PHILIPS

Computed Tomography

White paper

Advantages of a spectral detector-based CT system

Introduction

CT is routinely used in a diagnostic radiology department to help radiologists with a variety of disease conditions. Conventional CT uses Hounsfield units (HU) as a mechanism to differentiate between different tissues within the body. As a result, many different tissues with different chemical composition (or effective atomic number) and different densities may have overlapping HU values at a given peak tube voltage (kVp), making it difficult to differentiate between tissues. The two main mechanisms contributing to CT attenuation coefficients of materials are the photoelectric effect and Compton scattering. These effects are both energy- and material-dependent. The photoelectric effect predominates at lower photon energies and is heavily energy-dependent, while Compton scattering has a much weaker energy dependency than the photoelectric effect and dominates at high photon energies. The relative strength of photoelectric and Compton components also depend on atomic numbers. The photoelectric effect dominates the overall attenuation of high atomic number atoms such as iodine, while Compton dominates the attenuation of low atomic number atoms such as water (hydrogen and oxygen). Spectral CT measures the difference in attenuation of X-rays at two energy levels, high and low. Data collected simultaneously from these two energy levels can be used to determine the Compton scatter and photoelectric components of X-ray attenuation. These components, taken together, provide additional information about tissue density and atomic number that can be used to separate tissues with similar attenuation in a conventional image.

Approaches to spectral image generation

The two primary modes of generating spectral images are source-based and detector-based.

Source-based imaging

Source-based approaches use two different X-ray spectra (high and low energy), from two sources, or from a single source in a modulated fashion, to generate the two energy levels needed to create a spectral scan. Source-based methods include dual-source, kV-switching, twin beam and dual-spin.

- Dual-source CT uses two separate tubes integrated in the same scanner and operated at two different tube voltages.
- Fast kV-switching involves switching single tube potential to acquire interleaved data at two different tube voltages.
- Slow kV-switching involves switching the tube potential to acquire data at two different tube voltages with different kV settings per rotation.
- Twin beam shares the X-ray beam in the scanner with two different filters to generate two different spectra.
- Dual-spiral or dual-spin CT acquires two successive spiral or axial scans at different kVp and mA levels from a single source.

All source-based techniques require the clinician to preselect patients for dual-energy scanning. Because the acquisition of spectral data is dependent on the X-ray tube, certain trade-offs have to be made with respect to special modes, image quality and dose penalties. Some examples are slower rotation speed and reduced spatial resolution in fast kV-switching mode; longer scan time (low pitch) in slow kV- switching, in twin beam modes and in dual-source scanners (fast, high pitch scan); and giving up temporal resolution (1/4 rotation acquisition with two tube-detector systems) and field of view in dual-source scanners.

Detector-based spectral imaging

Unlike source-based spectral options, detector-based approaches simultaneously absorb and differentiate high and low energy, available in a single polyenergetic X-ray beam, at the detector level. Spectral results are acquired within a single scan without the need for special modes.

There are several advantages to detector-based spectral imaging.

- The clinician can adhere to the familiar conventional workflow, while retaining the same dose setting and usage of dose management tools, as well as the same rotation speed and pitch setting.
- The patient is scanned as usual and a true conventional image is generated, which is identical to conventional CT scanners.
- Full spectral information can be generated in addition to the true conventional images. The clinician only needs to select the generation of spectral information in addition to the conventional data. A retrospective reconstruction of the spectral information is also possible in case spectral data were not requested in the original reconstruction. Spectral is always on and always available.

Comparison of source-based and detector-based spectral imaging

Features in spectral mode	Dual-source (DSCT)	Fast kV-switching (KVSCT)	Slow kV-switching	Twin-beam	Dual-spiral or dual-spin	Detector-based spectral (DBSCT)
How it works	Special spectral mode must be preselected: 2 tubes 70-100/140-150 kVp	Special spectral mode must be preselected: fast kV-switching 80/140 kV from one projection to the next	Special spectral mode must be preselected: slow kV-switching 80/140 kVp from one rotation to the next	Special spectral mode must be preselected: X-ray beam split in 2 along Z axis with 2 different filtrations for low and high energy	Special spectral mode must be preselected: 1st spin at 80 kVp 2nd spin at 140 kVp	**No special spectral mode required:** **spectral is always on** **Scan at 120 kVp or 140 kVp** **dose-neutral**
Spectral mode	Prospective	Prospective	Prospective	Prospective	Prospective	**Prospective and retrospective**
Acquisition mode	Dual-energy	Dual-energy	Dual-energy	Dual-energy	Dual-energy	**Conventional**
Consistent data sets	Quarter rotation offset	Nearly perfect alignment	One rotation offset	Half collimation offset	Very large offset	**Perfect alignment**
Spectral decomposition	Image space	Projection space	Image space	Image space	Image space	**Projection space**
120 kVp conventional image	No	No	No	No	No	**Yes**
Dose modulation available	Yes	No	Yes	Yes	Yes	**Yes; tube mA modulation**
FOV	Limited	Full	Full	Full	Full	**Full**
Temporal resolution (for cardiac scanning)	Reduced to same as single-source scanner with same rotation speed as in spectral mode; information for the 2 detectors cannot be combined to build a single image with 2 times better temporal resolution	Reduced as fastest rotation speed not available with spectral mode	Not available for cardiac scanning	Not available for cardiac scanning	Not available for cardiac scanning	**Full; no limitations**

