Characterization of the Automated Tube-Current Modulation in a Dual-Source CT Scanner for Gated ECG Imaging of the Heart

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Purpose

Evaluation of dose and image quality based on input heart rate (HR) in cardiac imaging utilizing automated tube-current modulation (aTCM).

Introduction

Dose and Image Quality optimization tools have been integrated into new generation scanners. For instance, tube current modulation (mA) can help reduce the dose to the patients by evaluation the attenuation profile of the patient being imaged. The tube potential (kVp) can be selected to optimize the contrast-to-noise ratio. Additionally, cardiac gated studies take advantage of the cardiac cycle resting phase. In the worst case scenario, the beam stays on 90% of the cardiac cycle i.e. Retrospective Gating. Nevertheless, the dose can be reduced even further by having the beam on only 10% -20% of the cardiac cycle, i.e Prospective gating. Gated studies aim to use the moment in the cardiac cycle when the heart is moving the least to avoid introducing motion blur to the images. In dual source scanners, the temporal resolution is increased twofold. Furthermore, cardiac imaging can be improved by sampling the patient's heart rate and adjusting techniques such as pitch. Understanding how these techniques change as a function of heart rate can provide insight into protocol optimization.

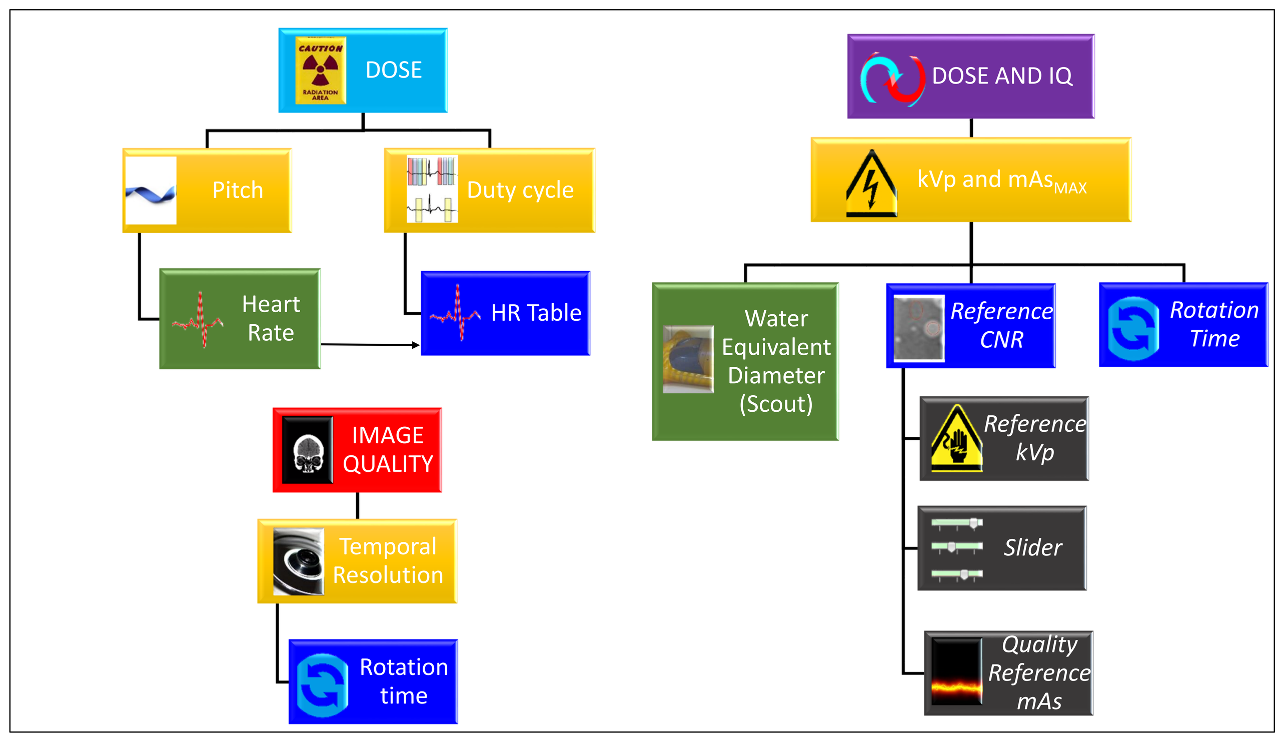

Figure 1. *Dose and Image Quality of Cardiac CT protocols are affected by the Acquisition Protocol Parameters (Yellow) which in turn are affected by Patient Factors (Green) and the Intrinsic Protocol Parameters (Blue).*

