the noise levels in the ROI of the scan with the current protocol. An average dose to each location with OEM on was calculated and a percent dose difference was calculated by:

\[ \% \text{Dose Savings} = \frac{\text{Current Dose} - \text{OEM Dose}}{\text{Current Dose}} \]

Equation 11. Calculation of Dose Savings with the use of OEM
The same procedure above was repeated for the head and neck CTA protocol at 100kVp. The current head and neck CTA protocol utilizes a 100kVp beam, 0.75s rotation time, small bowtie filter, and xy-dose modulation as seen in Table 3. The OEM protocol for the head and neck CTA protocol consists of the same techniques but this time with OEM turned on.

**Image Quality Measurement**

Since this study is based on the reduction of dose, image quality may be affected. With a decrease in dose, there is also an increase in image noise and the potential for image quality degradation. To evaluate image quality, scans of the ACR Accreditation CT phantom were made. The ACR phantom is a cylindrical phantom of 16cm length and a diameter of 20cm. The phantom consists of four modules. The first module measures CT number accuracy and slice thickness. This module contains five targets of various densities resembling water, air, bone, acrylic, and polyethylene. Module 2 measures low contrast resolution and image noise, and contains six different sized targets to assess low contrast. Module 3 assesses CT number uniformity and module 4 measures spatial resolution. Module 4 contains eight different high contrast line pair patterns [2].

![Figure 21. ACR CT Phantom](image)

[2]
The image quality scans used the same protocols as the dose measurement scans. To compare image quality, a scan using the current clinical protocol was made followed by a scan with the OEM protocol for both of the protocols listed above (Tables 3 and 4). From the images obtained, contrast-to-noise ratio and image noise was measured. Those measurements were performed in the images as depicted in Figure 22 below:

![Figure 22. CNR Measurements (left) and Noise Measurement (right)](image)

Image noise was measured in module 3 of the phantom. Small diameter ROIs were drawn at the top, bottom, left, right and center of the image as shown in Figure 22 above. Additionally, an ROI was drawn over the entire image. The noise taken as the standard deviation for each ROI was recorded. Contrast-to-noise ratio (CNR) was measured based on the procedure set forth in the ACR CT Accreditation manual. CNR is measured in module 2 and is calculated as followed:

\[
\text{CNR} = \frac{\text{target mean} - \text{background mean}}{\text{background std. dev.}}
\]


ACR requires that the CNR, for adult head scans, be a 1.0 or greater for accreditation in CT. Noise and CNR measurements for the current protocol were compared with the measurements from the OEM protocol with percent differences calculated. The noise
measurements and CNR measurements were performed using ImageJ (National Institute of Health), an image analysis software.
Chapter Five

Results

Dose to the lens of the eyes for the current clinically-used brain protocol was measured to be on average 42.0±4.9 mGy. The dose measured for the OEM brain protocol was 29.5±4.6 mGy, a dose reduction of 29.9%. Radiation dose to the sides and posterior of the head was reduced on average by 10.2±3.5%. These results can be easily seen in Figure 23 below:

However, as mentioned before, with the reduction in dose, there is also the increase in image noise and the potential for image quality degradation. Contrast-to-noise ratio was measured to be 1.822 for the current brain protocol and 1.547 for the OEM protocol, a reduction of 15.1%. Image noise was increased in the anterior of the image by 14.9%. Image noise was increased in the remainder of the image by an average of 4.1%. Noise throughout the entire image was measured to have increased by 8.3%. Tables 5 and 6 (below) show the image quality measurements for the head protocol.
<table>
<thead>
<tr>
<th></th>
<th>Current Protocol</th>
<th>OEM Protocol</th>
<th>% Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anterior</td>
<td>3.270</td>
<td>3.757</td>
<td>14.9%</td>
</tr>
<tr>
<td>Left</td>
<td>3.093</td>
<td>3.234</td>
<td>4.6%</td>
</tr>
<tr>
<td>Posterior</td>
<td>3.602</td>
<td>3.797</td>
<td>5.4%</td>
</tr>
<tr>
<td>Right</td>
<td>3.151</td>
<td>3.262</td>
<td>3.5%</td>
</tr>
<tr>
<td>Center</td>
<td>3.967</td>
<td>4.081</td>
<td>2.9%</td>
</tr>
<tr>
<td>Entire Image</td>
<td>6.330</td>
<td>6.854</td>
<td>8.3%</td>
</tr>
</tbody>
</table>