Comparison of source-based and detector-based spectral imaging						
Features in spectral mode	**Dual-source (DSCT)**	**Fast kV-switching (KVSCT)**	**Slow kV-switching**	**Twin-beam**	**Dual-spiral or dual-spin**	**Detector-based spectral (DBSCT)**
Scan time	Increased to same as single-source – ultra high pitch scanning combining the 2 helix from the 2 detectors not possible in spectral mode as with different tube voltage – or even more to get enough image quality on low energy acquisition	Increased as fastest rotation speed not available with spectral mode	At least 2 times more than conventional	At least 2 times more than conventional	At least 2 times more than conventional	**Same as conventional**
Obese patient	Limited due to low quality for kVp acquisition and reduced energy separation if using higher kVp for low energy (100/150Sn)	Very limited due to low quality for 80 kVp acquisition	Very limited due to low quality for 80 kVp acquisition	Very limited due to low loss power in tube filtrations	Very limited due to low quality for 80 kVp acquisition	**No limitations**

Table 1. A comparison of detector-based and source-based spectral approaches.

Advantages of a detector-based system

Radiation dose

The detector-based approach of the Philips IQon Spectral CT system allows the user full access to all dose management tools available in the conventional scanning mode, at all times. Even when collecting spectral information, the IQon Spectral CT adheres to the Philips DoseWise approach, which is a set of techniques, programs and practices based on the ALARA (as Low As Reasonably Achievable) principle to support outstanding image quality at low dose. During scanning, tube current modulation is used to change the X-ray dose during the rotation and along the scanned body, based on body regions. Image quality for each diagnostic task is specified by the DoseRight Index (DRI) for various scanning regions, to allow for appropriate dose to maintain consistent image quality within a single acquisition and between patients.

- Personalized doses for individual patients are suggested by the DoseRight automatic current selection.
- Longitudinal dose modulation is achieved using the DoseRight Z-DOM, which adjusts the tube current-time product (mAs) in the z-axis according to a patient's size and shape.
- DoseRight 3D-DOM (three-dimensional dose modulation) combines angular and longitudinal patient information to modulate dose in three dimensions (x-y-z-axis). It incorporates modulation of tube current-time product (mAs) according to changes in individual patient's size

and shape in the transverse (x-y-axis; angular) direction during helical scans, in addition to changes in the craniocaudal or caudocranial (z-axis; longitudinal) direction, as the tube rotates.

- Liver DRI is organ-based dose modulation that allows setting a different target image quality on the liver area than on the rest of the scan. During the acquisition, mAs value is modulated to account for the variations in patient body attenuation as with standard AEC but with different target image quality in the liver area than in the rest of the scan, such as chest or pelvis. The liver is automatically detected on the surview and better image quality is required for the assessment of the liver due to its lower contrast.
- Brain DRI is the same as liver DRI, but for head and neck scans.
- Dedicated pediatric protocols offer high-quality conventional images at low doses that take into account the pediatric patient's size and clinical indication. Note that to benefit from the spectral result added value, you need to set your protocols to 120 kVp and reduced mAs for the same CTDI if the initial protocol was with lower kVp. This can be done for most protocols to the limit of 0.9 mGy, corresponding to the lowest value available with 120 kVp. Lower CTDI protocols can still be used as on conventional scans but will not benefit from spectral information.
- Conventional images can use the full range of reconstruction options such as iDose[4] and iterative reconstruction IMR, as in conventional (single-energy) CT scanners. IMR can also be used for reduced patient dose while keeping the desired image quality.*
- Similarly, the spectral results can use noise suppression strength, named "Spectral-level." The resulting noise suppression of a specific spectral level is rather similar to the impact of iDose[4] at same level.

Spectral is always on

IQon Spectral CT is a detection-based spectral CT; therefore, spectral information is available for each and every scan (at 120 or 140 kVp). There are three methods to generate the spectral data.

1. Generate DICOMs of the spectral results that are planned to be used, for example monoenergetic (MonoE) results as a part of exam card on the scanner. These results will be sent to PACS, similarly to conventional images and can be used by any PACS and any viewer. Exam cards can include the type of spectral results to be generated.
2. Save SBI (spectral base images) along with the conventional images. This generates a compressed DICOM series from which all spectral results can be generated. SBI can be read by the IntelliSpace Portal thin client application and by the Spectral Magic Glass on PACS (sMGoP). A checkbox in the recon menu (and in the exam card setting) activates SBI saving.
3. Retrospective spectral reconstruction in case no spectral information was generated during the original reconstruction but needed later on, and can be used as long as the raw data of the case is available.

