

USING THE GAMMEX MERCURY 4.0™ PHANTOM FOR COMMON CLINICAL TASKS IN CT



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Automatic exposure control (AEC) is essential to ensure proper dose management for your patients.

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Section A: Introduction

In CT, automatic exposure control (AEC) is essential to ensure proper dose management for your patients. AEC implementations are ubiquitous on all modern multi detector CT scanners. AEC systems are essential because small changes in patient thickness produce large changes in the number of x-ray photons making it through the patient. The “rule of thumb” for diagnostic CT energies is that for every 4 cm of soft tissue added to the beam, the tube output needs to double to maintain the same x-ray fluence post patient. Considering the same CT scanner is commonly used to image newborns to bariatric adults, we need to understand how to optimize CT protocols for patient sizes ranging from 10 cm up to over 50 cm!

AEC systems in use today come in many different flavors and types. There are fundamental differences in the options for setting up an AEC system, the parameters that influence the behavior of the AEC, and the behavior of the AEC system as a function of patient size. Modern CT scanners can modulate tube current (mA) and select the optimal beam energy (kV) as a function of patient size and the clinical task. CT scanners assess the patient size using information obtained from a CT localizer radiograph, more commonly referred to as a “scout”, “topogram”, or “surview” by some of the major CT vendors. The localizer image contains information on both the “size” (i.e. the amount the patient’s body attenuates the CT beam) and position of the patient which lets the CT scanner estimate how much mA and what kV to use to properly image the patient. The CT scanner knows what level of image quality (e.g. noise) is required because the user selects a target level of image quality prior to scanning. For kV selection, the user tells the scanner what type of image is being acquired in terms of how important the visualization of Iodine contrast is.

Given the complexity of any single vendor’s implementation of AEC systems, understanding the behavior of these systems is not trivial. When tasked with optimizing protocols across many different vendor’s scanners makes, models, and software versions, a systematic approach to characterizing AEC is needed.

In this white paper, we discuss several clinically relevant tasks a CT protocol optimization team will carry out. We will discuss the role the Mercury 4.0 phantom can play in assisting with the completion of these tasks in detail. Examples will be given on analyzing scan data acquired using the phantom. While not required to effectively use the phantom, example computer code is provided in the appendix of this whitepaper to facilitate data analysis.

This whitepaper focuses on just AEC aspects of a CT scanner and how they can be characterized using the Mercury 4.0 phantom. The phantom contains many different inserts making it useful in characterizing spatial resolution, contrast, HU accuracy, and noise. The work of Wilson *et al.* [Wilson *et al.* 2013] describes these other important uses.

Section B: Phantom Description

Figure 1 depicts the Mercury 4.0 Phantom referenced throughout this whitepaper. The phantom was designed by Dr. Ehsan Samei at Duke University to meet the following demands:

- Performance and effectiveness characterization of Automatic exposure control systems
- Evaluation of iterative reconstruction methods
- Size based image quality evaluation



Figure 1. Mercury 4.0 Phantom

For the purposes of this whitepaper, the inserts are ignored and only the phantom's 5 different sized regions are utilized for AEC characterization. Each of the modules are separated by a 4 cm tapered region. Modules 1-5 have lengths of 7.5, 9, 6, 6, and 7.5 cm respectively. Module 2 is longer than the rest due to the z-resolution Solid Water 10 degree ramp between module 2 and the taper between modules 2 and 3 (not used in this study). Some technical specifications of the phantom are listed below.

Specifications

Material	Polyethylene
Diameters:	5 sections, from smallest to largest: 16, 21, 26, 31, and 36 cm
Overall Length	52 cm
Contrast inserts (present in each section)	Solid Water, Bone, Polystyrene, 10 mg/mL Iodine, air

Section C: Using the Mercury 4.0 phantom to evaluate the dynamic range of a protocol.

Clinical question addressed → For what patient size will my CT scanner reach a mA minimum or maximum?

Any CT scanner is forced to operate within a range of mA values. This range will change as a function of how “hot” the x-ray tube is at any given moment. For example, the maximum available mA following a multiphasic body exam may be lower due to the large amount of energy inside the tube housing relative to immediately after the tube sat dormant for 30 minutes. The tube may also protect itself by not allowing high mA values when it is not hot enough, i.e. when it is not warmed up properly. The effective output of a CT scanner is most commonly quantified using the effective mAs and CTDIvol. In helical/spiral mode effective mAs is the product of mA and tube rotation time divided by the pitch. In axial/sequence mode effective mAs is simply the product of the mA and rotation time. The CTDIvol is equal to a constant multiplied by the average effective mAs used for a scan. This constant will be a function of many things with beam energy (kV) having the largest influence. When an AEC system is used to image a patient, the scanner will try to select an effective mAs value appropriate for the patient’s size and the AEC setting selected by the user. Ideally, the mA needed to obtain the appropriate effective mAs is within the limits of what the scanner can produce. When the needed mA is outside of the limits of what the scanner can produce, the AEC system will not be able to deliver the image quality requested by the user. Different CT manufactures alert the user to this occurrence in different ways, or not at all. When the CT scanner cannot deliver enough mA, the mA can be thought of as hitting a mA “ceiling”, or “maxing out”. When the CT scanner cannot deliver a low enough mA, the mA can be thought of as hitting the mA “floor” or “minning out”. Hitting the ceiling/floor will result in image quality worse/better than the operator requested respectively. These are usually conditions the designer of a CT protocol wants to avoid. In some cases, it may be impossible to obtain the mA level needed by the scan, in such cases, it is useful to understand at what patient size a protocol will reach a mA ceiling or floor.

The Mercury 4.0 phantom represents a large range of patient sizes and can be used to:

1. Evaluate if a protocol will reach a mA ceiling or floor
2. Determine for what patient sizes a protocol will reach a ceiling or floor in mA

Table 1 shows an example of using the phantom to evaluate 4 different CT protocols. We show a routine adult abdomen pelvis, pediatric newborn, pediatric teenager, and a modified adult routine abdomen protocol in Table 1. At our institution, the first three protocols listed in Table 1 are used clinically, while the fourth is a modified version of the routine adult abdomen pelvis protocol. The modifications made resulted in a higher dynamic range of possible effective mAs (i.e. CTDIvol) values to be delivered. Figures 2a, 2b, 2c, and 2d plot the effective mAs and phantom size (i.e. water equivalent diameter (WED)) as a function of phantom module for each protocol. The routine abdomen protocol shown in Figure 2a reaches an effective mAs minimum while transitioning from the 3rd to the 2nd module. This informs us that this protocol is not well suited to image small patients as it doesn’t allow for small enough mA values to be delivered for modules 1 and 2. The newborn protocol results shown in Figure 2b demonstrate how module 1 is at the upper size range for the newborn pediatric protocol as the effective mAs is near its maximum for module 1 and at its maximum for all other modules. Fortunately, the manufacturer

is, as of this writing, developing a 10 cm module specific for newborn pediatrics. The pediatric teenager protocol results are shown in Figure 2c and demonstrate almost no effective mAs issues except for the largest module where they do reach the mA ceiling. The non-clinical protocol set-up to have a large dynamic range in effective mAs demonstrated no issues with mA ceilings or floors.

