Dosimetric Evaluation of Metal Artefact Reduction using Metal Artefact Reduction (MAR) Algorithm and Dual-energy Computed Tomography (CT) Method

by

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David Huang

Thesis submitted in partial fulfillment of the requirements for the degree of Master of Science in the Medical Physics Program Duke University and Duke Kunshan University

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ABSTRACT

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Abstract

Purpose: Computed Tomography (CT) is one of the standard diagnostic imaging modalities for the evaluation of a patient’s medical condition. In comparison to other imaging modalities such as Magnetic Resonance Imaging (MRI), CT is a fast acquisition imaging device with higher spatial resolution and higher contrast-to-noise ratio (CNR) for bony structures. CT images are presented through a gray scale of independent values in Hounsfield units (HU). High HU-valued materials represent higher density. High density materials, such as metal, tend to erroneously increase the HU values around it due to reconstruction software limitations. This problem of increased HU values due to metal presence is referred to as metal artefacts. Hip prostheses, dental fillings, aneurysm clips, and spinal clips are a few examples of metal objects that are of clinical relevance. These implants create artefacts such as beam hardening and photon starvation that distort CT images and degrade image quality. This is of great significance because the distortions may cause improper evaluation of images and inaccurate dose calculation in the treatment planning system. Different algorithms are being developed to reduce these artefacts for better image quality for both diagnostic and therapeutic purposes. However, very limited information is available about the effect of artefact correction on dose calculation accuracy. This research study evaluates the dosimetric effect of metal artefact reduction algorithms on severe artefacts on CT images. This
study uses Gemstone Spectral Imaging (GSI)-based MAR algorithm, projection-based Metal Artefact Reduction (MAR) algorithm, and the Dual-Energy method.

**Materials and Methods:** The Gemstone Spectral Imaging (GSI)-based and SMART Metal Artefact Reduction (MAR) algorithms are metal artefact reduction protocols embedded in two different CT scanner models by General Electric (GE), and the Dual-Energy Imaging Method was developed at Duke University. All three approaches were applied in this research for dosimetric evaluation on CT images with severe metal artefacts. The first part of the research used a water phantom with four iodine syringes. Two sets of plans, multi-arc plans and single-arc plans, using the Volumetric Modulated Arc therapy (VMAT) technique were designed to avoid or minimize influences from high-density objects. The second part of the research used projection-based MAR Algorithm and the Dual-Energy Method. Calculated Doses (Mean, Minimum, and Maximum Doses) to the planning treatment volume (PTV) were compared and homogeneity index (HI) calculated.

**Results:** (1) Without the GSI-based MAR application, a percent error between mean dose and the absolute dose ranging from 3.4-5.7% per fraction was observed. In contrast, the error was decreased to a range of 0.09-2.3% per fraction with the GSI-based MAR algorithm. There was a percent difference ranging from 1.7-4.2% per fraction between with and without using the GSI-based MAR algorithm. (2) A range of 0.1-3.2% difference was observed for the maximum dose values, 1.5-10.4% for minimum dose
difference, and 1.4-1.7% difference on the mean doses. Homogeneity indexes (HI) ranging from 0.068-0.065 for dual-energy method and 0.063-0.141 with projection-based MAR algorithm were also calculated.

**Conclusion:** (1) Percent error without using the GSI-based MAR algorithm may deviate as high as 5.7%. This error invalidates the goal of Radiation Therapy to provide a more precise treatment. Thus, GSI-based MAR algorithm was desirable due to its better dose calculation accuracy. (2) Based on direct numerical observation, there was no apparent deviation between the mean doses of different techniques but deviation was evident on the maximum and minimum doses. The HI for the dual-energy method almost achieved the desirable null values. In conclusion, the Dual-Energy method gave better dose calculation accuracy to the planning treatment volume (PTV) for images with metal artefacts than with or without GE MAR Algorithm.
Dedication

For my parents, Cesar and Edna Laguda, I thank them for their unwavering support and genuine love. No words can describe the sacrifices they have made for me. It is not only hard work and studying that put me where I am. I will always be deeply grateful to them for the privilege they have given me.
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List of Abbreviations

CT - Computed Tomography
MRI - Magnetic Resonance Imaging
HU - Hounsfield Units
GSI - Gemstone Spectral Imaging
MAR - Metal Artefact Reduction
GE - General Electric
VMAT - Volumetric Modulated Arc Therapy
PTV - Planning Target Volume
HI - Homogeneity Index
CSF - Cerebrospinal Fluid
MDCT - Multi-detector Computed Tomography
FBP - Filtered Backprojection
FP - Forward Projection
CR - Computed Radiography
VM - Virtual Monochromatic
RED - Relative Electron Density
ABR - American Board of Radiology
MLC - Multi-leaf Collimator
Gy  - Gray
TCP  - Tumor Control Probability
CBCT - Cone Beam Computed Tomography
MTF  - Modulation Transfer Function
OBI  - On-board Imager
PLA  - Polylactic Acid
TPS  - Treatment Planning System
Acknowledgements

I wish to thank my advisor, Dr. Fang-Fang Yin, who always encourages me to work independently and strive for excellence. Dr. Yin has taught me to think creatively and work passionately as a researcher. I am also grateful to Dr. Xiangpeng Zheng, my co-advisor, for his encouragement, guidance, and hands-on training throughout this project. Many thanks to Huadong Hospital, China, especially Jian Jian Niu, for providing training and assistance.