The recommended mode is a combination of (1) and (2) for the relevant reconstructions. Send the spectral results that are in routine use (per scan type, etc.) as regular DICOM series, and generate SBI to enable other spectral results that may be retrospectively required.

*In clinical practice, the use of IMR may reduce CT patient dose depending on the clinical task, patient size, anatomical location, and clinical practice. A consultation with a radiologist and a physicist should be made to determine the appropriate dose to obtain diagnostic image quality for the particular clinical task. Lower image noise, improved spatial resolution, improved low-contrast detectability, and/or dose reduction, were tested using reference body protocols. All metrics were tested on phantoms. Dose reduction assessments were performed using 0.8 mm slices, and tested on the MITA CT IQ Phantom (CCT183, The Phantom Laboratory), using human observers. Data on file.

Spectral image quality

Energy separation and spectral separation

Spectral performance of the scanners determines the quality of spectral results. The energy separation between the high- and the low-energy spectrums as well as the noise level within the detected signal from each spectrum determine the spectral performance.

The energy separation is the difference in effective energies between the two measured attenuations that will allow spectral analysis. In principle, the lower energy spectrum measures the low-energy attenuation, dominated by the photoelectric effect, and the higher energy spectrum measures the high-energy attenuation, dominated by the Compton scattering effect. The choice of the low- and high-energy spectrums is limited by the available spectra from the X-ray tube. For example, the 80 kVp spectrum overlaps the 140 kVp spectrum. The difference in average or effective energies depends on the tube filtration as well as on patient size. A dual-source scanner allows using 70 kVp for the low-energy spectrum and up to 150 kVp with specific filtration to increase the average energy of the high-energy spectrum. The problem of 70 kVp is that for average and large patients, most of the X-rays are absorbed by the patient, irradiating him or her and resulting in poor signal on the detector. Higher energy spectrums from 80, 90 or even 100 kVp are then needed, reducing the energy separation.

However, energy separation is not the only parameter that drives the quality of the spectral results. Signal-to-noise ratio (SNR) determines the ability to use the dual-energy data and derive quality spectral results. In the case of dual-energy spectral analysis, the signal is the difference in attenuation between the low- and high-energy spectrums. The noise is the square root of the sum of the variances. Therefore, the signal with the higher noise dominates the SNR. Using low kVp for the low-energy spectrum usually leads to high noise, impacting the overall SNR. Another noise source, only relevant for the dual-source DECT, is cross-scatter. The dominant effect is radiation from tube-1 that is scattered from the surface of the scanned body and captured by the detection system of tube-2 (and vice versa). The detection system captures both the desired attenuated radiation from its dedicated X-ray source (primary signal), and the un-attenuated scattered radiation from the other X-ray source. Proper algorithms can estimate and subtract the scattered radiation from the primary signal, however, the noise of the scattered radiation is added to the noise of the primary signal.

In dual-layer detector-based spectral CT, the low- and high-energy spectrums have more overlap compared with source-based DECT. Therefore, the energy separation is lower than the separation of dual-source CT. However, the noise of the low- and high-energy spectrums is comparable, and there is no cross-scatter noise. In addition, the low- and high-energy data are simultaneously acquired (perfectly aligned in space and time) enabling projection domain spectral analysis which has inherent advantages over image domain analysis (Figure 1). A publication from Maaß et al. demonstrates that material decomposition is inherently better in projection domain as compared to image domain.[1]

IQon Spectral CT spectral decomposition and reconstruction include noise suppression methods based on proper noise modeling and specific iterative techniques resulting with high SNR. This results in high-quality spectral results as demonstrated by the comparison studies published by Sellerer et al.[2], or Hua et al.[3] in comparison to Almeida et al.[4]

Spatial and temporal coherence and projection based versus image based spectral results

Detector-based spectral imaging has the advantage of simultaneously acquired (spatially and temporally) measurements of high- and low-energy projection data sets, across the two detector layers. This enables spectral decomposition at the projection domain, which has an inherent advantage over image-domain decomposition. The noise of the coherent spectral data is almost purely quantum (Poisson) noise. The noise characteristics, including spectral anti-correlative noise, are fed into model based iterative methods, which effectively suppress image noise of all spectral results. One of the unique advantages of this approach is low image noise at low keV monoenergetic images and hence CNR keeps improving as MonoE energy is reduced.

Low noise virtual monoenergetic image (VMI) for improved CNR

The advantage of spectral detector CT in terms of low image noise over the whole range of VMI was further demonstrated in an independent study, conducted by Sellerer et al.[2] comparing the different spectral image generation approaches.

In this study, a semi-anthropomorphic abdomen phantom was imaged using a spectral detector CT (IQon), a dual-source dual-energy system (DSCT), and a fast kVp-switching dual-energy system (KVSCT).

Scans were acquired for three patient sizes at $CTDI_{vol}$ levels of 10 mGy, 20 mGy and 30 mGy. Noise and quantitative accuracy of HU and iodine concentration were evaluated for a range of monoenergetic images. The results of the study demonstrated important differences among approaches.