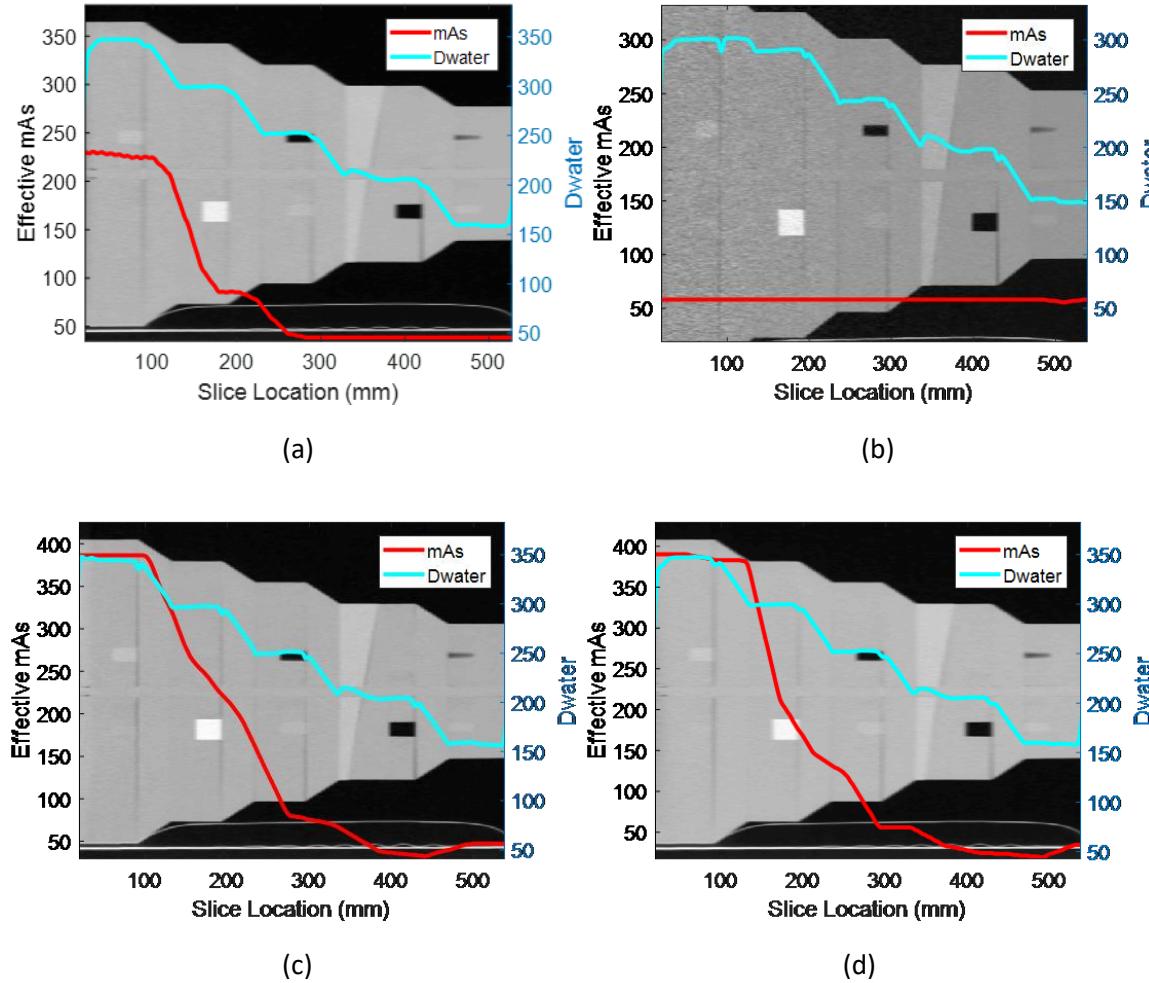


Figure 2a. Effective mAs and WED versus phantom module location for the routine adult protocol detailed in Table 1. Figure 2b. Effective mAs and WED versus phantom module location for the newborn pediatric abdomen/pelvis protocol detailed in Table 1. Note, the WED shown in this figure reaches an artificial maximum as the reconstruction field of view for the scanner was smaller than the size of the largest phantom module. Figure 2c. Effective mAs and WED versus phantom module location for the teenager abdomen/pelvis protocol detailed in Table 1. Only the largest module forces this protocol to request a scanner effective mAs value at the protocol's ceiling. Figure 2d. Effective mAs and WED versus phantom module location for the modified abdomen/pelvis protocol detailed in Table 1. This protocol differs from that used in Figure 2a as the rotation time and mA limits have been increased, allowing for a greater dynamic range in effective mAs output. The phantom image in the background of all the figures is a sagittal slice down the center of the phantom.

Section F of this whitepaper correlates modules within the Mercury 4.0 phantom with patient BMI and weight for different body regions. We also have WED data from several body regions summarized and correlated with the phantom. Before actually scanning any patients, one may use plots of effective mAs, and the data correlating the phantom modules to actual patient sizes, to evaluate their clinical protocol for potential AEC issues.

Practical use cases:

Based on the results shown in Figure 2a, the effective mAs for the routine adult abd/pelvis protocol reaches a minimum right below the size of module 3. This corresponds to a WED of 252 mm. For our institution, this corresponds perfectly with our known WED lower limit on adult abdomen pelvis patients [Burton and Szczykutowicz 2018]. This protocol would not be appropriate for imaging pediatric patients, as we would expect the effective mAs to hit the mA floor due to the mA minimum of the protocol being set at 50 mA and the low pitch value that keeps the effective mAs higher relative to a higher pitch. For the clinical practice at UW-Madison, these phantom results confirmed the protocols in Table 1 columns 1-3 are appropriate for the patient sizes they are designed for.

Practical use instructions:

1. Set up the phantom on your CT scanner's couch. Level the phantom using the provided supports and align its long axis with the z axis of the couch. Center the phantom using your CT scanner's laser system.
2. Select a protocol for evaluation. The protocol should not be modified for this experiment in any acquisition parameter or scan phase. The only exception is turning off bolus tracking and scan delays, it is okay not to have these contrast related scan parameters turned on. You must ensure a CT localizer radiograph series is acquired of the phantom before the tomographic (helical/spiral or axial/sequential) phase is acquired. It is important to acquire the CT localizer radiographs in the same number and order as will be used clinically, as both the order, angle, and number of CT localizer radiographs has been shown to alter the AEC systems of most CT vendors [Merzan et al. 2016].
3. Prescribe a scan range that covers all of the phantom's modules.
4. Scan the phantom.
5. You can either manually navigate to the module you desire to calculate the effective mAs for and record the values for each module, or you can export the scan data and use the script provided in the appendix to plot the effective mAs for each module as shown in Figure 2. As observed in Figure 2, there will be some variation in the effective mAs from slice to slice, so within a single module, several slices should be averaged together to obtain a reliable effective mAs estimate.
6. You should now have the effective mAs for each module in a tabular format (i.e. if you manually record these values from the images) or plotted as shown in Figure 2. It is now easy to understand for which module your protocol reaches an effective mAs ceiling or floor.

