Assessing spectral performance of a dual source dual energy CT scanner

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Aims and objectives

Recently, several techniques for the acquisition of dual energy CT images have become available on clinical CT scanners. In all cases, the same axial slice of the patient is scanned with two different x-ray spectra within a short time interval. However, the scanners use different methods to generate the spectral separation either at the X-ray source or at the detector side. It is therefore important to define a relevant and easy to use quantity for assessing spectral performance.

In principle all required information can be derived from the X-ray tube voltage, beam filtration and detector response, but these properties are often not provided to the user with sufficient precision. In addition the impact of these quantities on the quality of the obtained Dual Energy CT result images is not obvious.

Instead, we propose to use a standardized phantom for the assessment of spectral performance. The technical requirement is that - after a scan with defined CTDIvol - low and high energy images of the phantom must be provided separately by the scanner, so that the iodine enhancement at the two energies can be measured. From this information, the "iodine enhancement ratio" can be calculated, which is the ratio of the iodine CT-value at low kV and the iodine CT-value at high kV.

In order to assess the impact of the iodine enhancement ratio, we have also performed a basic material decomposition into iodine and water in order to obtain virtual non-contrast (VNC) images from the same scan. The obtained noise in the VNC image is a direct consequence of the dose efficiency of the used Dual Energy scanner and its correlation with the iodine enhancement ratio will be evaluated.
Fig. 1: Evaluation phantom (configuration with 20cm diameter) and image processing steps: the low kV image (top left) and the high kV image (top right) result directly from the Dual Energy scan and must be available from independent reconstructions. The mixed image (bottom left) is a weighted average of these images and the primary diagnostic image with best contrast-to-noise ratio. From the same image, iodine contrast can be subtracted by means of base material decomposition (bottom right). In the absence of noise reduction algorithms, noise is increased in this process.
**Methods and materials**

Our evaluation phantom consists of a small sample of diluted iodine contrast agent (15 mg/ml iodine) with a diameter of 2cm. This sample mimics the typical dimensions and enhancement of an iodine filled aorta in a CT angiography scan.

The iodine sample was either scanned in air or in water-equivalent rings with 10/20/30/40cm outer diameter (Fig. 1 on page 6). In all cases the sample was at the center of the phantom as well as at the iso-center of the gantry to obtain a reproducible and well-defined X-ray spectrum.

All dual energy CT scans were performed on a third generation Dual Source Dual Energy CT scanner using the following parameters: abdomen sequence protocol with 128x0.6mm collimation (including z-flying focal spot), 0.5s rotation time, 1.5mm slice thickness, 1mm slice increment, Qr40 kernel, field of view = phantom diameter, dose modulation with CareDose 4D. Five different combinations of spectra were delivered by the 2 x-ray tubes: 80kV and 140kV or 70/80/90/100kV together with Sn150kV, where "Sn" denotes the use of an additional tin filter for hardening of the 150kV spectrum.

For all scans a single rotation was used; the quality reference mAs (default values) at 100/Sn150kV were kept constant for all phantom diameters to achieve a realistic radiation dose, while at the other voltage combinations quality reference mAs were adjusted to get the same CTDIvol at the same phantom diameter as for 100/Sn150kV. In this way, overall radiation dose was the same for all voltage combinations at the same diameter, while the ratio of the low energy mAs relative to the high energy mAs was the same as for a patient of this diameter. According to the patient protocol, the CTDIvol value for 100/Sn150kV at 2, 10, 20, 20, 40cm, respectively, was 2.1, 2.1, 2.5, 7.5, 19.0 mGy.

The obtained low and high kV images were processed with a MATLAB script consisting of the following steps:

- automatic detection of the iodine sample in each slice

- measurement of the central iodine CT-value using a circle ROI with 1cm diameter

- measurement of water CT-value (if applicable) and image noise in a larger ROI surrounding the iodine sample
· water-scaling of the measured iodine enhancement to correct for small drifts of the scanner calibration

· calculation of the diagnostic mixed images with linear image weighting as recommended by the vendor

· calculation of the iodine enhancement ratio

· calculation of VNC images using the measured iodine enhancement ratio (this requires an image based base material decomposition into water and iodine)

· statistical evaluation of all slices to get mean image noise and standard deviation of image noise for the same ROI in water/air as above

It should be noted that the basic VNC images that are obtained from a base material decomposition for each individual voxel are usually not available on commercial products because of the comparatively high image noise. Instead, non-linear noise reduction methods are applied, which obscure the performance of the acquisition system. It is therefore necessary to perform this calculation by hand in order to obtain comparable results for the image noise in the VNC images from different scan modes.
Fig. 1: Evaluation phantom (configuration with 20cm diameter) and image processing steps: the low kV image (top left) and the high kV image (top right) result directly from the Dual Energy scan and must be available from independent reconstructions. The mixed image (bottom left) is a weighted average of these images and the primary diagnostic image with best contrast-to-noise ratio. From the same image, iodine contrast can be subtracted by means of base material decomposition (bottom right). In the absence of noise reduction algorithms, noise is increased in this process.