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1. Introduction

Radiation therapy is a common treatment given to cancer treatment other than surgery and chemotherapy. The goal of radiation therapy is to provide curative or palliative treatments precisely. To provide such treatments, better image quality of position and size of the tumor and lesions is desirable. CT images of patients with metal implants produce bad image quality due to metal artefacts. Metal artefacts change the electron density information on CT images. This change leads to incorrect treatment radiation dose calculation. Incorrect calculation is undesirable for attaining a precise radiation treatment. This is a two-part research study that looks into the dosimetric impact of three metal artefact reduction (MAR) methods. The two parts were done in different hospitals and in different countries -- China and United States of America (USA). The first part was implemented with a GSI-based algorithm in China while the second part evaluated a projection-based algorithm and the dual-energy method in the USA.

1.1 Computed Tomography (CT) and Metal Artefacts

Computed tomography (CT) is one of the standard imaging modalities due to its fast acquisition time, high spatial resolution, and ability to eliminate overlapping
structures common in conventional radiography [1]. However, like most imaging modalities, CT is disadvantaged by a patient’s motion and susceptibility to artefacts. While the former issue will not be discussed in this thesis, this research focuses on specific methods of correcting artefacts. Images produced by CT are presented in gray scales, which vary depending on energy, attenuation, and electron density of objects (e.g., soft tissues, bones) under observation. Each gray scale that corresponds to a unique value is computed in Hounsfield units:

\[
HU = 1000 \left( \frac{\mu - \mu_w}{\mu_w} \right)
\]

where \( \mu \) and \( \mu_w \) are the linear attenuation coefficients of an object under observation and water, respectively [2]. The most common HU values of medical concern are shown below. These values are not unique.

**Table 1: Common HU Values. Table from Freeman [33].**

<table>
<thead>
<tr>
<th>Substance</th>
<th>HU</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lung (air)</td>
<td>-700 to -1000</td>
</tr>
<tr>
<td>Fat</td>
<td>-100 to -50</td>
</tr>
<tr>
<td>Water</td>
<td>0</td>
</tr>
<tr>
<td>Cerebrospinal Fluid (CSF)</td>
<td>15</td>
</tr>
<tr>
<td>Soft Tissue</td>
<td>+100 to +300</td>
</tr>
<tr>
<td>Dense Bone</td>
<td>+1000 to +3000</td>
</tr>
</tbody>
</table>
The presence of metals such as dental fillings, spinal implants, surgical clips, shoulder implants, and hip prostheses mainly causes severe metal artefacts in CT. This, in turn, degrades the image quality. The severity of the artefacts depends on the size and material of the object. HU value of the object under observation is explicitly dependent on linear attenuation - the higher the attenuation, the higher the HU value.

Metal substantially attenuates photon beams, causing insufficient photon detection. Insufficient photon detection results in photon starvation and incorrect HU values surrounding the metal. Incorrect HU values misrepresent the tissue electron density information. According to Wei et al., erroneous electron density information results in incorrect dose calculations leading to undesirable radiation treatment plans [3]. Metal Artefacts are more evident in steel and gold and lesser in titanium and aluminum [4]. Even if sufficient photons are detected after passing through a highly-attenuating metal, high quantum noise is still observed, leading to streaking artefacts, photon starvation, and beam hardening [5]. Figure 1 shows an example of metal artefacts seen in a phantom.
1.2 Metal Artefacts Correction: Image Processing Approaches

Researchers have devoted their efforts to finding a solution to metal artefacts that could potentially lead to large dose deviations. The most common method applies corrections to Radon transform without scanner’s specifications. Radon transformation of the image is the sum of the radon transformation of each independent pixel. It uses interpolation to decrease signal in areas that are associated with projections of metal objects [6]. The technique results in a decrease of noise and artefacts in CT scan. Also, the technique removes bright and streak artefacts, resulting in an improved image quality. However, the images might still not be clinically acceptable. A similar technique known
as Forward Projection (FP) algorithm uses a modern method for Multi-detector Computed Tomography (MDCT) that corrects original raw projections and not the artificial raw data. Metal was identified by an upper and lower threshold of 2000 and 500 HU, respectively [2]. However, the technique was computationally inefficient (time-consuming) making it unsuitable for clinical purposes. It sometimes produces image artefacts due to incomplete and imprecise data.