Practical advice:

If you find you have a protocol that is not capable of modulating the effective mAs for the size range you desire, simple changes to the acquisition parameters can often be made to increase your protocols effective mAs/CTDIvol dynamic range.

For each specific protocol, always try to use the smallest rotation time possible that will still allow a high enough maximum scanner output for the indication/patient size range. Small rotation times better mitigate motion artifact relative to longer rotation times.

If you have a protocol with an effective mAs/CTDIvol floor low enough for your intended patient population and indication, but the ceiling is much higher than needed, it is likely you can decrease the protocol's scan time by decreasing the tube rotation time or increasing the pitch. In other words, you don't want a protocol intended for scanning teenagers to have the ability to deliver enough dose for a bariatric adult due to too long of a rotation time being used in the protocol.

Increasing the rotation time will raise the effective mAs floor, increase the effective mAs ceiling, and increase the scan time.

Increasing the pitch will lower the effective mAs floor, lower the effective mAs ceiling, and decrease the scan time. Please refer to Ranallo and Szczykutowicz 2015 for an overview of pitch related recommendations. Often, however, lower pitches deliver better image quality when scan speed is not a concern. Lower pitches usually allow for better slice sensitivity profiles, less helical artifact, and lower mA values which decrease focal spot blurring.

Increasing the beam collimation will have no effect on the effective mAs but it will decrease the scan time. It may change the CTDIvol due to geometric efficiency changes.

Increasing the kV will have no effect on the effective mAs, will raise the CTDIvol floor, will raise the CTDIvol ceiling, and will have no effect on scan time. Changing the kV may introduce tube heating problems. Note, in practice, changing kV may change effective mAs in cases where the allowable mA range changes because of a kV change.

When changes are made to scan time, it is likely the mA limits will also change. In general, when one increases the scan time, the maximum possible mA must be decreased. This is why some CT vendors provide mA maximum tables as a function of exposure time. In these tables, you will see longer exposure times corresponding to smaller maximum mA limits. For example, it may not be possible to set up a protocol that requests the absolute maximum mA your scanner vendor reports in their technical specifications. In practice, the tube will always be "hot" and in order to protect itself, it will not let the user scan with the highest mA for a long scan range. It is therefore best to design your scanner protocols knowing this, in other words, never expect the scanner will deliver the maximum mA for anything other than of a single phase short scan range exam. Experience with your scanner will let you know what mA levels can be requested realistically in a high throughput clinical environment.

If your protocol is maxing out at values below your scanner stated maximum and the protocol is a multiphasic protocol, you should turn back on any inter-scan timing delays. These delays may have been turned off in step 2 of the practical use instructions above. Those delays may also provide the tube housing/anode time to cool off allowing you to use higher mA values in addition to providing the needed contrast dynamics.

Using a wide area detector (i.e. ~16 cm coverage) with the Mercury 4.0 phantom is fine, but be sure to center the axial scans over individual phantom modules. Centering the detector over multiple modules, or over a module plus tapered section will make interpreting the results difficult. The effective mAs will be constant for all slices, but the size of the phantom will vary over your scan range.

Table 1. Scan Parameters using in Figure 2. A Discovery HD 750 (GE Healthcare, Chicago, USA) CT scanner was used to acquire the examples shown in all the figures shown in this whitepaper.

	Abdomen Pelvis Medium Adult	Pediatric Abdomen Pelvis Newborn	Pediatric Abdomen Pelvis 13-18 year old	Modified Abdomen Pelvis to increase effective mAs dynamic range
Scan Type	Helical	Helical	Helical	Helical
Detector Coverage (mm) Beam Collimation (mm)	40.0	20.0	40.0	40.0
Detector Configuration	64x0.625	32x0.625	64x0.625	64x0.625
Scan FOV	500.0	320.0	500.0	500.0
Pitch	0.516	1.375	1.375	0.984
Speed (mm/rot)	20.625	27.5	54.999	39.375
Rotation Time (sec)	0.4	0.4	0.7	1.0
kV	120	80	100	120
Smart/ Auto mA or Manual mA	smartmA	smartmA	smartmA	smartmA
Smart mA/ Auto mA Range	50-500	15-200	60-760	10-835
Noise Index	18	12	17	15
(Manual mA)	250	100	55	20
% Dose Reduction Guidance	Not used	Not used	Not used	Not used
Slice Thickness (mm)	5.00	3.75	3.75	5.00
Interval (mm)	3.00	2.25	2.25	3.00
<i>Recon 1:</i>				
DFOV	40	32	40	40
Recon Kernel	STANDARD	DETAIL	DETAIL	STANDARD
Recon Option	Plus Mode On	Plus Mode On	Plus Mode On	Plus Mode On
ASiR Setup	SS40	SS40	SS40	SS40

Section D: Using the Mercury 4.0 phantom to understand how changes in scan parameters affect CTDIvol.

Clinical question addressed → I am asked to decrease the dose by 20% on a protocol, how do I do that?

An AEC system must be provided with some form of user requested image quality target. How this information is given to the scanner varies by vendor and scanner model. An excellent overview of the factors effecting AEC systems by many of the major CT vendors was reported by Merzan *et al.* 2016.

CT AEC systems are not as simple as one would hope. **There are multitudes of factors that will produce different scanner output for the exact same size patient scanned using the exact same AEC image quality target!** The main control “knob” defining the AEC image quality target for GE/Siemens/Canon/Philips is referred to as NI/Quality ref.mAs/SD/DRI respectively. Within each vendors’ AEC system, the following factors, if changed, may all produce different scanner output for the same patient being scanned with the same AEC image quality target: scanner model, software model, body region, slice thickness, patient positioning, bowtie filter (scan field of view or body region usually determines this on most vendors), pitch, CT localizer radiograph technique (mA/kV/angle), coverage of CT localizer radiograph compared to requested scan coverage, scan direction (superior→ inferior or inferior→ superior), kV, collimation, rotation time, scan mode (axial/sequence verse helical/spiral modes), reconstruction field of view, and iterative denoising level. It is naive to believe setting up a scanner’s AEC system is a trivial task. The Mercury 4.0 phantom allows a controlled way to study how a CT scanners AEC system will respond to the factors outlined above.

The phantom represents a controlled subject which can be scanned under a variety of different conditions to systematically understand how your CT fleet’s AEC systems operate.