- A steep increase in image noise was observed for keV levels below 80 keV for fast kVp-switching and dual-source approaches, while the change in noise was rather low in the case of detector-based approach (Figure 2).
- Virtual monoenergetic image quality was comparable for detector-based and dual-source based approaches for spectral imaging at mid- to high-energy levels, and image quality was significantly reduced for the fast kVp-switching approach.

Why such good results despite limited energy separation?

Spectral separation versus energy separation

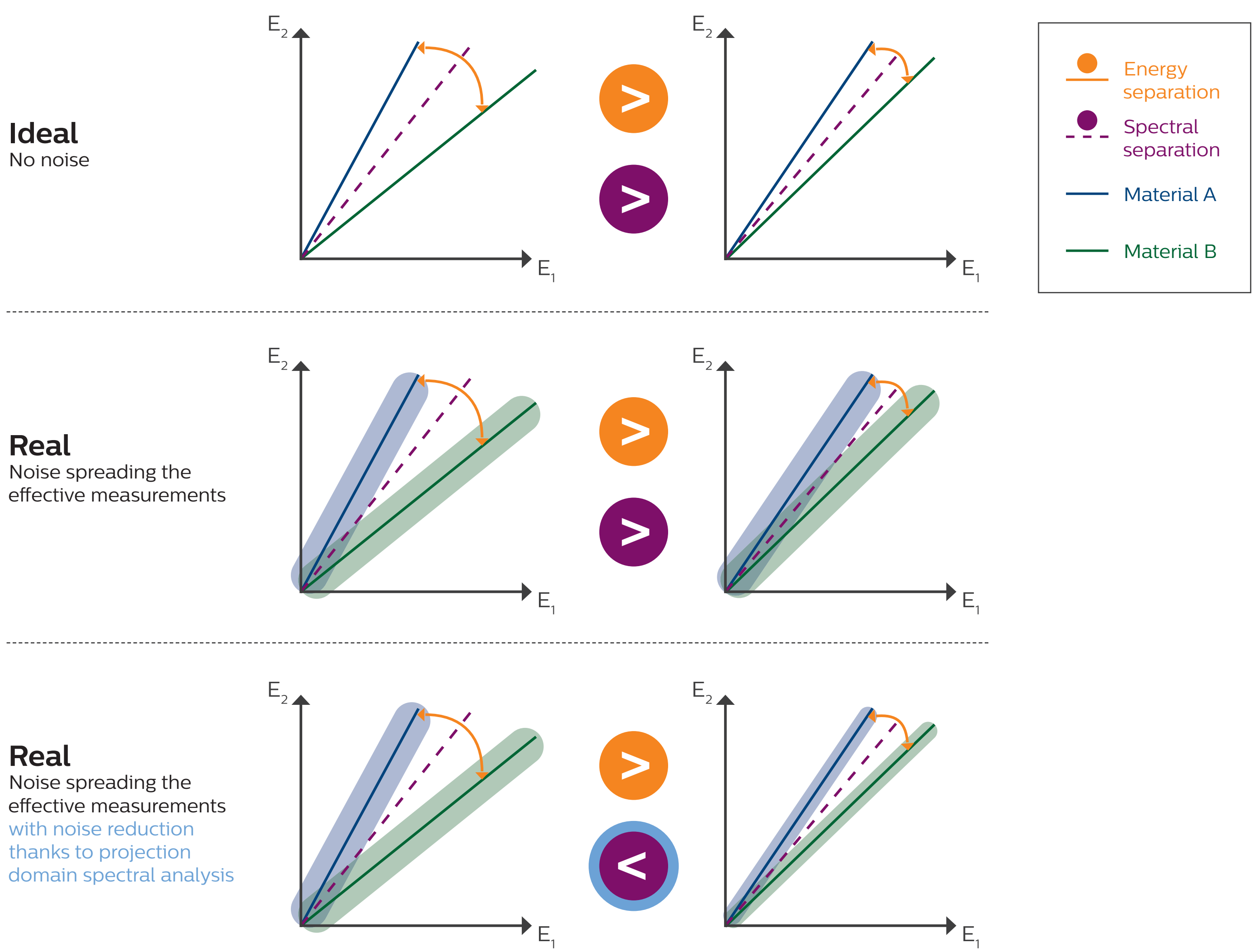
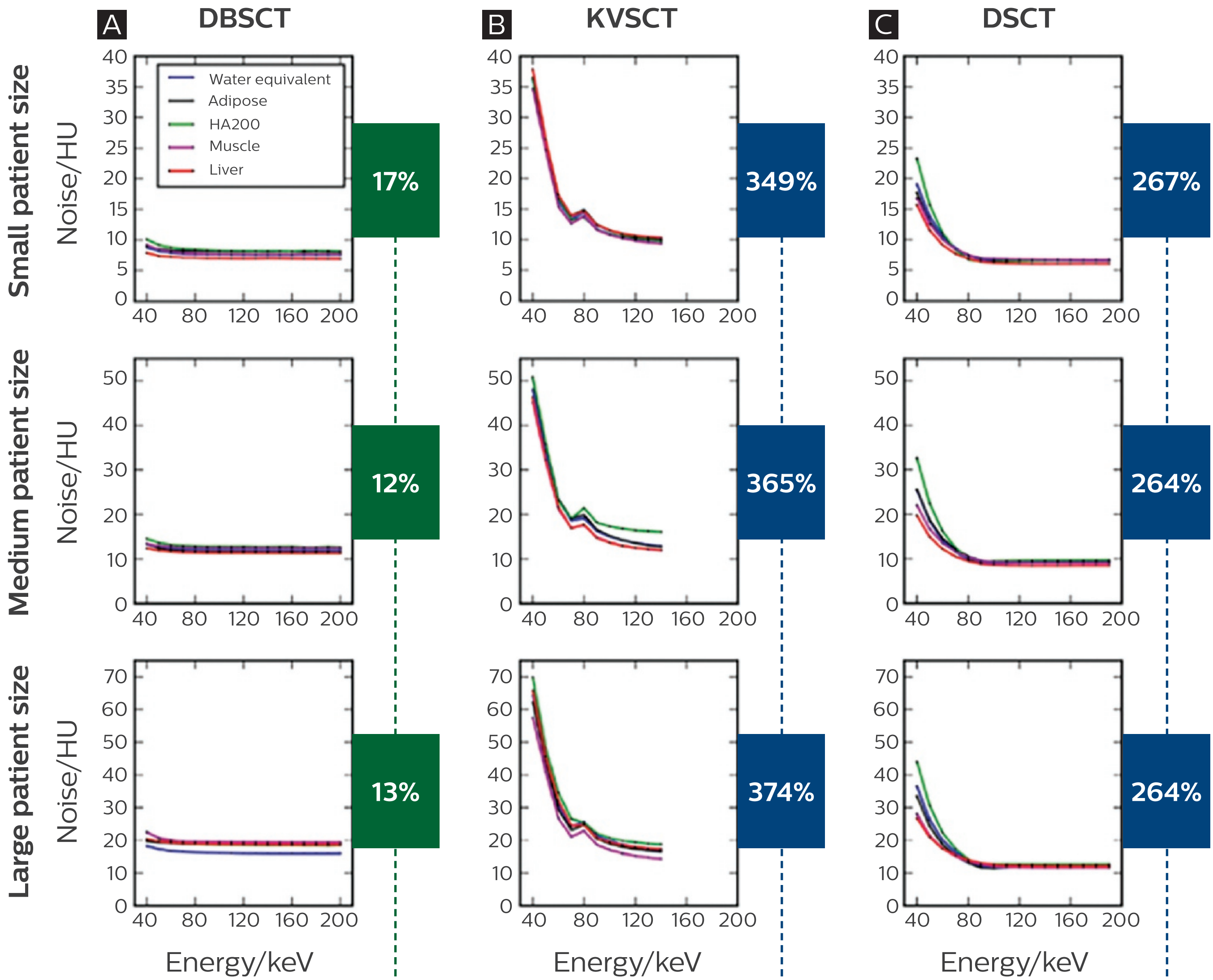