Methods and Materials

An electron density phantom representing the size and density of an adult abdomen with inserts of various concentrations of iodine were scanned using aTCM on dual-source CT scanner using single segment helical reconstruction techniques. A cardiac wave generator was used to set the HR. Scans were acquired as a function of HR. The data was reconstructed to optimize the image quality based on EKG waveform. All parameters were extracted from the image header and then evaluated to measure the contrast to noise ratio of each insert in the phantom. A 0.6cc thimble chamber was used to sample the pulse width for each acquisition. In addition, 19 consecutive patient scans were collected over a six-month time period. kV CNR correction factors were established by scanning the phantom in semi mode at all available kV stations and the contrast reference station for our protocol (slider bar =9).

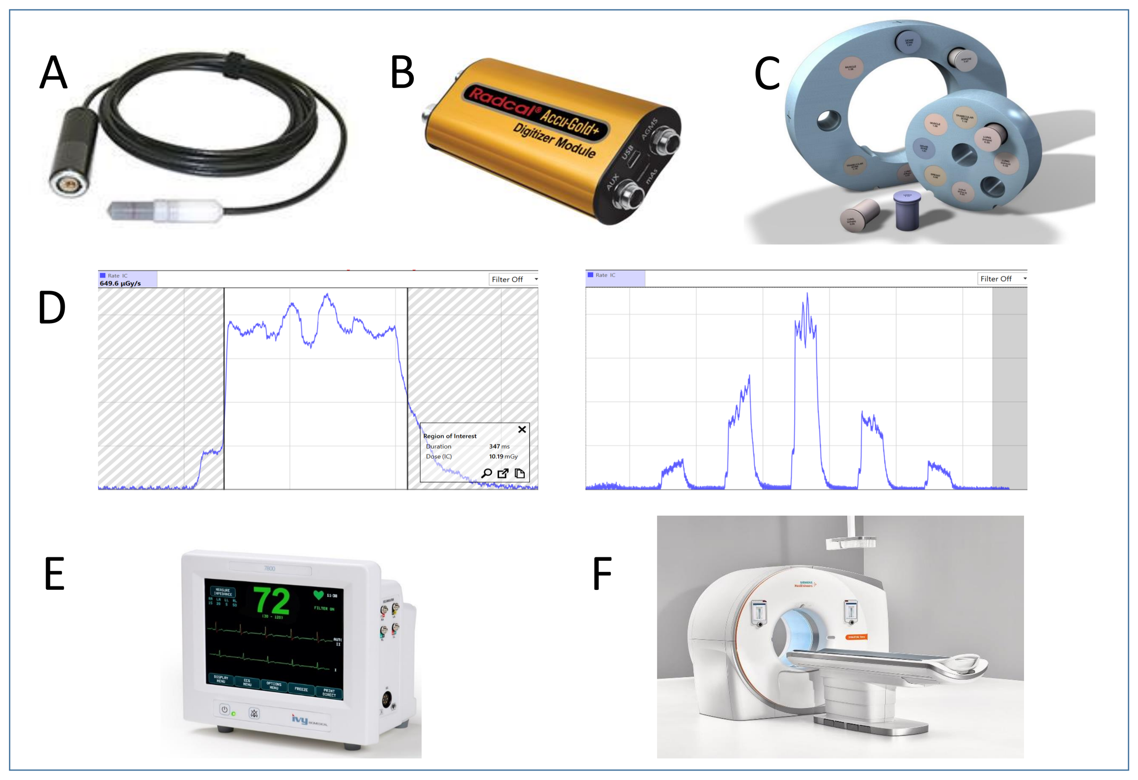

Figure 2. *(A) Radcal thimble ionization detection chamber and (B) digitizer module. (C) CIRS electron density phantom with different materials inserts. (D) Dose measuring software from Radcal Accu-Gold 2. (E) Heart rate signal generator. (F) Siemens SOMATOM Force CT Scanner.*

Results

The scanner output is specified by weighted CTDI to eliminate pitch effects and is corrected for patient size and kV-dependent contrast factor. The kV correction (kVc) adjusts the dose based on the system maintaining a certain level of Contrast-Noise-ration (CNR) which is prescribed in the protocol. The mA correction (mAc) corrects the dose based on patient water-equivalent diameter. This system adjusts the mA differently if the patient is larger or smaller than the reference size. In the case of the phantom CTDIw is only corrected for kVp because the mA is not modulated because there are no thickness changes. The beam is only on at certain intervals of the heart cycle. The fraction of time (f_{HR}) is fraction relative to the R-R interval that beam must be on to acquire enough scan data.

