Evaluating the Complex Relationship of Automated Tube Current Modulation, Noise Index, Image Noise and Phantom Size

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Purpose
To determine the influence of phantom size on automated tube current modulation (ATCM) performance.

Introduction
Automated tube current modulation (ATCM) dynamically optimizes radiation exposure by modulating x-ray tube current during scanning based on a patient’s varying anatomic attenuation, both in-plane and along the z axis. ATCM plays a central role in controlling radiation dose and maintaining image quality.

While the concept of ATCM is similar across manufacturers, how ATCM is applied, and the parameters used to control it vary (1).

In our study, we utilized a GE Healthcare HD750 CT scanner (General Electric, Milwaukee). ATCM exposure control parameters for this scanner include a quality setting called “noise index” (NI) and minimum/maximum tube current levels. The NI parameter is expected to have a direct, linear relationship with noise and we expected that, for a given NI value, the ATCM algorithm would result in similar image noise regardless of patient size. However, previous studies have reported varying correlation between tube current modulation settings and resultant image noise (2,3).

The purpose of this study was to evaluate the effect of NI on a measure of image noise (standard deviation) across a range of phantom sizes.

Materials & Methods
Four tissue-equivalent abdominal CT dose phantoms (CIRS 007TE) of different sizes were scanned using a GE HD750 scanner. (Figure 1). To simulate an “extra-large” patient size, a 5th phantom was created by wrapping a fat-equivalent ring around the large adult phantom.

We used scanning parameters similar to our routine abdominal CT protocol: 129kVp, 0.8 rotation time, 40mL beam width, 0.964 pitch, and 2.5 mm image thickness. A large scan field-of-view was utilized. Auto-ma and Smart-ma (General Electric, Milwaukee WI) were enabled and Noise Index (NI) was varied. Images were reconstructed using the standard reconstruction kernel.

For each phantom size/NI combination, three regions of interest (ROIs) were placed in an off-center, consistent portion of 10 consecutive images. The standard deviation of each ROI was recorded as our measure of noise. The average of these 30 standard deviation values was calculated and then plotted against NI for each phantom size (Figure 2).

Results
For a given phantom size, noise increased linearly as NI value increased (R2 = 0.989-0.999). However, the slopes (ranged 0.47-1.26) of these plots differed between phantom sizes (Figure 2).

Using a constant NI value and the same scan protocol, noise levels decreased with increasing phantom size. For the small to medium phantom sizes (circumferences of 71, 86, and 96cm), the differences in slopes (1.26, 1.21, and 1.11) were relatively minor. The slopes (0.85 and 0.47) for the large and extra-large phantoms (circumferences of 116 and 138cm) were substantially less compared to the small-medium size phantoms, and also quite different from each other, resulting in three distinct sets of lines on the noise (standard deviation) vs NI plot (Figure 2).

In summary, for large and extra-large phantoms at a given NI, image noise (standard deviation) is less than that seen for small to medium phantoms.

Discussion
ATCM plays a central role in controlling image quality and radiation exposure. Our study shows ATCM is complex. We expected that, for a given fixed NI, the resulting images would have a similar level of noise regardless of phantom size. This was not the case. Our study showed a major difference in ATCM behavior between small-medium and large-extra-large phantom sizes, as well as the diverging linear plots between phantoms in Figure 2. This figure illustrates that, for a constant level of noise across images of different phantom sizes, much larger NI values are needed for large-extra-large phantoms. It is notable that the findings in the literature for ATCM, patient size, and noise have been mixed (2,3). We believe one explanation is that the extra-large phantom is larger than some phantoms and patients previously evaluated, which may explain the discrepancy between our study and another study that showed constant noise across varying patient sizes (2).

Our findings have significant implications for protocol design, particularly given the increasing number of large patients being scanned. Israel et al. showed that patients in the 90th percentile for weight received 3 - 4 times the organ doses of patients in the 10th percentile during ATCM use (4,5).

Our evaluation suggests it may be possible to scan large patients with much higher NI values than probably used previously while preserving diagnostic image quality. We have already used these results to create a “Noise Index Lookup Table” for our low dose surveillance abdomen/pelvis protocol which also utilized Model-Based Iterative Reconstruction (MBIR).

[Refer to separate poster titled, Veo (MBIR) Implementation: Process Description and Lessons Learned]

Further clinical evaluations are needed to determine optimal scan parameters for larger patients. Authors have reported higher subjective image quality with increasing patient size, even with constant image noise (5). This, coupled with our results, may suggest the possibility of significant further dose reduction related to scanning of large patients.

We would note that this study has several limitations. First, we studied only one CT scanner model from a single manufacturer. While the results cannot be directly extrapolated to other scanners, our approach may be useful for assessing the behavior of ATCM for other manufacturers. Secondly, the use of five phantoms does not fully demonstrate the range of patient sizes and shapes encountered clinically. Thirdly, our extra-large phantom was created by the addition of a fat-equivalent layer to the next largest size phantom. The other phantoms, in contrast, differed as a result of scaling. Finally, our study was based on phantoms. We therefore advise caution in implementing our findings when scanning patients. For instance, we used a process of using progressively larger noise indices for large patients rather than going directly to large NI values.

Conclusion
Use of ATCM with a given noise index results in lower image noise for larger phantoms than for smaller phantoms in the CT scanner we studied.

To maintain a more consistent amount of image noise across patient sizes, our data suggests utilizing larger NI values for larger patients.

References