Another MAR algorithm was implemented on filtered backprojection (FBP), which is generally termed as a filtering technique. It uses linear and polynomial interpolation techniques to employ “missing” projection data. However, the threshold that is commonly used leads to information lost. Kalender, Hebel, and Ebersberger [7] used linear interpolation within each projection while Roeske et al. [8] applied a cubic spline fit to correct for artefacts in images containing a Fletcher-Suit applicator.

Yazdi used a technique that automatically detected missing information in the corrupted sinogram. Neighboring projections were used at different angles to correct for metal artefacts in CT images. This technique is good for a typical hip prosthesis shape but not for irregularly shaped ones [9]. However, Bal and Spies [10] used reduced artefacts by in-painting missing information into the corrupted sinogram. The main problem lies on needing a prior tissue model. One of the most sophisticated techniques
is the segmentation-based method for MAR. Executed by Henyong Yu, et al., they used mean-shift technique and feedback strategy [11].

Prell, D. et al. [12] implemented an algorithm dedicated to metal artefact with interventional flat-panel detectors. They used cadavers to demonstrate this artefact reduction algorithm near a prosthesis and showed desirable results. Bazalova, M. et al. [13] studied the influence on Monte Carlo dose calculations where they used a “sinogram interpolation metal streaking algorithm on several phantoms of exact-known compositions on a prostate patient with two hip prostheses.” There was also a therapeutic research method that studied these kinds of situations by implementing coplanar intensity-modulated radiotherapy in prostate cancer patients with and without nodal involvement [2]. However, inhomogeneity corrections were not applied, metal artefacts were not corrected and the research only applied on one type of cancer.

With fast-growing technology, General Electric (GE) developed and is still developing novel metal artefact reduction techniques. One of their commercially available methods is known as SMART Metal Artefact Reduction method. This is a three-step process that includes identifying corrupted projections due to metal artefacts, replacing these projections with corrected ones, and comparing the two projections acquired to estimate the correct projection of the set [14].
1.3 Dual-Energy Imaging

The dual-energy imaging technique has been known since the 1970s. However, due to the unavailability of the technology in par with the knowledge at that time, it was harder to implement the technique. It is a technique that takes advantage of the relationship between the attenuation coefficient and energy. Images produced at different energies differ on which structure(s) is (are) more evident. An obvious problem seen in a CT image taken at 120-kVp, a common energy used for CT imaging, is its “crowded” information. This image shows both soft tissue and bone structure information, which may lead to inaccurate medical diagnosis. Since attenuation is inversely proportional to the third power of energy, one can produce bone-subtracted and soft-tissue subtracted images [15].

Contrasts such as iodine and barium are ideal for illustrating Dual-Energy demonstration due to their K-shell binding energies 33.2 keV and 37.4 keV, respectively. Note that these values are close to the mean energy of most diagnostic X-ray beams and have higher mass attenuation coefficients. [15].
Dual-energy imaging has been the primary technique used to eliminate any anatomical “noise” seen in radiographic images. According to Dobbins et al., more than a quarter of the nodules are overlooked on conventional radiographs not because of quantum noise and low contrast but due to poor conspicuity [16]. This method is realized in two ways. One method uses a sandwich detector of two computed radiographs (CR) with a hardening filter in the middle. The other uses two separated flat-panel detectors with two different energies (low and high). Both of these methods give high spatial resolution but are very susceptible to motion and may increase patient dose [16].

As stated above, GE has been developing techniques to reduce metal artefacts. One of these techniques uses the concept of dual-energy. The technique is known as Spectral Imaging. It generates virtual monochromatic CTs for dual-energy imaging subtraction method needed to eliminate unwanted structures as described by Alvarez et al. [17]. This method uses fast-kVp switching to acquire dual-energy imaging simultaneously. Also, a dual-energy technique developed at Duke University based on CBCT images was used in this research [28].
1.4 Computed Tomography (CT) and Cone-Beam CT

CT images are taken slice-by-slice projection image of an object while CBCT is a volumetric CT reconstruction of on-board imager (OBI) images. The differences between CBCT and CT images are the images acquisition and application. CT images are acquired using a fan beam angle geometry while CBCT is a cone beam angle geometry. Due to CBCT’s increased x-ray scatter, image quality and uniformity are decreased. According to Yoo et al., these problems misrepresents HU values in 3D reconstruction. However, CBCT-based and CT-based treatment plans are dosimetrically equivalent [34].

The relative electron density (RED) should first be determined to evaluate the dose deposited in each pixel of a CT data set. This is supported by a CT number curve that assigns the HU of each pixel to an associated RED value [35].