$$f_{HR} = \frac{PW}{T_{RR}} \qquad (1)$$

The relationship between HR and output is best described by the fraction of R-R interval that the beam is on, where image quality is expressed as weighted CTDI, not the standard CTDIvol. This is because the table speed must be adjusted based on the HR to avoid any gaps in acquisition between scans. This relationship begins to breakdown with increasing variability of the HR. The protocol uses an algorithm to maintain CNR at various kVp stations. Therefore, dose is increased at higher kVp and for patients with larger diameter. To correct for this, the CTDIw was corrected using a kVp and mA correction factor (for only patient scans). The kVp tended to be higher for smaller HR's (55-70 bpm).

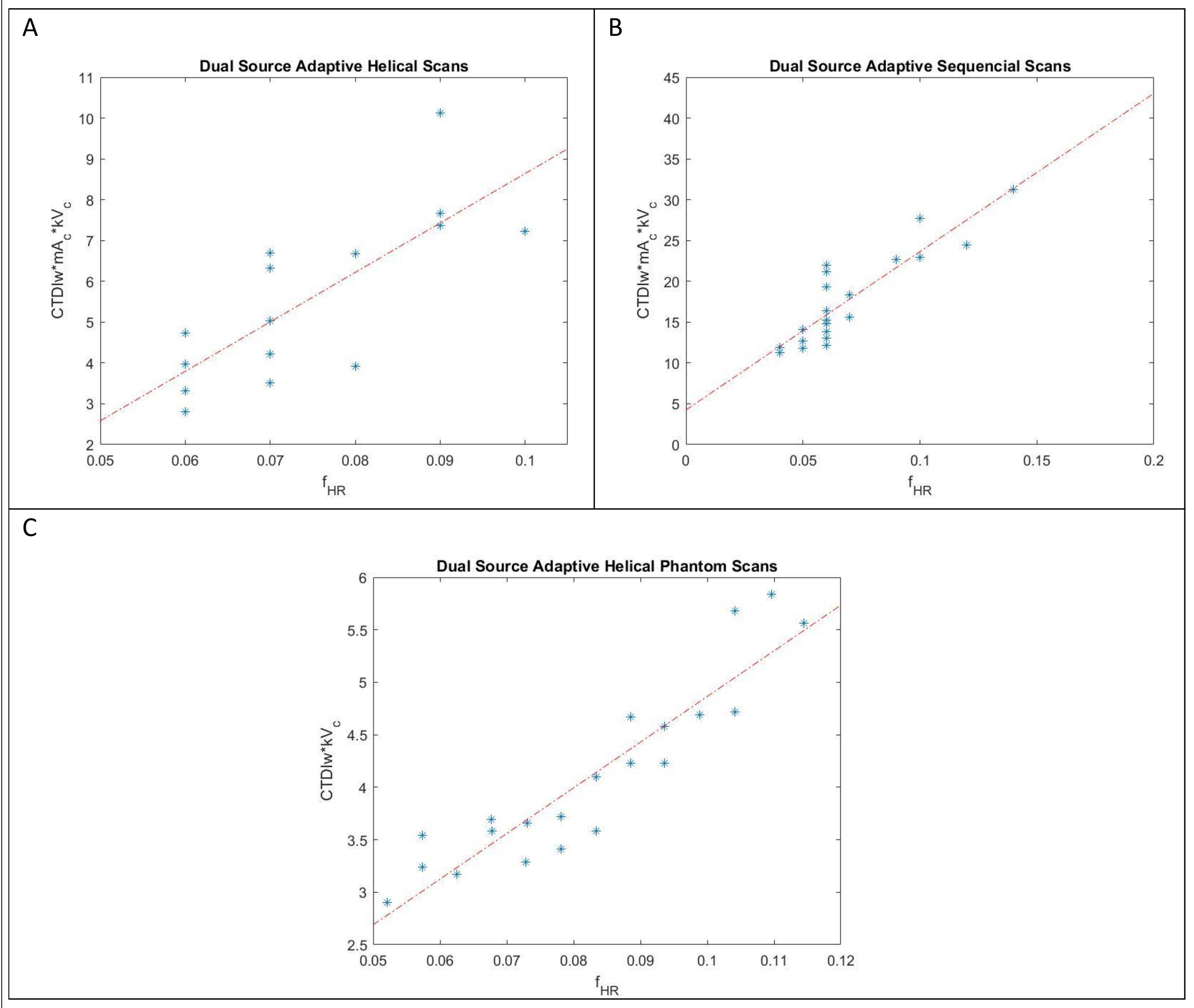

Figure 3. *Plots of the corrected weighted CTDI as a function of fractional heart rate time (f_{HR}). Where (A) contains data form patient helical acquisitions, (B) contains data from patient sequential acquisitions, and (C) contains data form phantom helical acquisitions. All data sets had a p-values less than 0.05.*