Figure 1. Spectral separation and noise.

Low noise virtual monoenergetic images

Percentage of noise increase from 140 keV to 40 keV for different patient sizes at 20 mGy $CTDI_{vol}$ (averaged over all available tissue equivalents listed).



Mean value percentage of noise increase 14% 363% 265%

Figure 2. Noise comparison across different source- and detector-based dual-energy/spectral scanners.

Dual-energy CT: a phantom comparison of different platforms for abdominal imaging. T Sellerer, P B Noël, M Patino, A Parakh, S Ehn, S Zeiter, J A Holtz, J Hammel, A A Fingerle, F Pfeiffer, D Maintz, E J Rummeny, D Münzel, D V Sahani; Eur Radiol. 2017

Another study by van Hamersvelt et al.[5] demonstrated that virtual monoenergetic images (VMI) from fast kVp-switching, dual-source and spectral detector CT techniques exhibit a strong increase in iodine attenuation at low energies (Figure 3) as expected from the iodine attenuation curve. When looking at CNR values, with kVp switching, the maximum CNR was for 70 keV and was lower than with 80 kVp conventional images; with the dual-source technique, CNR increased when going to low VMI, but the increase was less than expected, and spectral detector CT showed a strong CNR increase when going to low VMI (Figure 4).

Tsang et al.[6] confirmed in a clinical study that CNR from 50 keV VMI from spectral detector CT was two times more than CNR at 120 kVp, and CNR at 40 keV VMI was three times the CNR at 120 kVp. This CNR increase is very important for many clinical applications with iodinated contrast enhancement to improve lesion and vessel assessment.