1.5 Objective

Although various techniques have been developed for metal artefact reduction, only limited information is available to understand the impact on the dose calculation based on CT of these correction techniques. This paper evaluates the dosimetric effect of metal artefact correction on the planning target volume (PTV) using three different metal artefact reduction (MAR) methods. These methods include GE Spectral Imaging method,
GE SMART MAR algorithm, and the Dual-energy Method. The Dual-energy computed tomography algorithm technique was developed at Duke University. For the first part, the research evaluates the dosimetric errors on the PTV between the mean doses calculated by the treatment plan and the absolute doses measured during the dose delivery. For the second part, parameters for dose evaluation are the treatment plan doses (mean dose, minimum dose, and maximum dose) and the homogeneity index (HI).
2. General Electric (GE) Spectral Imaging

2.1 Motivation

The Virtual Monochromatic (VM) technique is the most common technique used for processing dual-energy imaging method. The VM technique generates an image of an object irradiated by a single energy. A set of these images is known as monochromatic images. CT monochromatic images are processed before reconstruction to improve image quality of CT images with severe metal artefacts [18]. A new technique developed by General Electric embedded in GE Discovery CT750 HD (GE Healthcare, Milwaukee, Wisconsin) uses the concept of dual-energy with fast-kVp switching between 80 and 140 kVp. Monochromatic images created have energies between 40 and 140 keV [19].

This technique improves lesion detection obtained in comparison to conventional CT [20]. It obtains relative electron density (RED) from virtual monochromatic CT images by GSI mode with fast-kVp switching. Interpolation uses dual energy acquisition. Ogata et al. confirmed the feasibility of the treatment planning but object size is not considered and high density materials, such as metals, are not used [21]. Radiation Dose for GSI-based MAR Algorithm is below the limit of American Board of Radiology (ABR) for abdominal imaging. However, CT radiation dose of GSI-based
MAR algorithm increases to 20% higher than conventional CT for this type of imaging [22].

2.2 Materials

Materials used in this part of the research included a water phantom and four syringes. Figure 3 shows the water phantom set-up. The four syringes were filled with an iodinated contrast agent that imitates the metal component. Medical tape, two slabs of solid water, and water were also included. Medical tape was used to stick four syringes onto the slabs parallel and opposite to each other. Iodine contrast agent was chosen to take advantage of its k-edge property and because of its common presence in clinical settings. The phantom was filled with water.

![Figure 2: Set-up of the phantom. Photo taken at Huadong Hospital, China.](image-url)
2.3 Methodology

After setting up the phantom, a CT scan was taken (shown in Figure 5). Here, fiducial markers are very important. Three fiducial markers were placed, one on both sides and one on the top. The set-up of the phantom while CT images were taken is shown in Figure 4.

![Set-up of the phantom while taking its image using CT scan. Photo Taken at Huadong, Hospital, China](image)

The target (PTV) and iodinated contrast agents were first delineated on the images reconstructed with and without GSI-based MAR Algorithm shown in Figure 5. An additional four rings were drawn a few cm (0.2, 0.8, 1.2, and 2) away from the PTV. The rings served to investigate the dose gradient at the boundary of the PTV.
Figure 4: CT Image with (upper) and without (lower) GSI-based MAR. CT Image Taken at Huadong, Hospital, China
The two plans were implemented using two sets of arc treatment. The first set was a multi-arc plan and a single arc plan that avoided the metals using VMAT. The Partial arc and a full arc plan were composed of four plans of $\frac{1}{4}$ arc, $\frac{1}{2}$ arc, $\frac{3}{4}$ arc, and 1 full arc. The four plans passed through the metals while the single arc plan avoided the metals. VMAT was used since it improved treatment quality and efficiency compared to 3D-conformal RT [29]. This part of the research used Styrofoam (polystyrene) instead of solid water first but images produced were too dark and thus not considered for this research. Planning target volume (PTV) was contoured at the center of the four areas where metal artefacts were evident. Thus, the study did not consider lesions near the metals on the water phantom. OCTAVIUS 4D (PTW, Freiburg, Germany) with accompanying software VERISOFT (PTW, Freiburg, Germany) was used for Intensity Modulated Radiation Therapy Quality Assurance (IMRTQA). Octavius 4D measures the dose plane at isocenter while the phantom rotates with the gantry. These measurements were collected and reconstructed during the volumetric modulated arc therapy delivery (VMAT) into a 3D dose volume. The reconstructed dose was compared against the 3D dose matrix from the TPS but with the same geometry and beam fields using VERISOFT.
VMAT treatment plan was optimized. Optimization was implemented by optimizing the beam intensity at each beam angle. Three goals were needed to achieve this optimization. First was to attain dose conformity, then to observe a fast dose gradient fall-off and lastly, to have dose homogeneity. On the other hand, normal tissue objectives include decreasing dose levels away from the PTV, limiting dose levels, preventing hot spots outside the PTV, and obtaining a sharp dose gradient around the PTV. Beam optimization includes a structure model parameter given by