During phantom measurements it was found that the measured pulse width did not change regardless of the HR i.e. 150 milliseconds. However, this was not the case in clinical images. In additional measurements (table 1), it was found that by modifying the duty cycle range in the heart rate table of the scanner, the output pulse width (PW) could be altered in two ways for cardiac protocols,

$$PW_{\%} = Duty\ cycle + 150\ [ms] \qquad (2)$$

$$PW_{AT} = Absolute\ Time + 150\ [ms] \qquad (3)$$

where $PW_{\%}$ is the increase in PW based on the duty cycle range i.e. Percent Mode and the PW_{AT} is the increase in PW by a fixed time interval i.e. Absolute Time mode. These adaptive prospective gated scans have a buffer or base pulse width of 150 ms, but it can be increased based on the patient heart rate when a range duty cycle greater than 1% or by a constant amount. In equation 2, a percentage of the patient cardiac cycle is added to the base pulse. This portion of time in the cardiac cycle where the output is maximized and it can be calculated by,

$$Duty\ Cycle\ [ms] = \frac{\%_{end} - \%_{start}}{HR(bpm)} * 600[ms] \quad (4)$$

In general, duty cycle is expressed as a percentage range of the cardiac cycle, but in this case the percentage range was converted to a time interval. Conversely in equation 3, a fixed time interval (AT) can be added to the base pulse which is given by the difference of two arbitrary time intervals,

$$Absolute\ Time = AT_{end} - AT_{start} \qquad (5)$$

The pulse widths measured and the calculated pulse widths were compared and all measurements were within 10% of the calculated value (see table 1).

Table 1. *Measured output pulse widths compared to the calculated pulse widths for different heart rates and different duty cycle inputs in the Heart Rate Table.*

HR (bpm)	Heart Rate Table input	Measured Pulse Width (ms)	Added time (ms)	Calculated Pulse Width (ms)	% Difference
60	200-400 ms	347	200.0	350.0	-0.9%
60	65%-75%	255	100.0	250.0	2.0%
60	70%	147	0.0	150.0	-2.0%
75	400-600 ms	351	200.0	350.0	0.3%
75	100-300 ms	320	200.0	350.0	-8.6%
75	200-400 ms	337	200.0	350.0	-3.7%
87	70-75%	186	34.5	184.5	0.8%
87	65-70%	186	34.5	184.5	0.8%
87	60-80%	282	137.9	287.9	-2.1%
87	65-75%	231	69.0	219.0	5.5%
100	70-75%	193	30.0	180.0	7.2%
100	65-75%	211	60.0	210.0	0.5%
100	65-80%	250	90.0	240.0	4.2%
100	60-80%	282	120.0	270.0	4.4%
100	200-400 ms	345	200.0	350.0	-1.4%

The pitch was another parameter affected by the change in heart rate. The relationship between pitch and heart rate can be defined as follows,

$$Pitch\ (HR) = RotationTime * \frac{HR - 10}{60} * \frac{Slices_{total} - Slices_{overlap}}{Slices_{total}}$$