Improved iodine enhancement at low keV

Low noise, low energy virtual MonoE images with IQon Spectral CT

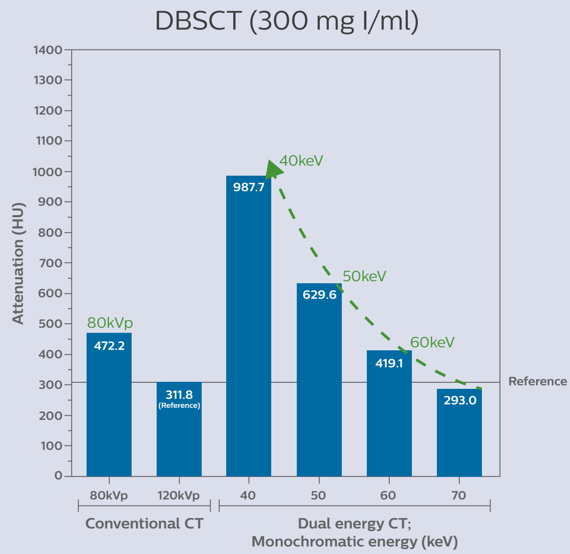
51 pediatric patients received contrast-enhanced CT for radiation therapy planning.

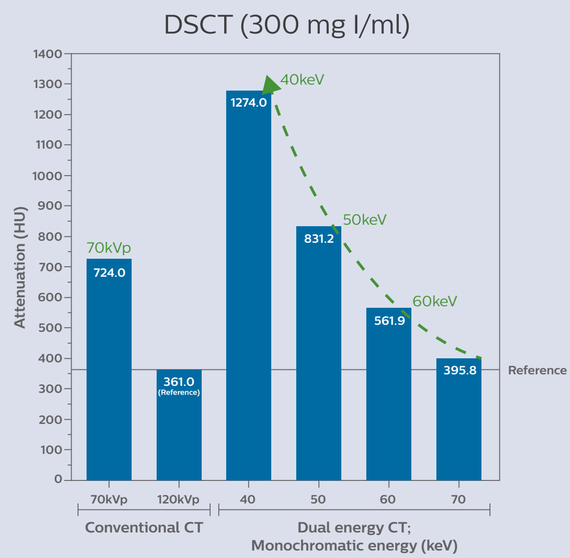
- **2.4-fold mean CNR increase at 50 keV compared to 120 kV**
- **3.3-fold mean CNR increase at 40 keV compared to 120 kV**

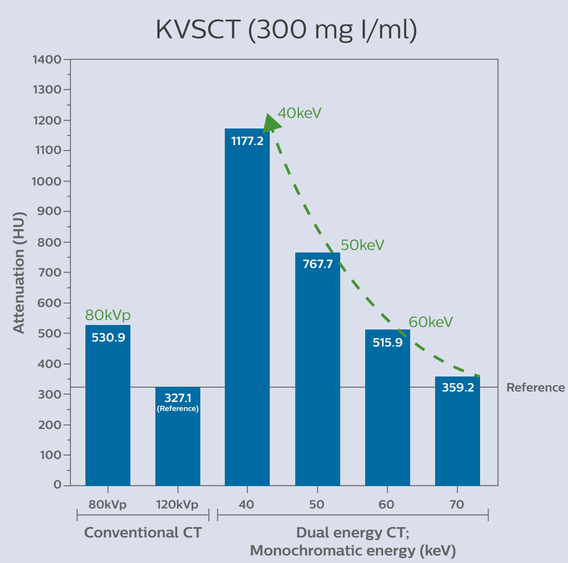
(Right) Contrast agent concentration optimization in CTA using low tube voltage and dual-energy CT in multiple vendors: a phantom study. R. van Hamersvelt et al., The International Journal of Cardiovascular Imaging, 2018. https://doi.org/10.1007/s10554-018-1329-x

Figure 3. Attenuation comparisons for different scanners at low kVps and low MonoEs.

Improved iodine enhancement at low keV – attenuation (HU)