$$points = \frac{Volume}{resolution}$$

where points are the number of pixels. This parameter has a default resolution value of 4.5 mm for structures with volumes greater than 5000 cm$^3$ and structures with volumes less than 5000 cm$^3$ to 3 mm$^3$ while the minimum number of points is 2000. Since there is a limiting value, there is evidently a tradeoff. A resolution as small as possible is desirable. However, making the resolution too small reaches the minimum required number of points. On the other hand, a large number of points slow down the optimization time since it covers a lot of data memory. An alternative is to increase the resolution of organs, thus making their points decrease. However, therapeutic ratio should be considered. Another parameter is the upper and lower objectives. The upper objective restricts the dose and the lower objective is the desired dose level. The time
limits and iterations are set at 100 minutes and 1000, respectively. Calculation stops when it reaches these limits.

Figure 6 shows the set-up before the treatment. The ionization chamber (PTW 30010, Freiburg, Germany) was placed in the middle of the set-up to measure absolute dose. During the dose delivery, absolute doses were measured and recorded for both the single arc and multi-arc plans.

Figure 5: Set-up of the phantom before treatment. Photo Taken at Huadong, Hospital, China
The comparison between the two techniques was calculated using the formula below,

\[
\% \text{error} = \frac{|EV - SV|}{SV}
\]

Where EV is the calculated mean dose using the TPS while SV is the absolute dose measured during the dose delivery. The above formula was used to see percent error between the measured dose and the calculated dose.

To further assess the result, the percent difference between dose calculations using with and without projection-based MAR algorithm was evaluated using the formula below,

\[
\% \text{difference} = \frac{|Mars - No Mars|}{No Mars}
\]

Where Mars is the calculated mean dose with MAR correction using the TPS for each single and multi-arc plan while No Mars is the calculated mean dose without MAR correction. This serves as a percent difference investigation of MAR and without MAR calculated treatment plan doses.

During the dose delivery, LINAC Trilogy (Varian Medical Systems, Palo Alto, CA) was used with Millennium 120 Multi-leaf collimator (MLC) and Rapid Arc Technology. Absolute doses were measured at the isocenter (the center of the target...
volume) using the ion chamber mentioned above. This was a point dose measurement. Beam was delivered with 6 MV photon energy. The energy is often preferred over 15 MV due to its high conformal, good PTV coverage while sparing healthy tissues shown in studies for carcinoma of cervix doses [23] and spinal cord and lung doses [24].

2.4 Analysis and Discussion

The treatment machine was calibrated with measured output of 0.998 cGy/MU, which is within the tolerance of 1 cGy/MU. Pressure and temperature corrections were applied to the output measurement for this calibration.

Table 2: Table showing calculated mean dose in cGy with and without MAR. It includes measured absolute dose during treatment and its respective percent errors.

<table>
<thead>
<tr>
<th>Plans</th>
<th>Mean Dose (cGy) MAR</th>
<th>Mean Dose (cGy) Without MAR</th>
<th>Experimental Dose (cGy) Absolute dose</th>
<th>Percent error (%) MAR to absolute dose</th>
<th>Percent error (%) Without MAR to absolute dose</th>
<th>Percent difference (%) MAR to without MAR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Single-arc plan</td>
<td>Avoid</td>
<td>208.9</td>
<td>200.5</td>
<td>211.9</td>
<td>1.4</td>
<td>5.7</td>
</tr>
<tr>
<td>Partial Arc and full Plan</td>
<td>1/4 arc</td>
<td>194.6</td>
<td>186.5</td>
<td>196.6</td>
<td>1.02</td>
<td>5.4</td>
</tr>
<tr>
<td></td>
<td>½ arc</td>
<td>196.1</td>
<td>190</td>
<td>197.4</td>
<td>0.65</td>
<td>3.9</td>
</tr>
<tr>
<td></td>
<td>¾ arc</td>
<td>194.9</td>
<td>190.2</td>
<td>196.7</td>
<td>0.09</td>
<td>3.4</td>
</tr>
<tr>
<td></td>
<td>1 arc</td>
<td>192.2</td>
<td>188.9</td>
<td>196.8</td>
<td>2.3</td>
<td>4.2</td>
</tr>
</tbody>
</table>
Using Spectral Imaging MAR Algorithm, there was a percent error ranging from 0.09-2.3 to the mean dose obtained from TPS in comparison to the absolute dose obtained during the dose delivery. Meanwhile, without using MAR, the percent error ranges from 3.4-5.7. There was a percent difference ranging from 1.7-4.2 between with and without MAR. Since the two plans were designed to avoid or minimize influences from high-density materials, the relationship between the arcs was observed. From the data above, the error from absolute dose of the single arc plan that avoided the metals (1.4%) is much higher than the ¾ plan (1.02%). We can infer that avoiding the metal may not reduce the effect of high-density materials. Thus, a relationship between arcs and dose delivered cannot be strongly established or may not exist at all. A large deviation from the delivered dose from the calculated dose is evident. Therefore, using MAR gives accurate treatment dose calculation to the patient and is very beneficial. GE Spectral Imaging showed superior dose calculation than without using any algorithm at all. Numerically, if a conventional treatment is planned with a prescribed dose of 60 Gray (Gy), the PTV will only be irradiated at 56.58 Gy. If metal artefacts exist and do not receive proper measures to eliminate or avoid it, such percent dose difference from absolute dose is concerning and might lead to reduced tumor control probability (TCP).
3. Dual-Energy Imaging Method