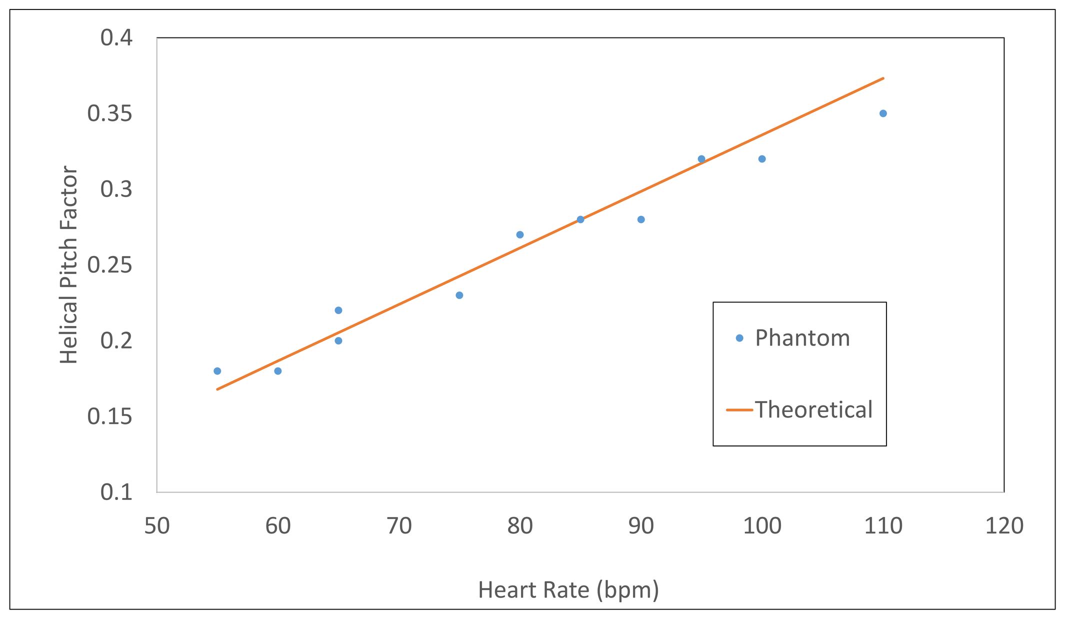

Figure 4. *Pitch change for Routine Coronary protocol as a function of heart rate.*

The image quality was evaluated by calculating CRN of the iodine inserts of the electron density phantom and comparing it the pitch factor. There was no significant trend in the data.

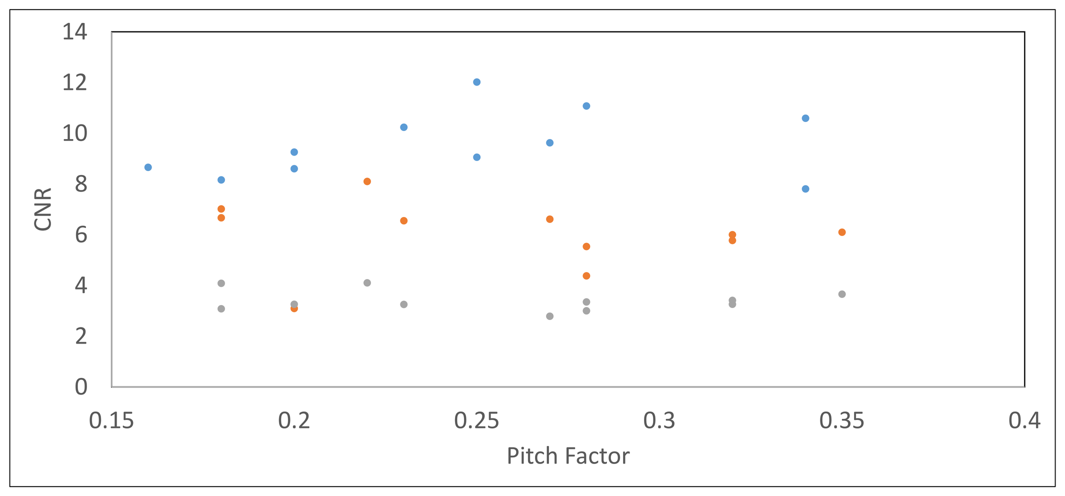

Figure 5. *CNR data for different phantom inserts containing different concentrations of iodine. The CNR values are in different colors for each concentration of iodine.*

While browsing the many extractable DICOM parameters we listed a few that would be useful for making a DICOM standard for Cardiac Imaging. Some of these parameters were buried into a single DICOM tag and had to be sorted out by a script to extract them. Some examples these useful tags are:

1)Image reconstruction time (0020-4000)
2)R-R interval (TRR) (0018-0022): length of time in seconds between consecutive R waves on a EKG waveform
3)Heart rate values (0018-0022): Max BPM/Min BPM/Avg BPM/Variability
4)Series description (0008-103E): displays the Best Phase for Reconstruction
5)Generator Power (GP) (0018-1170) : Helps determine the maximum mA
6)Exposure Modulation Type (0018-9323): PULS_EC_MIN or PULS_EC

Discussion

It was found that image quality is independent on the helical pitch factor. Also, the selected kVp is a determinant on patient dose. The small temporal resolution of the image puts extra burden on the tube current to achieve consistent image noise levels. Different amounts of time can be added as the patient situation changes. For instance, heart rate thresholds can be set in the heart table to add time to the pulse width i.e. Low, moderate and high HR. Dose and image quality are functions of both heart rate and patient size, which determine acquisition techniques. Therefore, when evaluating Cardiac CT protocols for radiation exposures, one needs to consider the scan mode, heart rate, and patient size. In the future, the speed and motion correction algorithms will need to be tested and compared to single source CT Scanners.