Figure 3 depicts an example of using the phantom to evaluate how changes in the main AEC image quality target “knob” on a GE CT scanner platform effects image quality. The phantom was imaged using three different AEC settings for the NI parameter as outlined in Table 2. On a GE scanner, the NI reflects the standard deviation observed in a water phantom using the GE “standard” reconstruction kernel at the slice thickness used for “recon 1”. According to the GE user manuals, standard deviation is related to effective mAs as $NI \approx Standard\ deviation \propto \frac{1}{\sqrt{mAs_{eff}}}$, so we would expect to see this functional dependence on effective mAs as we change NI. All of the major vendors can quantify similar relationships for their CT scanners, albeit their functional form may differ. Here, we use the GE terminology and results as an example of one use of the phantom. Figure 3a plots the effective mAs for each module for NI values of 11, 15, and 20. In Figure 3b, we show the ratio of the effective mAs values to the effective mAs from NI = 15 scan. As expected, the scanner output (i.e. effective mAs) is higher/lower for the NI = 11/20 scans relative to the NI = 15 case. Percentage wise, we expect the change from 15 to 11 in NI to produce an effective mAs increase of 85%. Increasing the NI from 15 to 20 should decrease the effective mAs to 56% of its value at a NI of 15. We see these anticipated changes reflected in the actual scan data shown in Figure 3. We can see some size dependence in the ratio of effective mAs as shown in Figure 3b, but the average effective mAs change agrees nicely with the predicted changes in NI.

Practical use cases:

The results shown in Figure 3 demonstrate an easy way to verify and/or characterize claims made by a vendor on the operation of their AEC system. Even if a vendor provides details on how their AEC system responds to changes in AEC control parameters, they may not provide details on how their systems respond to patient sizes. The Mercury 4.0 phantom allows this type of characterization to be made.

A typical task assigned to a clinical medical physicist is to “decrease the dose by X percent”. This instruction is usually given by a radiologist wishing to decrease the dose for a specific protocol. Using the phantom, we can make such a change with confidence. The phantom can be imaged using the clinical protocol, and then the AEC control knob adjusted to where the scanner vendor recommends one would achieve the desired dose decrease. The actual dose reduction, as a function of patient size, can then be determined by comparing the effective mAs between the original and the altered protocol. This is important given the large number of factors effecting AEC system response. For example, the same control knob adjustment that caused a 20% dose reduction in an axial head protocol using an iterative denoising algorithm may not produce the same dose reduction for a helical/spiral protocol set-up for thoracic imaging without iterative denoising.

Practical use instructions:

1. Set up the phantom on your CT scanner’s couch. Level the phantom using the provided supports and align its long axis with the z axis of the couch. Center the phantom using your CT scanner’s laser system.
2. Select a protocol for evaluation. The protocol should not be modified for this experiment in any acquisition parameter or scan phase. The only exception is turning off bolus tracking and scan delays, it is okay not to have these contrast related scan parameters turned on. You must ensure a CT localizer radiograph series is acquired of the phantom before the tomographic (helical/spiral or axial/sequential) phase is acquired. It is important to acquire the CT localizer radiographs in the same number and order as will be used clinically, as both the order, angle, and number of CT localizer radiographs has been shown to alter the AEC systems of most CT vendors.
3. Prescribe a scan range that covers all of the phantom’s modules.
4. Scan the phantom.
5. Repeat the scan with the modified protocol. The modification may be a different AEC system image quality target as shown in Figure 3, or a change in the factors itemized in this section (pitch, collimation, field of view, etc.).
6. For each scan,
 - a. You can either manually navigate to the module you desire to calculate the effective mAs and record the values for each module, or you can export the scan data and use the script provided in the appendix to plot the effective mAs for each module as shown in Figure 3. As observed in Figure 3, there will be some variation in the effective mAs from slice to slice, so within a single module, several slices should be averaged together to obtain a reliable effective mAs estimate.
 - b. You should now have the effective mAs for each module in a tabular format (i.e. if you manually record these values from the images) or plotted as shown in Figure 3.

- Once you measured the effective mAs for each module from each scan, you can divide them to calculate the change in scanner output between the two modes.

The above procedure allows one to calculate the ratio of effective mAs. Under many scanner acquisition and reconstruction conditions, this ratio is equal to the ratio of scanner output (i.e CTDIvol or DLP). There are, however, several exceptions to this:

- Note, here we are assuming the different scan modes were using the same kV if we want to consider the ratio of effective mAs to be equal to the ratio of scanner output. If the beam energy is changed between the scan modes, or if a change is made from single energy CT (SECT) to dual energy CT (DECT) or vice versa, changes in effective mAs no longer reflect changes in scanner output. In other words, the ratio of effective mAs doesn't reflect a change in CTDIvol, DLP, or patient dose when the kV changes. Most vendors provide a look up table in their manuals for calculating the CTDIvol/DLP change when the kV is changed.
- Note, we are assuming the geometric efficiency is the same between scan modes if we want to consider the ratio of effective mAs to be equal to the ratio of scanner output. For example, one can make a change from a 40 mm collimation to a 10 mm collimation and keep the effective mAs constant. The scanner output, as measured by CTDIvol/DLP will increase for the smaller collimation due to the lower geometric efficiency of the smaller beam collimation. Therefore, do not assume the ratio of effective mAs always reflects the ratio of scanner output.
- Note, we are assuming the bowtie filter or any other filtration present in the CT scanner, is unchanged between the scans if the ratio of effective mAs is to be considered the ratio of scanner output. Some vendors provide tables allowing one to scale scanner output as a function of scanner filtration.

Practical advice:

In practice, a human observer cannot actually detect small changes in image noise [Massoumzadeh *et al.* 2009]. In other words, small changes in AEC image quality target control knobs will not be noticeable. Additionally, changes to image acquisition or reconstruction options that result in small changes to scanner output for a fixed AEC image quality target control setting will also not be noticeable. Defining exactly what "small" means in terms of noise standard deviation or scanner output is not well established in the Radiology community. For the practicing clinical physicist, however, knowing how changes in AEC, image acquisition, or reconstruction will affect scanner output should not be guessed. Therefore, we advise using the Mercury 4.0 phantom to characterize the scan mode changes you actually use in your clinical practice, and to characterize new options.

Since so many different scanner options affect AEC, we recommend performing tests as outlined above for each major group of acquisition parameters. For example, it is likely your site has a preferred set of scan modes and beam collimations for imaging the major body regions [Szczykutowicz *et al.* 2015]. For example, for head imaging you may use axial/sequential mode at 20 mm beam collimation. For chest imaging you may use helical/spiral mode and 40 mm beam collimation at a high 1.5 pitch. For abdominal imaging, you may use helical/spiral mode, 38.4 mm beam collimation, and a pitch of 1. Keeping these parameters fixed for head, thorax, and abdominal imaging reduces the number of free parameters for you to explore for each body

region. Once you settle on scan modes and collimations for each body region, it is likely the remaining choices will have to do with overall noise and contrast level, which are primarily determined by AEC image quality target, iterative denoising level, and kV. You can use the Mercury 4.0 phantom to investigate how changing AEC image quality target, kV, and iterative denoising strength affect scanner output. Focusing on clinically relevant parameters to explore makes sense and saves time.