3.1 Motivation

Metal Artefact Reduction algorithms were commonly implemented using CT. This research is designed to use Cone-Beam Computed Tomography (CBCT). CBCT has reduced dynamic range, lower modulation transfer function (MTF), and increased scatter compared to CT. CBCT is simply a volumetric Reconstruction of on-board imager (OBI) projection images. According to Landry et al. [32], dual energy CBCT can provide high dose accuracy due to its ability to correct homogeneity problems through online IGRT and target localization. Also, according to Hao Li et al. [28], using their novel dual-energy algorithm technique, dual-energy CBCT can produce good metal artefact reduction with the same total dose as a single energy CBCT. In the diagnostic range, attenuation coefficient is a linear combination of photoelectric component and Compton Scatter component. These components are approximated by basis materials A and B. Using dual-energy (one high and one low), an incident radiation $I_0$ and exit radiation $I$ with equivalent composition material thicknesses $x_A$ and $x_B$ with their respective attenuations $\mu_A$ and $\mu_B$ can be modeled as:

$$\ln\left(\frac{I}{I_0}\right) = -\mu_A(E)x_A - \mu_B(E)x_B$$
Which can be further generalized for monochromatic beams as

\[
\frac{x_A}{x_B} = \left[ -\mu_A(E_L) - \mu_B(E_L) \right]^{-1} \ln \frac{I_L}{I_0} \left[ -\mu_A(E_H) - \mu_B(E_H) \right]^{-1} \ln \frac{I_H}{I_0}
\]

However, for polychromatic beams, a set of equations are produced which makes the solution more complicated. Cardinal et Al. [28] devised an approximate solution as

\[
x_A = a_0 + a_1 L + a_2 H + a_3 L^2 + a_4 LH + a_5 H^2
\]

\[
x_B = c_0 + c_1 L + c_2 H + c_3 L^2 + c_4 H + c_5 H^2
\]

Where \( H = \ln \left( \frac{I_H}{I_0} \right) \) and \( L = \ln \left( \frac{I_L}{I_0} \right) \) and constants a-d are experimentally determined. These equations are further broken down into [28]:

\[
L = L_L - aL_H = \log(I_L) - a\log(I_H)
\]

\[
I_L = I_0 e^{-\mu_L T - \mu_{BL}}
\]

\[
I_H = I_0 e^{-\mu_L T - \mu_{BH}}
\]

Where \( \alpha \) is the weighting factor, \( I_L \) and \( I_H \) are the initial images, \( \mu_L \) and \( \mu_H \) are the attenuation coefficients, \( L_L \) and \( L_H \) are the logarithmically processed images.
3.2 Materials

A pig’s head, pig’s leg, and a water bucket were used for the second part of the research. After using a water phantom, the study was engaged to testing projection-based MAR and dual-energy method using a water bucket and animal tissues as shown in Figure 7. The titanium used was 1.5 cm in diameter and 12 inches in length inserted into the pig’s leg and 0.125 inches in diameter and 12 inches in length inserted in the pig’s head. Also, four 2 cm alloys were placed inside the water bucket. Polylactic Acid (PLA) that served as the tumor was also inserted in the pig’s leg and head. Materials used are shown in Figure 7.
Figure 6: Materials used: (a) Pig’s Head, (b) Pig’s leg, (c) Water Bucket, (d) titanium, (e) Polylactic Acid (PLA), and (f) Alloy
3.3 Methodology