Table 5. Image Noise Measurements – Head

<table>
<thead>
<tr>
<th></th>
<th>Current Protocol</th>
<th>OEM Protocol</th>
<th>% Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Signal</td>
<td>90.166</td>
<td>90.148</td>
<td></td>
</tr>
<tr>
<td>Background</td>
<td>83.578</td>
<td>83.703</td>
<td></td>
</tr>
<tr>
<td>SD (Noise)</td>
<td>3.616</td>
<td>4.165</td>
<td></td>
</tr>
<tr>
<td>CNR</td>
<td>1.822</td>
<td>1.547</td>
<td>-15.1%</td>
</tr>
</tbody>
</table>

Table 6. CNR Measurements - Head

In addition to the lens of the eyes, the thyroid was also measured during the scans of the head and neck. The dose to the lens of the eyes for the current clinically-used protocol was measured to be 59.1±0.84 mGy and dose to the thyroid was measured to be 17.38±1.2 mGy. The dose measured for the head and neck protocol with OEM for the eye lens was measured to be 45.0±0.69 mGy, a dose reduction of 23.8%. Dose to the thyroid for the OEM head and neck protocol was measured to be 21.8±0.9 mGy, an increase of 25.6%. Figure 24 (below) shows the measurements for the head and neck CTA protocol.
Contrast-to-noise ratio was measured to be 1.123 for the current clinically-used head and neck protocol. The CNR was measured to be 0.922 for the OEM head and neck protocol, a reduction of 17.9%. Image noise was increased by 16.6% in the anterior of the image. Image noise was also increased in the remaining locations of the image by an average of 17.0%. Once again, noise throughout the entire image was increased. This set of image showed an increase in noise of 6.9% for the entire image. Tables 7 and 8 (below) show the image quality results for the head and neck CTA protocol.

<table>
<thead>
<tr>
<th>OSLD Location</th>
<th>Current Protocol</th>
<th>OEM Protocol</th>
<th>% Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anterior</td>
<td>2.478</td>
<td>2.889</td>
<td>16.6%</td>
</tr>
<tr>
<td>Left</td>
<td>2.692</td>
<td>2.874</td>
<td>6.8%</td>
</tr>
<tr>
<td>Posterior</td>
<td>2.554</td>
<td>3.146</td>
<td>23.2%</td>
</tr>
<tr>
<td>Right</td>
<td>2.577</td>
<td>2.987</td>
<td>15.9%</td>
</tr>
<tr>
<td>Center</td>
<td>2.836</td>
<td>3.456</td>
<td>21.9%</td>
</tr>
<tr>
<td>Entire Image</td>
<td>4.082</td>
<td>4.362</td>
<td>6.9%</td>
</tr>
</tbody>
</table>

Table 7. Image Noise Measurements – Head and Neck
<table>
<thead>
<tr>
<th></th>
<th>Current Protocol</th>
<th>OEM Protocol</th>
<th>% Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Signal</td>
<td>137.304</td>
<td>137.187</td>
<td></td>
</tr>
<tr>
<td>Background</td>
<td>133.341</td>
<td>133.523</td>
<td></td>
</tr>
<tr>
<td>SD (Noise)</td>
<td>3.528</td>
<td>3.973</td>
<td></td>
</tr>
<tr>
<td>CNR</td>
<td>1.123</td>
<td>0.922</td>
<td>-17.9%</td>
</tr>
</tbody>
</table>

Table 8. CNR Measurements – Head and Neck
Chapter Six

Discussion

This study was intended to investigate the use of organ exposure modulation to reduce the radiation dose to radiosensitive organs in head and neck CT. The study compared radiation dose and image quality for current clinically-used protocols and OEM protocols. The protocols investigated in this study were the adult brain protocol and the adult head and neck CTA protocol. Radiation dose reductions were seen in both eye lenses for the OEM brain scan, with an average dose savings of 29.9%. Dose reductions were seen in both eye lenses for the OEM head and neck scan, with an average dose savings of 23.8%. With the current dose threshold of 0.5Gy [11] and an average lens dose in head scans at UTMC of 50mGy, as determined from Radimetrics, the number of scans a patient may receive before cataract formation is 10 scans. However, with a lens dose reduction to 29.5mGy, the number of CT scans a patient may receive before cataract formation is increased up to 17 scans. However, with the use of OEM on the head and neck protocol, the dose to the thyroid is increased on average by 25.6%. It is unknown what caused the dose to the thyroid to increase, as it was expected to decrease similar to the eye lens. This increase in dose to the thyroid was however, reproducible over three different trials. Radiologists would need to decide whether the benefits outweigh the dose increase to the thyroid if OEM were to be used in the head and neck protocol.