For a more detailed review of using patient scan data to understand AEC parameter changes see Szczykutowicz *et al.* 2015. If you desire to use the phantom to convert one vendor's AEC image quality target parameter to another, you should familiarize yourself with the work of McKenney *et al.* 2014, Sookpeng *et al.* 2016, and Söderberg *et al.* 2010a, 2010b, 2013, 2016 who characterize a wide variety of CT scanners makes and models and present methods for converting AEC parameters between vendor and for characterizing individual scanner model's systems.

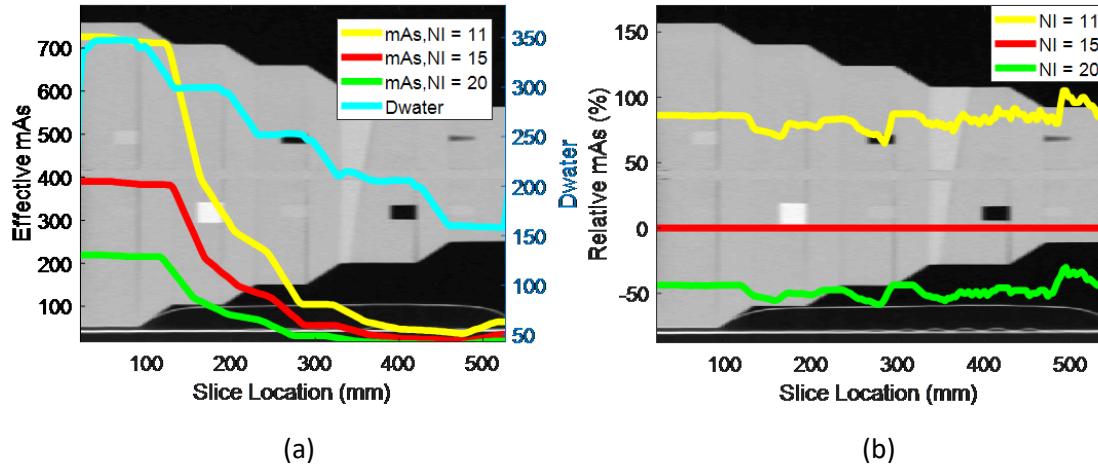


Figure 3. Figure 3a depicts the phantom scanned using the protocols listed in Table 2. The three different scans represent three different levels of requested target image quality, in this example defined using the GE noise index (NI) parameter. Figure 3b. The ratio of the NI 11 and 20 effective mAs curves shown in Figure 3a to the scan acquired using NI = 15. As expected, requesting a lower noise (i.e. NI = 11) relative to the NI = 15 scan resulted in a higher effective mAs. Requesting a scan with higher noise level (i.e. NI = 20) reduced the effective mAs relative to the NI = 15 scan. The phantom image in the background of both figures is a sagittal slice down the center of the phantom.

Table 2. Scan parameters used to generate the results shown in Figure 3. A Discovery HD 750 (GE Healthcare, Chicago, USA) CT scanner was used to acquire the examples shown in all the figures shown in this whitepaper.

	Abd/Pelvis (Full Range)	Abd/Pelvis (LOW NI)	Abd/Pelvis (High NI)
Scan Type	Helical	Helical	Helical
Detector Coverage (mm) Beam Collimation (mm)	40.0	40.0	40.0
Detector Configuration	64x0.625	64x0.625	64x0.625
Scan FOV	500.0	500.0	500.0
Pitch	0.984	0.984	0.984
Speed (mm/rot)	39.375	39.375	39.375
Rotation Time (sec)	1.0	1.0	1.0
kV	120	120	120
Smart/ Auto mA or Manual mA	smartmA	smartmA	smartmA
Noise Index	15	11	20
(Manual mA)	20	20	20
% Dose Reduction Guidance	Not used	Not used	Not used
Slice Thickness (mm)	5.00	5.00	5.00
Interval (mm)	3.00	3.00	3.00
<i>Recon 1:</i>			
DFOV	40	40	40
Recon Kernel	STANDARD	STANDARD	STANDARD
Recon Option	Plus Mode On	Plus Mode On	Plus Mode On
ASiR Setup	SS40	SS40	SS40

Section E: Using the Mercury 4.0 phantom, how can I predict the CTDI_{vol} for a given patient size?

Clinical question addressed → The Joint Commission requires you to know the expected dose range for all your protocols, how can you predict that a priori?

Using an AEC system can be scary to some, you are leaving the choice of how much ionizing radiation to deliver to your patient up to a computer algorithm. This algorithm is very complex, not publicly disclosed, and known to have a large number of often unanticipated dependencies [Merzan et al. 2016]. Often, the only way a site can understand the range of doses delivered by a given CT protocol is by doing a retrospective review of scanner data from a dose monitoring database. Such a review will yield meaningful data on the expected dose ranges for your patients on a scanner and protocol basis but is not an optimal approach. An optimal approach would be to use retrospective data gathered from one's dose monitoring system to confirm their expectations, not define them. The Mercury 4.0 phantom can be used to define expected dose range expectation.

For a single scan using the same focal spot size, beam collimation, pitch, rotation time, kV, and filtration (i.e. bowtie filter) the CTDI_{vol} reported by the scanner will be proportional to the average effective mAs used to acquire the scan data as $\text{CTDI}_{\text{vol}} = k * \overline{\text{mAs}}$. For a scan with N slices, where we know the effective mAs for each slice, this can be written as $\text{CTDI}_{\text{vol}} = \frac{1}{N} \sum_{n=1}^N k * \text{mAs}(n)$. The constant k is given by the ratio of the total scan CTDI_{vol} to the average effective mAs. Since the scanner reports the total CTDI_{vol} and we can get the average effective mAs by averaging the effective mAs from each slice in the scan, we can then calculate the CTDI_{vol} for any slice using $\text{CTDI}_{\text{vol}}(n) = \frac{\text{CTDI}_{\text{vol}}}{\overline{\text{mAs}}} * \text{mAs}(n)$. This derivation only holds true for scan regions where the AEC modulation is non-existent [Dixon and Boone 2013], i.e. within the uniform modules present on the phantom. Over regions for which the mA changes due to AEC responding to change in patient anatomy, the CTDI_{vol} at a given point is not given solely by the effective mAs at that point due to the tails of the dose profiles from adjacent slices as explained by Dixon and Boone 2013. For our purposes of using the phantom to predict the dose in units of CTDI_{vol} for an arbitrary patient size, this is not an issue.

Practical use cases:

For a protocol using AEC without an automatic kV selection algorithm, the Mercury 4.0 phantom can be easily used to produce a look-up table going from patient size to patient dose. Since the phantom represents a large range of patient sizes from small child to large adult, it can provide one with an expected dose range for any patient size. One simply scans the phantom, and then uses the methodology explained above to calculate the conversion factor allowing an arbitrary effective mAs value to be converted to CTDI_{vol}. Or, in practice, allowing one to go to a specific module of the phantom corresponding to the patient size you wish to predict the dose and convert the effective mAs used for that module to CTDI_{vol}.