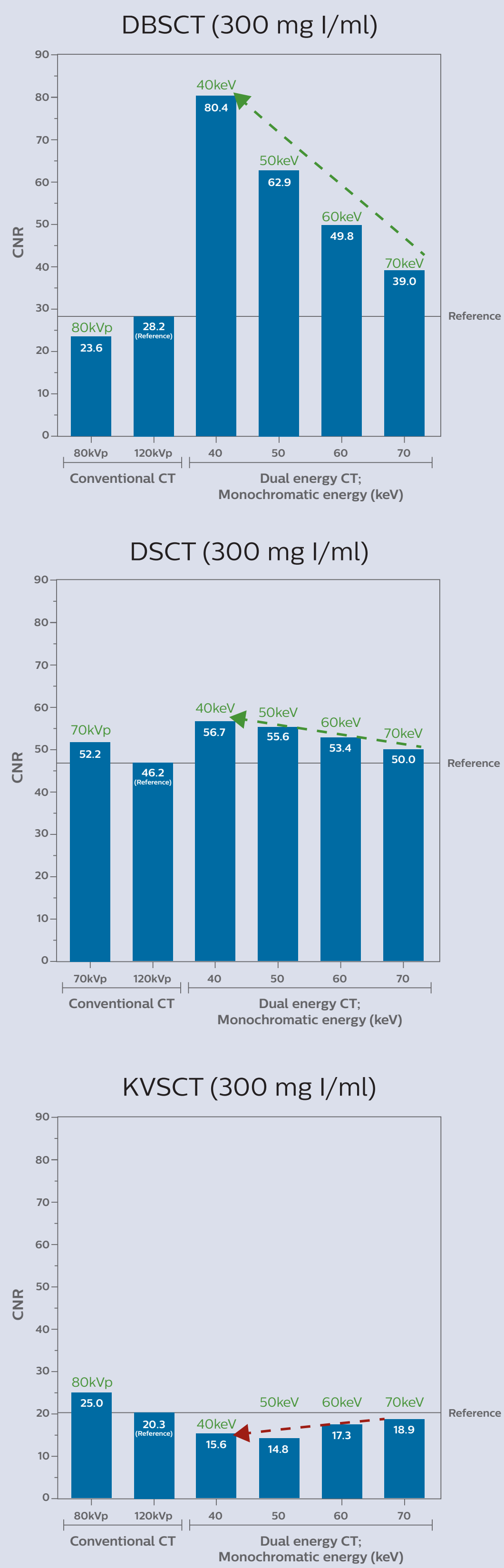






Improved iodine enhancement at low keV – CNR

IQon low noise MonoE images translate the benefit at low energy in strong CNR increase.



Moving to true quantitative imaging

Publication by Sellerer et al.[2] showed that in abdomen phantom studies, detector-based and fast kVp-switching approaches deliver comparable absolute errors of measured iodine concentration. The mean errors in source-based spectral CT are notably larger. The root mean standard deviation (RMSD) of iodine concentrations was lowest for the detector-based approach in eight of the nine patient size/dose configurations tested, and comparable to the kVp-switching approach of spectral imaging in the ninth configuration. All three platforms show very good correlation between measured and true iodine concentration. However, IQon shows the lowest error range of 0.42 compared to 1.15 for KVSCT and 1.88 for DSCT. Detector-based spectral CT is superb compared to the other techniques for large patient size phantom and low dose (Figure 5.)

(Left) Contrast agent concentration optimization in CTA using low tube voltage and dual-energy CT in multiple vendors: a phantom study. R. van Hamersvelt et al., The International Journal of Cardiovascular Imaging, 2018. https://doi.org/10.1007/s10554-018-1329-x

Figure 4. CNR comparisons for different scanners at low kVps and low MonoEs.

Moving to true quantitative imaging

Quantitative iodine perfusion

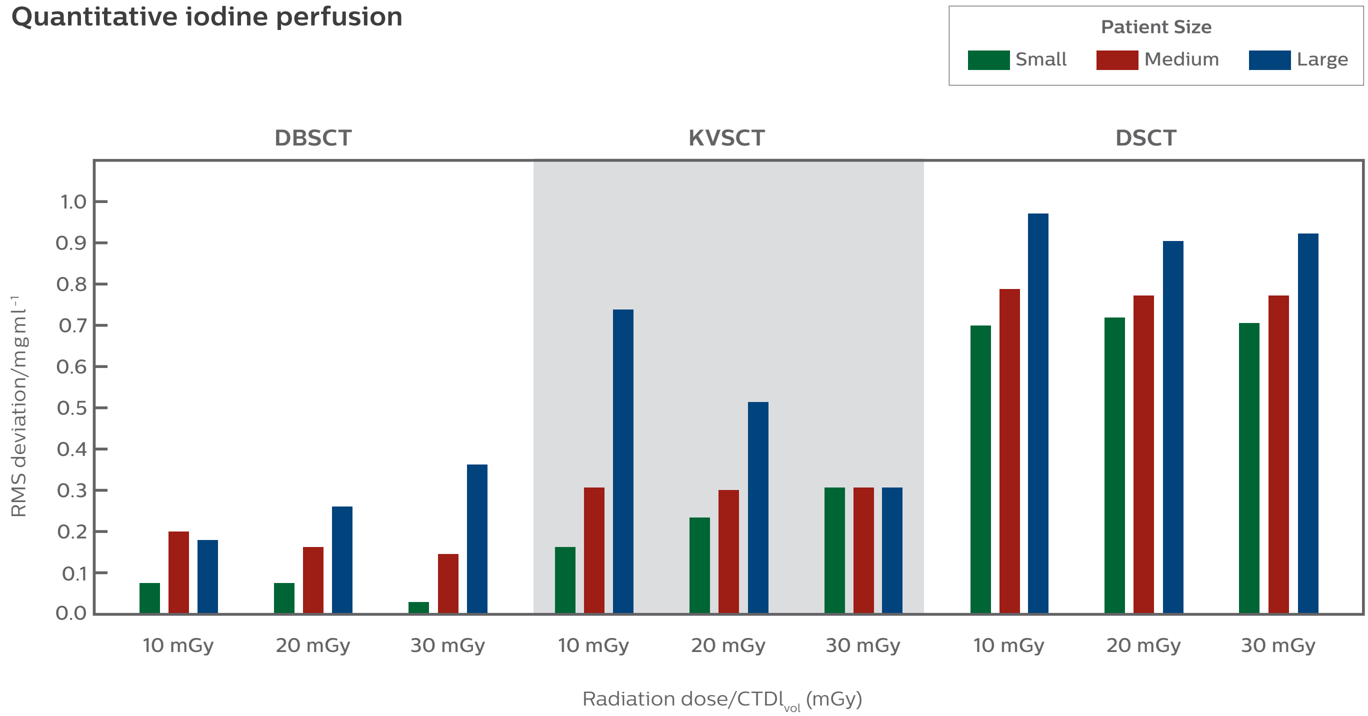


Figure 5. Root mean square deviation (RMSD) of observed iodine concentrations (with respect to true values) for different measurement configurations and CT systems. Deviations are given in mg/ml.