CT images (shown in Figures 9 to 13) were taken using Discovery CT590 RT (GE Healthcare, Milwaukee, WI, USA) using its Metal Artefact Reduction (SMART MAR, GE Healthcare) Algorithm. The CT machine has 100 kW high-powered tubes. Energies used were 80 and 120 kV with slice thicknesses of 0.625 mm and 1.25 mm for the water bucket, and 1.25 mm for both pig’s leg and head. The reconstructed images were taken with and without MAR Algorithm correction. CBCT Images were taken from TrueBeam (Varian Medical Systems, Corona, CA), using energies 80 and 140 kV, and currents-exposure time of 1mAs and 0.8mAs, respectively, for the measurements. Slice thickness was 1.25 mm for both the pig’s leg and head. The Treatment Planning System (TPS) used was Eclipse (version 13.6, Varian Medical Systems, Corona, CA). A Dual-Energy reconstruction code was developed at Duke University for generation of dual energy reconstruction images of the pig’s leg and head. The image generated was chosen at 60 kVp. MatLab (The Mathworks, Inc., Natick, Massachusetts, USA) codes for header and CBCT normalization with CT HU values were created. The Matlab code header was needed to make the images compatible with the computer configuration. Figure 8 shows the normalization curves obtained for the pig’s leg and head.
Figure 7: Normalization Curves for the Pig's leg (top) and Pig's head (below)
According to Katharia et al. [31], a parameter can be used to objectively measure homogeneity called the homogeneity index. The Homogeneity Index (HI) is defined as

\[
HI = \frac{D_{\text{max}} - D_{\text{min}}}{D_p} \times 100
\]

Where \(D_{\text{max}}\) is the maximum dose while \(D_{\text{min}}\) is the minimum dose, \(D_p\) is the prescribed dose. From the above equation, to get a homogeneous dose across the PTV, a null value is desirable. This is another parameter used in this part of the research to evaluate results.

### 3.4 Analysis and Discussion

Results showed a tradeoff between maximum, minimum, and mean dose to the planning target volume (PTV). The pig’s leg (MAR, without MAR, and Dual-Energy), Pig’s head (Dual-Energy), and 0.625 mm slice Water Bucket (MAR) were able to meet the condition for maximum dose on the PTV while other experimental combinations failed to meet the criteria set (see Appendices A and B). As for the minimum dose, no one was able to meet the criteria. However, mean dose criteria was met by all images using all techniques.
Table 3: With and without projection-based MAR, Dual-Energy, and Homegeneity index for each technique

<table>
<thead>
<tr>
<th>Technique</th>
<th>Max Dose (%)</th>
<th>Min Dose (%)</th>
<th>Mean Dose (%)</th>
<th>Homogeneity Index (HI)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water Bucket</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(0.625 mm slice)</td>
<td>MAR</td>
<td>101.8</td>
<td>95.5</td>
<td>100.3</td>
</tr>
<tr>
<td></td>
<td>Without MAR</td>
<td>108.2</td>
<td>87.2</td>
<td>99.3</td>
</tr>
<tr>
<td>Water Bucket</td>
<td>MAR</td>
<td>103.6</td>
<td>90.5</td>
<td>98.7</td>
</tr>
<tr>
<td>(1.25 mm slice)</td>
<td>Without MAR</td>
<td>106.5</td>
<td>80.1</td>
<td>98.2</td>
</tr>
<tr>
<td>Pig’s Head</td>
<td>MAR</td>
<td>105.1</td>
<td>91.0</td>
<td>99.6</td>
</tr>
<tr>
<td></td>
<td>Without MAR</td>
<td>106.1</td>
<td>87.9</td>
<td>98.3</td>
</tr>
<tr>
<td></td>
<td>Dual Energy</td>
<td>102.9</td>
<td>96.4</td>
<td>99.7</td>
</tr>
<tr>
<td>Pig’s Leg</td>
<td>MAR</td>
<td>103.3</td>
<td>93.7</td>
<td>99.7</td>
</tr>
<tr>
<td></td>
<td>Without MAR</td>
<td>101.9</td>
<td>86.7</td>
<td>98.0</td>
</tr>
<tr>
<td></td>
<td>Dual Energy</td>
<td>102.0</td>
<td>95.2</td>
<td>99.7</td>
</tr>
</tbody>
</table>
Table 4: Percent Difference between different Techniques

<table>
<thead>
<tr>
<th>Technique</th>
<th>Max Dose (%)</th>
<th>Min Dose (%)</th>
<th>Mean Dose (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Water Bucket (0.625 mm slice)</td>
<td>MAR and Without MAR</td>
<td>6.4</td>
<td>8.3</td>
</tr>
<tr>
<td>Water Bucket (1.25 mm slice)</td>
<td>MAR and Without MAR</td>
<td>2.9</td>
<td>10.4</td>
</tr>
<tr>
<td>Pig’s Head</td>
<td>MAR and Without MAR</td>
<td>1</td>
<td>3.1</td>
</tr>
<tr>
<td></td>
<td>MAR and Dual-Energy</td>
<td>2.2</td>
<td>5.4</td>
</tr>
<tr>
<td>Pig’s Leg</td>
<td>MAR and Without MAR</td>
<td>1.4</td>
<td>7.0</td>
</tr>
<tr>
<td></td>
<td>MAR and Dual-Energy</td>
<td>1.3</td>
<td>1.5</td>
</tr>
</tbody>
</table>

Deviation percent difference for mean doses between different techniques ranges from 0 to 1.7. However, a large deviation of 1.5 to 10.4% is seen for minimum dose and 1-6.4% for maximum dose. Projection-based MAR technique gave a more accurate dose calculation for both 0.625 and 1.25 mm slice water bucket images. There is less dose homogeneity since minimum doses for both slices gave large deviations of 8.3% and 10.4%, respectively. This means that dose calculation on CT without MAR is 10% less
homogeneous than with MAR. If this is translated in numbers, for a conventional IMRT fraction size of 60 Gy, the PTV is only getting 56 Gy on the PTV.