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Results

Image noise in the mixed image for all voltage combinations and diameters is shown in Fig. 2 on page 9.

For diameters of 20cm or above, the voltage combination 80/140kV performs worse at same dose than all other voltage combinations. Image quality of the voltage combination 70/Sn150kV at 40cm is visibly degraded, which appears to be the reason for the exceptionally low measured image noise for this case. At all other patient diameters image noise of the mixed image is comparable for all voltage combinations with tin filter.

In Fig. 3 on page 9 the iodine enhancement in the mixed image using the recommended default weighting is shown for all voltage combinations.

While the voltage combinations 80/Sn150kV, 90/Sn150kV and 100/Sn150kV have very similar iodine enhancement (designed to be 120kV equivalent), the voltage combinations 80/140kV and 70/Sn150kV correspond to an iodine enhancement obtained at a much lower tube voltage. For example at 20cm, the iodine contrast in the mixed image at 70/Sn150kV is 28% higher than at 100/Sn150kV.

It should be noted that iodine contrast depends strongly on the selected image weighting, which was 0.3 / 0.5 / 0.4 / 0.5 / 0.6 (weight of the low kV image) for the 5 voltage combinations. It can also be adjusted, but choosing higher values usually also increases image noise so that the two voltage combination 80/140kV and 70/Sn150kV in general can be expected to have the highest iodine contrast to noise ratio in the mixed image.

The iodine enhancement ratio (= iodine enhancement at low energy / iodine enhancement for high energy) for all voltage combinations and all patient diameters is shown in Fig. 4 on page 10.

At 20cm water diameter, for which best image quality is expected due to the tuning procedure of the scanner, the measured iodine enhancement ratio was 1.87, 4.03, 3.40, 2.93 and 2.60 at 80/140kV and 70kV, 80kV, 90kV, 100kV in combination with Sn150kV, respectively.

At all patient diameters, the iodine enhancement ratio was lowest at 80/140kV and highest at 70/Sn150kV.
Between 10cm and 30cm diameter, the iodine ratio increased by 0.010/cm for 80/140kV. For the voltage combinations with tin filter it decreased by 0.001/cm, 0.008/cm, 0.009/cm, 0.013/cm, respectively.

Image noise in the VNC image as a function of the voltage combination is shown in Fig. 5 on page 11.

At the same dose, the voltage combination 80/140kV has much higher image noise in the VNC image than all voltage combinations with tin filter. For example at 30cm diameter, image noise is 61% higher than at 80/Sn150kV and 10% higher than at 100/Sn150kV. Again, the data point at 40cm and 70/Sn150kV is not reliable because of the poor image quality.

In Fig. 6 on page 12, the measured ratio between noise in the VNC and in the mixed image is shown as a function of the iodine enhancement ratio.

There is a good correlation between the two quantities, which means that a higher iodine ratio usually also corresponds to less image noise in the VNC image relative to the mixed image. This explains the better noise performance at fixed dose for the scan modes with tin filter in figure 5.
**Fig. 2:** Image noise in the mixed image at same CTDIvol as a function of tube voltage for all phantom diameters.

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**Fig. 3:** Iodine CT-value as a function of voltage combination for all phantom diameters.

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**Fig. 4:** Iodine enhancement ratio as a function of the phantom diameter.

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**Fig. 5:** Image noise in the virtual non-contrast image (VNC) at same CTDIvol as a function of the voltage combination for all patient diameters.

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Fig. 6: Ratio of noise in the VNC image and noise in the mixed image as a function of the iodine enhancement ratio for all voltage combinations and patient diameters. A linear fit is indicated as a dashed line.

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Conclusion

The iodine enhancement ratio can be easily measured using a standardized phantom. For the studied scanner and scan modes, measured values at 20cm phantom diameter were between 1.87 and 4.03 and the phantom diameter only had a small impact on this ratio.

At the same time it has been shown that image noise in the virtual non contrast image (water/iodine decomposition) correlates strongly with the iodine enhancement ratio, so that a high iodine enhancement ratio corresponds to low image noise at same dose. The iodine enhancement ratio therefore gives a good impression of the scanner performance with no need for more complicated mathematics.

For the specific scanner type in our phantom study, the use of a tin filter allows for less noise at same dose in the mixed image as well as in the VNC image. The voltage combination 70/Sn150kV is apparently the best for patient diameters of up to 30cm, since much higher iodine contrast is observed, while dose and image noise are comparable with 80/Sn150kV. At large diameters (40cm), 90/Sn150kV and 100/Sn150kV achieve visibly the best image quality.
References

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