For a protocol using AEC with tube current modulation and automatic kV selection, it is also possible to obtain a correspondence between patient size and predicted dose. In this case, however, each module of the Mercury 4.0 phantom should be scanned separately so the scanner's kV selection algorithm is allowed to choose a kV for each module resulting in a unique k factor for each kV/module combination.

Practical use instructions:

1. Set up the phantom on your CT scanner's couch. Level the phantom using the provided supports and align its long axis with the z axis of the couch. Center the phantom using your CT scanner's laser system.
2. Select a protocol for evaluation. The protocol should not be modified for this experiment in any acquisition parameter or scan phase. The only exception is turning off bolus tracking and scan delays, it is okay not to have these contrast related scan parameters turned on. You must ensure a CT localizer radiograph series is acquired of the phantom before the tomographic (helical/spiral or axial/sequential) phase is acquired. It is important to acquire the CT localizer radiographs in the same number and order as will be used clinically, as both the order, angle, and number of CT localizer radiographs has been shown to alter the AEC systems of most CT vendors.
3. Prescribe a scan range that covers all of the phantom's modules.
4. Scan the phantom.
5. Record the exam CTDIvol reported by the scanner for the scan you just acquired.
6. Obtain the effective mAs for each slice of the exam. Make sure not to limit your reconstruction range along the z-axis, you need to record the effective mAs from every slice acquired in the exam.
 - a. Make sure you are recording effective mAs values and not just mA values. This will vary scanner vendor to scanner vendor. Effective mAs values should represent the average mA, used for a given reconstructed image, times the tube rotation time divided by the helical pitch. In axial/sequential scanning, the effective mAs is just the average mA used for a given reconstructed image times the rotation time.
7. Divide the CTDIvol you obtained in step 5 by the average of all the effective mAs values obtained in step 6.
8. You can now calculate the CTDIvol for any location within the scan range by multiplying the effective mAs corresponding to that location by the k factor calculated in step 7.
9. If you calculate the CTDIvol received for each module, you can use the Tables presented in section F of this whitepaper to correlate them with actual patient sizes.

Practical advice:

You can forgo any plotting or calculations if you simply desire to measure the CTDIvol for one module. To do this, you would simply scan a single module and obtain the scanner reported CTDIvol used for that module. So long as the effective mAs used for that module was relatively flat over the module, this method should predict the CTDIvol a patient equal in attenuation properties to the module would obtain.

In the real world, you should not expect the scanner to produce the exact same dose on a patient of a size equal to the phantom module you test following these guidelines. No patient will have a uniform size over their entire scan range making picking a phantom module an approximation of the real patient size. Even for a patient of the same average water equivalent diameter as a Mercury 4.0 phantom module, we would not expect the CTDIvol to match exactly. The CT vendor will likely have an algorithm that adjusts AEC both in the z direction and in the angular direction. The angular modulation will be a function of the ellipticity ratio the vendor measures from the patient's CT localizer radiographs [Burton and Szczykutowicz 2018]. Another clinical factor is patient positioning. Changes in patient positioning will cause AEC differences,

albeit the changes are highly dependent on scanner make and model and the direction and number of CT localizer radiographs used [Merzan *et al.* 2016, Szczykutowicz *et al.* 2017, Toth *et al.* 2007, Matsubara *et al.* 2009, Gudjonsdottir *et al.* 2009, Li *et al.* 2007]. We hesitate to provide guidance on how close one can expect the measured CTDI_{vol} using the methods we describe here to match real patient scans due to the clinical issues and AEC system operation details presented here. Since many CT vendors only report their reported CTDI_{vol} values within +/- 15% expected deviation, +/-20% maximum deviation, it is reasonable to expect one could predict a patient's dose with within these ranges of uncertainty.

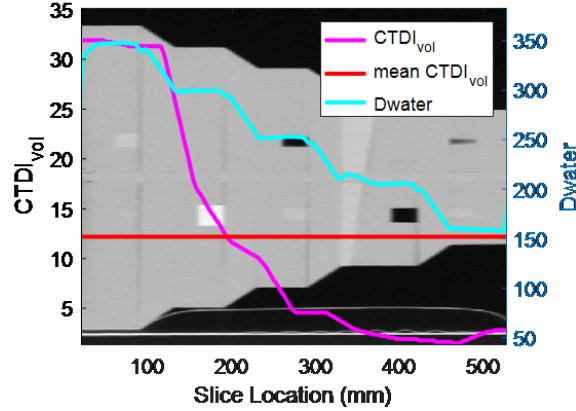


Figure 4. An example of calculating the CTDI_{vol} for each module using the methods described in Section E. The mean CTDI_{vol} value was the scanner reported CTDI_{vol} value for the entire scan. The phantom image in the background of the figure is a sagittal slice down the center of the phantom.

Section F: Matching patient size surrogates to the sizes of the Mercury 4.0 AEC phantom

Clinical question addressed → My patient has a BMI of 19, what module of the Mercury 4.0 phantom does that correspond to?

In medical imaging we use phantoms for several tasks, many in which we desire to have the phantom mimic a patient in some way. With the phantom, we are trying to mimic different patient sizes using polyethylene cast to different circular diameters. As discussed multiple times in this whitepaper, we need to be able to correlate module sizes of this phantom with patient size.

The phantom has five modules of differing dimensions, as seen in Figure 5 and Table 3. Table 3 compares the modules to patient sizes quantified using BMI and weight. Table 4 lists the WED and geometric sizes (AP/LAT based size surrogates) for a wide range of patient ages and body regions. Tables 3 and 4 allow one to identify what module of the phantom to use when a specific patient size is needed to investigate AEC performance over as described in Sections C-E.

WED values will change slightly with beam energy; WED will increase as the kV is lowered and decrease as the kV is raised. This dependence of WED on kV will also be present in humans. For a purely water phantom one should not see a kV dependence on WED as vendors should calibrate CT number to remain constant for water with changes in beam energy.

Table 3 A comparison of the water equivalent diameters (WED) and physical sizes of the Mercury 4.0 phantom to published size surrogates for pediatric and adult patients. WED values listed for the phantom were measured at 120 kV. Note, the smallest module size is not shown in this table as no published study has shown the relationship between size surrogates like WED/ED and BMI/weight for pediatric patients. ¹Measured size surrogate of the phantom. ²BMI and weight values calculating using fit equations provided in Table 2 of Menke 2005.