The use of OEM reduced dose to anterior projections without compensating for the dose reduction in the posterior or lateral projections by increasing dose to those projections. In a reconstructed image, this reduction results in an increase in image noise dependent upon the decrease in the effective mA of the OEM protocol. Image quality was
measured and compared, for each protocol with and without OEM, by scanning the ACR CT Accreditation phantom. Image noise was measured in four peripheral locations, the center and the entire image, and CNR was measured according to the procedure set forth in the ACR CT Accreditation Manual. Image noise for the OEM brain protocol was increased in the anterior by 14.9% with an overall increase of 8.3%. The CNR was reduced by 15.1% from 1.822 to 1.547. Image noise for the OEM head and neck protocol was increased 16.6% in the anterior, and an overall increase of 6.9%. The CNR was also reduced by 17.9% from 1.123 to 0.922. ACR Accreditation requires a minimum CNR of 1.0 for adult head protocols. Therefore the Head and Neck protocol would not meet ACR Accreditation requirements and therefore should be investigated further.

Current dose reduction techniques (gantry tilt) in head scans have proven to be relatively ineffective at reducing dose to the eye lens. At UTMC approximately 98% of those CT scans that utilize gantry tilt still contain the orbits in the image. The 2% of scans that the orbits are not visible in the image, does not mean that the orbit do not receive any radiation. A scan of a CR plate was made with metal BB’s marking the scan range. Analysis of the CR plate showed that due to the helical nature of the CT scan, the x-ray field was seen to extend nearly 3cm on either end of the scan range. This means that even if the orbits are just outside the scan range, part or all of the orbitals still receive a partial or full dose of radiation. Scans of the RANDO phantom have shown that OEM is a viable option for reducing dose the lens of the eyes in head (specifically brain) scans. Image quality was clearly reduced as noise increased and CNR decreased. However, even with a decrease in CNR, to 1.547, CNR remained above the minimum limit of 1.0 set forth by the ACR.
SureExposure 3D dose modulation is the current form of dose reduction in the head and neck protocol. SureExposure 3D modulates the mA based on the size, shape, and density of the patient based on data acquired from the scout image [6]. With the use of OEM, dose can further be reduced to the anterior of the patient. As seen in the head scan, image noise increased throughout the image, as expected, and CNR decreased in the head and neck protocol. The CNR was reduced from 1.123 to 0.922. This value for CNR would not pass the ACR Accreditation requirements of 1.0. Therefore further investigation into a proper technique must be completed if OEM were to be used for this protocol.

All-in-all, this investigation proves that OEM functioned as it was intended and reduced dose to the anterior of the “patient” without increasing dose to the posterior. The OEM brain protocol showed the most promising results with a dose reduction to both eye lenses and adequate image quality that passes minimum ACR Accreditation requirements. The OEM head and neck protocol showed a dose reduction in the eye lenses as well. However, dose to the thyroid was significantly increased and image quality dose not pass the minimum ACR Accreditation requirements. If the OEM head and neck protocol were to be used in the clinical setting, dose to the thyroid would need to be monitored. In addition, further investigation into a proper technique would need to be completed to increase the CNR above the minimum value of 1.0 set forth by the ACR for accreditation.

Dose reduction in head and neck CT is more vital in the present time due to the lowering of the dose threshold for cataract formation. Cataract formation is a deterministic effect, meaning severity of the injury increases with dose beyond a
threshold dose. In 2007, ICRP publication 103 suggested a dose threshold dose of 5Gy and 8Gy for acute and fractionated exposures, respectively [10]. However, in 2012, ICRP publication 118 lowered that dose threshold to 0.5Gy for acute and fractionated exposures [11]. The use of organ exposure modulation reduced the dose to the lens of the eye, hence increasing the number of scans a patient may receive before developing a radiation induced cataract.
Chapter Seven

Conclusions

1. Use of an OEM protocol reduces dose to the lens of the eye by 29.9% for the head CT protocol and by 23.8% for the head and neck CTA protocol.

2. Use of an OEM protocol caused an increase in image noise by 14.9% in the anterior and 8.3% over the entire image for the head protocol and 16.6% in the anterior and 6.9% over the entire image for the head and neck CTA protocol.

3. Use of an OEM protocol decreased contrast-to-noise ratio by 15.1% to 1.547 for the head protocol and 17.9% to 0.922 for the head and neck CTA protocol.

4. Dose to the thyroid was increased by 25.6% with the use of OEM in the head and neck CTA protocol.

5. OEM is a viable option for clinical use to reduce dose to the lens of the eye. However, for protocols that include the thyroid, the increase in dose to the thyroid may inhibit the usefulness of OEM.
References


5. Angel, E. PhD, Senior Manager, Clinical Collaborations. Organ Effective Modulation OEM. Lecture presented at Toshiba Medical Systems.