For the dual-energy imaging method, there were deviations of 1.5% and 5.4% between MAR and the Dual-Energy Method for pig’s leg and head, respectively. From the previous paragraph, it was claimed that MAR was better than without using any algorithm at all. However, results showed that improvements could still be made using the dual-energy method. Between the pig’s leg and pig’s head, the deviation is more evident on the latter. It may be accounted for by the position of the metals inside the pig’s leg and head. As for the pig’s head, the metal was located way deeper in the tissue than in the pig’s leg. The more tissue the radiation needs to pass through, the more evident the effect will be. For all the techniques, there is no strong evidence of deviation on the mean dose calculations. The mean dose percent difference only differs from 0-1.7%. Also, the homogeneity indexes of dual-energy method for both pig’s leg and head are closer to the desirable null value.

Overall, the dual-energy method provided more accurate dose calculation and has better PTV coverage than the MAR Algorithm.
4. Concluding Remarks

For the first part of the research study, the percent error between the mean dose and the absolute dose with metal artifact correction using the GE Spectral Imaging (GSI) method ranged from 0.09-2.3 per fraction. On the other hand, without using GSI, the percent error ranged from 3.4-5.7% per fraction. There is a percent difference ranging from 1.7-4.2% per fraction between with and without GSI. The GSI-based MAR technique improves treatment dose calculation. The percent error of single arc plan was 1.4% which has higher value than the ¾ arc plan with 1.02 %. Thus, avoiding the metal does not guarantee less effect of high-density materials on the dose calculation accuracy. The relationship between the multi-arc plan and the single arc plan was not established.

The second part of the research uses the GE Metal Artefact Reduction (MAR) Algorithm and a Dual-Energy Imaging Method. Calculated Doses (Mean, Minimum, and Maximum Doses) to the PTV were compared. A range of 0.1-3.2% difference was observed for the maximum dose values, 1.5-10.4% for minimum dose difference and 1.4-1.7% difference on the mean dose. A homogeneity index ranging from 0.068-0.065 for dual-energy method was calculated. HI values of 0.063-0.141 with and 0.152-0.264 without projection-based MAR algorithm were also computed. Based from direct numerical observation, there is no large deviation between mean doses of different
techniques but it is evident on the maximum and minimum doses. HI for dual-energy method almost achieved the desirable null values. In conclusion, the dual-energy method improved dose calculation accuracy of the planning treatment volume (PTV) for images with metal artefacts compared to GE MAR Algorithm.
Appendix A: Treatment planning for water bucket

The four metals seen separated in the CT water bucket images were delineated as lower left, lower right, upper left, and upper right. A ring was contoured 0.2 cm away from the PTV but avoiding PTV. The treatment planning used four-field (box) plan for the water bucket. The beam angles were 0, 90, 180, and 270 degrees. The plan was optimized with lower and upper objectives for PTV at 98% and 102%, respectively with both priorities at 100.

Figure 8: Water Bucket Image taken with projection-based MAR
Figure 9: Water Bucket Image taken without projection-based MAR
Figure 10: 1.25 mm slice Images taken at energies 80kV (upper left), 140 kV (upper right), and the reconstructed 60 kV (bottom)
Appendix B: Treatment planning for pig’s head and leg

After the dual-energy images were generated, MatLab was used to make the DICOM images compatible with Eclipse treatment planning system. Tumor contour was based on the shape of the PLA seen on the CT images. Titanium was also delineated as bone. Like the water bucket treatment planning, a ring was contoured 0.2 cm away from the PTV but avoiding PTV. Rings’ purpose was to eliminate hotspots outside the target volume. The treatment planning used four-field (box) plan for with beam angles 0, 90, 180, and 270 degrees. The plan was optimized with lower and upper objectives for PTV at 98% and 102%, respectively with both priorities at 100.
Figure 11: 1.25 mm slice Images taken with (upper left) and without (upper right) projection-based MAR algorithm, and the reconstructed 60 kV (bottom) of the pig’s leg.
Figure 12: 1.25 mm slice Images taken with (upper left) and without (upper right) projection-based MAR algorithm, and the reconstructed 60 kV (bottom) of the pig’s head.
Reference


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