Phantom Module	WED (mm) ¹	AP (mm) ¹	LAT (mm) ¹	$\sqrt{LAT * AP}$ (mm) ¹	BMI (kg/m ²) Thorax ²	BMI (kg/m ²) Abdomen ²	BMI (kg/m ²) Pelvis ²	Weight (kg) Thorax ²	Weight (kg) Abdomen ²	Weight (kg) Pelvis ²
1	348	360	360	360	43	36	36	127	103	108
2	299	310	310	310	32	27	27	93	78	78
3	253	260	260	260	20	18	18	59	54	49
4	206	210	210	210	9	10	9	26	29	19

Table 4 Typical (mean and min-max values from cited sources) patient size surrogates obtained from the literature.¹Menke 2005 values are <median (min–max)>. ²Burton and Szczykutowicz 2018 values are <mean (min – max)>. The equations relating AP and LAT to age for pediatric patients are from Kleinman *et al.* 2010.

Age and Body region	WED (cm) ¹	WED (cm) ²	AP+Lat (cm) ²	AP (cm)	Lat (cm)
Adult Thorax	28.3 (22-33)	27.3 (19.8-34.8)	62 (48-77.6)	25.5 (18-32) ¹	33.9 (25-40) ¹
Adult Abdomen	29.8 (22-34)	30.6 (22.5-36.9)	61.2 (42-76.2)	23.3 (16-30) ¹	34.1 (26-41) ¹
Adult Abdomen Pelvis		31.1 (23.3-39.1)	59.7 (43.3-77.6)		
Adult Pelvis	29.6 (23-35)			22 (14-29) ¹	35 (26-45) ¹
Adult Head		17.1 (14.4-20.3)	30.8 (25.8-37.6)		
Pediatric head				$16.35 \times \text{age}^{0.068}$ age0.0760	$13.16 \times \text{age}^{0.068}$
Pediatric thorax				$0.6 \times \text{age} + 11.7$	$0.92 \times \text{age} + 16.2$
Pediatric abdomen				$0.57 \times \text{age} + 10.7$	$0.93 \times \text{age} + 15$
Pediatric Abdomen/Pelvis		18.8 (13.6-26.3)	35.8 (26.7 – 50.8)		
Pediatric Pelvis				$0.61 \times \text{age} + 9.9$	$1.15 \times \text{age} + 14.5$

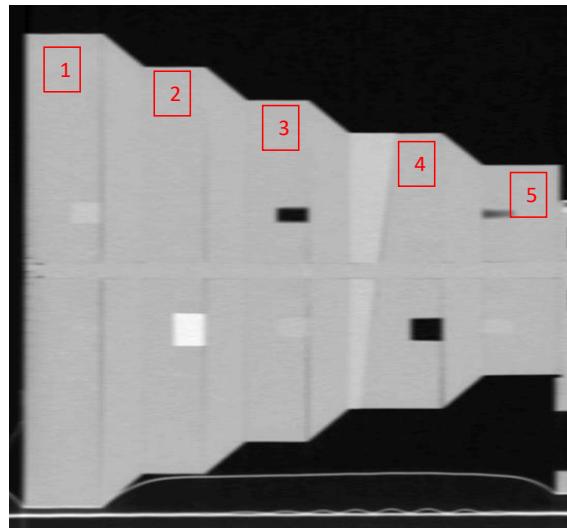


Figure 5. Sagittal reformat of the Mercury 4.0 phantom with the 5 modules labeled. See Table 3 for WED and geometric size surrogate data for these modules.

Practical use cases:

As outlined in Sections C-E, many clinical tasks related to AEC function will require a mapping between Mercury 4.0 phantom module and patient size/body region. Tables 3 and 4 allow this size comparison to be made.

Practical use instructions:

The mapping between phantom module and patient size should be done using WED if possible. WED will correlate the best with a CT vendor's AEC system response, therefore it is the preferred metric to map phantom module to patient size/body region.

Obtaining WED from a patient scan

1. WED information is an output of most CT dose monitoring tools. If you are not familiar with obtaining WED information from your dose monitoring solution, ask the applications team from your dose monitoring vendor how to find it.
2. Some CT vendors now store the WED in the DICOM header of the axial image series, CT localizer radiograph, and or the Dose Slide/SR Dose Report.
3. You can manually calculate the WED using your PACS viewing station following the guidance of AAPM Report 220 [McCollough *et al.* 2014].
 - a. Either choose the central reconstructed image from the patient scan volume, or follow steps b-e for every few slices to get a reliable estimate of the patients WED. AAPM report 220 recommends calculating WED for every slice, but also says using the center slice is a reliable estimate of a scan's WED [Leng *et al.* 2015].
 - b. Make sure the entire patients cross section is contained inside the slice. In other words, if the patient's skin line cannot be seen all the way around the patient, the WED you calculate will be artificially lowered as demonstrated in Figure 2b. It is common in clinical practice to "zoom in" to obtain better detail on exams like spines, temporal bones, angiography, and chest exams. Zoomed in recons usually will not include the skin line all the way around a patient.
 - c. Draw an ROI around the entire patient.
 - i. There are pros and cons to including the patient couch, head holder, or other supporting structures that represent attenuation but are not part of the patient. When patients are actually scanned, support structures like the CT couch or table top will contribute to the WED the CT scanner "sees" and increase the scanner output. In other words, a patient scanned with and without the CT couch would see an effective mAs drop in the "no couch case" if an AEC system was used. In general, if you will be using the Mercury 4.0 phantom to take measurements on the same support structure (i.e. CT couch or table top or head holder) as the patients you are measuring WED, you should only include the phantom and try to exclude support structures from your patient WED measurement.
 - d. Use your PACS tools to measure
 - i. The mean CT number within the ROI (\overline{ROI})
 - ii. The area of the ROI (A_{ROI})
 - e. The WED is equal to $2\sqrt{\left[\frac{1}{1000}\overline{ROI} + 1\right]\frac{A_{ROI}}{\pi}}$

WED is a bit cumbersome to obtain from a PACS system via manual measurement. The effective diameter, AP+LAT, (AP+LAT)/2, AP, or LAT can all easily be calculated from CT localizer radiograph images or non-truncated axial or reformatted CT images. The correlations between the WED and geometric sized based metrics can be found in Burton and Szczykutowicz 2018 if you desire to convert these geometric metrics to WED, or the Tables provided in AAPM Reports 204 and 220 can be used [Boone *et al.* 2011, McCollough *et al.* 2014].

Practical advice:

For head imaging, the difference in attenuation between the CT couch and head holder can be quite large. Figure 6 shows an example of this. This effect should also be expected to be seen to varying degrees between different CT scanner models and within the same scanner model when different couch/table tops (i.e. radiation therapy versus diagnostic) are used.

Obtaining data on the size distributions of the patients at your clinic is most easily performed using the output from your CT dose monitoring vendor. They should allow you access to a “data dump” or “csv export” or perhaps even a direct data base connection. Using any of these tools, you should be able to filter down to a specific protocol and patient size/age range to characterize the size patients you see in your clinic.

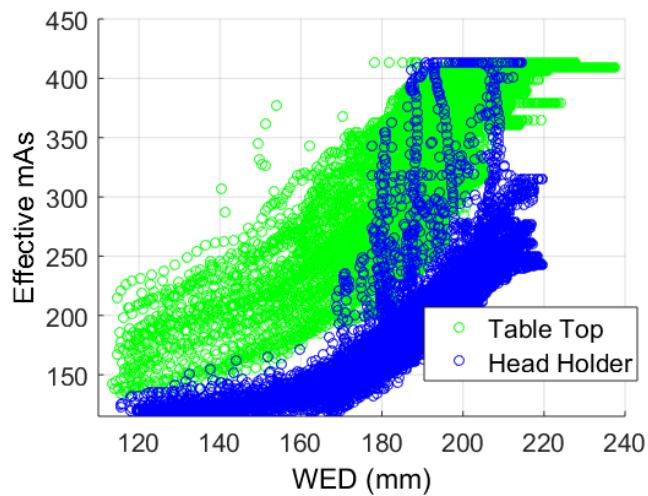


Figure 6. Example of how the scanning of the same anatomical region (i.e. in this case adult heads) scanned using the same protocol with no changes in AEC image quality target on the same scanner can produce different results as a function of what supporting structure was present. In this example, there is a ~1.6 times difference in scanner output for the same sized adult heads between being scanned in the head holder ($n = 193$) and on the CT couch ($n = 71$). Each circle in the plot represents a single CT slice's effective mAs and WED. Unpublished data from the authors.

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Code Appendix

Function to read-in axial CT slices and obtain size surrogates and effective mAs

Pseudo code color coded to match actual Matlab® (The Mathworks, Natick, MA) code is provided below. This code assumes you have a folder containing a stack of axial CT image slices with DICOM headers containing the following fields:

DICOM field tag number	DICOM field tag name
0028,1052	Rescale Intercept
0028,0030	Pixel Spacing
0028,0011	Columns
0008,0008	Image Type
0018,1151	X-ray Tube Current
0020,1041	Slice Location
0018,0060	kVp
0018,9305	Revolution Time
0018,9311	Spiral Pitch Factor

Pseudocode

Input: file location of a stack of axial DICOM CT images

Output: a structure containing: slice location, effective mAs, WED, mA, LAT/AP ratio (ellipticity ratio [Burton and Szczykutowicz 2018]), ED, AP, LAT

For each image in your exam

Read in the DICOM header data

Read in the image data

Create a binary image out of the original DICOM image

Remove the CT couch, blankets, ecg wires etc. from the data

Calculate the AP, LAT, ED dimensions of the axial slice

Calculate the WED diameter using $WED = 2 \sqrt{\left[\frac{1}{1000} ROI + 1\right] \frac{A_{ROI}}{\pi}}$

Pull out information from the DICOM header needed to obtain the effective mAs

End

Re-order the structure containing the metrics we calculated by slice position

Matlab® code

```
function exam = Data_Extraction_From_Axial_Slices(fileDir)
tmp = dir(fullfile(fileDir,'*.dcm'));%returns DICOM files with extension
".dcm"
```

```

for j=1:(size(tmp,1))
    %read in image
    tmp2 = dicominfo([fileDir tmp(j).name]);
    PixData = double(dicomread([fileDir w])) +tmp2.RescaleIntercept;

    %create a binary image for AP, Lat, ED, and AP+Lat calculations
    temp = size(PixData);
    PixData2 = PixData;

    for y = 1:temp(1)
        for t = 1:temp(2)
            if PixData(y,t) < -150 %HU threshold here is 150
                PixData2(y,t) = 0;
            end
        end
    end
    imgg = PixData2(:,:,1);
    imgg(imgg>0) = 1;

    % remove all connected components (objects) that have fewer
    % than 10000 pixels from the binary image imgg
    imgg = bwareaopen(imgg, 10000);

    % Calculating Lateral & Anterior-Posterior dimensions
    widthLat = sum(sum(imgg,1)>0).*tmp2.PixelSpacing(1);
    widthAP = sum(sum(imgg,2)>0).*tmp2.PixelSpacing(1);

    % Calculating Effective Diameter
    A = pi.*((widthLat/2).*((widthAP/2));
    Effdia = sqrt(widthLat.*widthAP);
    prodAPLAT = widthLat.*widthAP;
    sqrprodAPLAT = sqrt(prodAPLAT);
    %ellipticity ratio calculation
    ratio = widthLat/widthAP;

    %calculate WED
    %on a GE CT scanner, the padding values outside the central
    reconstructed part of the image have a value of -3024, these values
    cannot be included in the WED calculation. You should check this value for
    your specific CT scanner make and model, it may differ from -3024. We set to
    -1000 since that is the CT number of air
    PixData(PixData == -3024) = -1000;
    % Limit PixData to only the central part of the image
    countt = 1;
    for h=1:512
        for K=1:512
            if ((h-(tmp2.Width/2))^2+(K-(tmp2.Width/2))^2 <
    (tmp2.Width/2)^2)
                tmpp(countt) = PixData(h,K);
                countt = countt +1;
            end
        end
    end

    %Calculating the water diameter
    meanct = mean2(tmpp);

```

```

Aroi = pi* (double(tmp2.Width)*double(tmp2.PixelSpacing(1))/2)^2;
AWater = .001 * meanct * Aroi + Aroi;
Dwater = 2 * (AWater / pi)^(1/2);

%Pull and store all necessary data from dicom info
if (strcmpi(tmp2.ImageType,'ORIGINAL\PRIMARY\AXIAL')) %make sure
we are looking at an axial image, not a localizer or something else
    exam.mA(j) = tmp2.XrayTubeCurrent;
    exam.sliceLocation(j) = tmp2.SliceLocation;
    exam.kVp = tmp2.KVP;
    exam.T = tmp2.RevolutionTime;
    tmpppp =
cell2mat(strfind(fieldnames(tmp2), 'SpiralPitchFactor'));%axial scans do not
have this
    if (sum(tmpppp)>=1)
        exam.P = tmp2.SpiralPitchFactor;
    else
        exam.P = 1;
    end
    exam.mAs(j) = (exam.imData(count) * exam.T) / exam.P;
    exam.width = tmp2.Width;
    exam.pixelSpacing = tmp2.PixelSpacing;
    exam.ratio(j) = ratio;
    exam.Dwater(j) = Dwater;
    exam.Effdia(j) = Effdia;
    exam.widthLat(j) = widthLat;
    exam.widthAP(coujnt) = widthAP;

end
end %over all images within a given exam

% Reorder Exams Along Z-Axis
[a b] = sort(exam.sliceLocation);
exam.sliceLocation = exam.sliceLocation(b);
exam.mAs = exam.mAs(b);
exam.Dwater = exam.Dwater(b);
exam.mA = exam.mA(b);
exam.ratio = exam.ratio(b);
exam.Effdia = exam.Effdia(b);
exam.widthLat = exam.widthLat(b);
exam.widthAP = exam.widthAP(b);

```