Temporal resolution

The temporal resolution on IQon Spectral CT is as low as 34 mSec using adaptive multicycle reconstruction, and since there is not a spectral mode to acquire spectral data, there is no compromise in temporal resolution for IQon Spectral CT. This is not the case for the source-based dual-energy scanners where temporal resolution in dual-energy mode is compromised as compared to conventional scans, which could be due to slowing down the scanner for kVp switching or not having high pitch mode or flash mode of scanning due to two tubes operating at two different kVps.

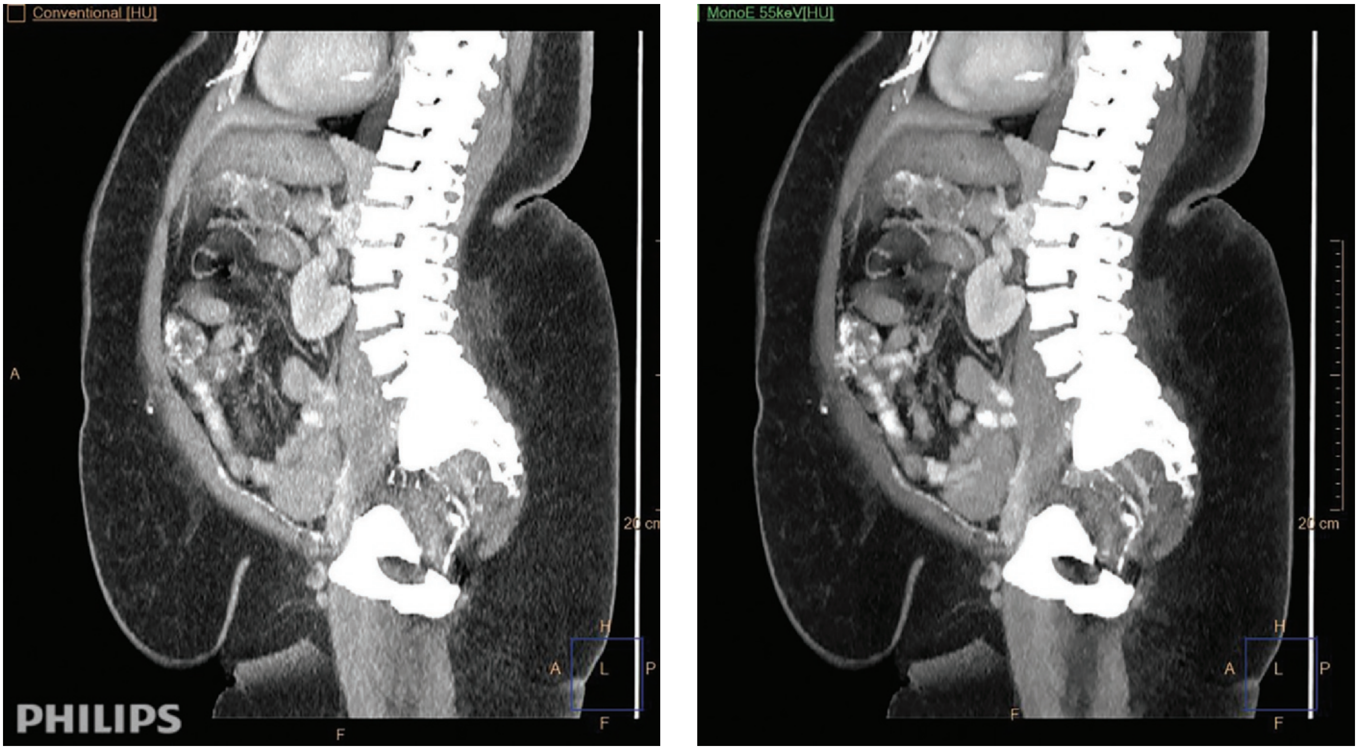
Scan time

Scan times on IQon Spectral CT are the same whether you are acquiring conventional data or spectral data. There is no special scan mode for spectral, and therefore, the scan parameters (including the scan time) are the same.

As for source-based systems, the scan time is longer when the system is operating in dual-energy mode as compared to conventional scans. This change in scan time can be attributed to several acquisition parameters that need to be changed when the system operates in dual-energy mode e.g., rotation time has to increase, and pitch has to be decreased in order to obtain dual-energy data.

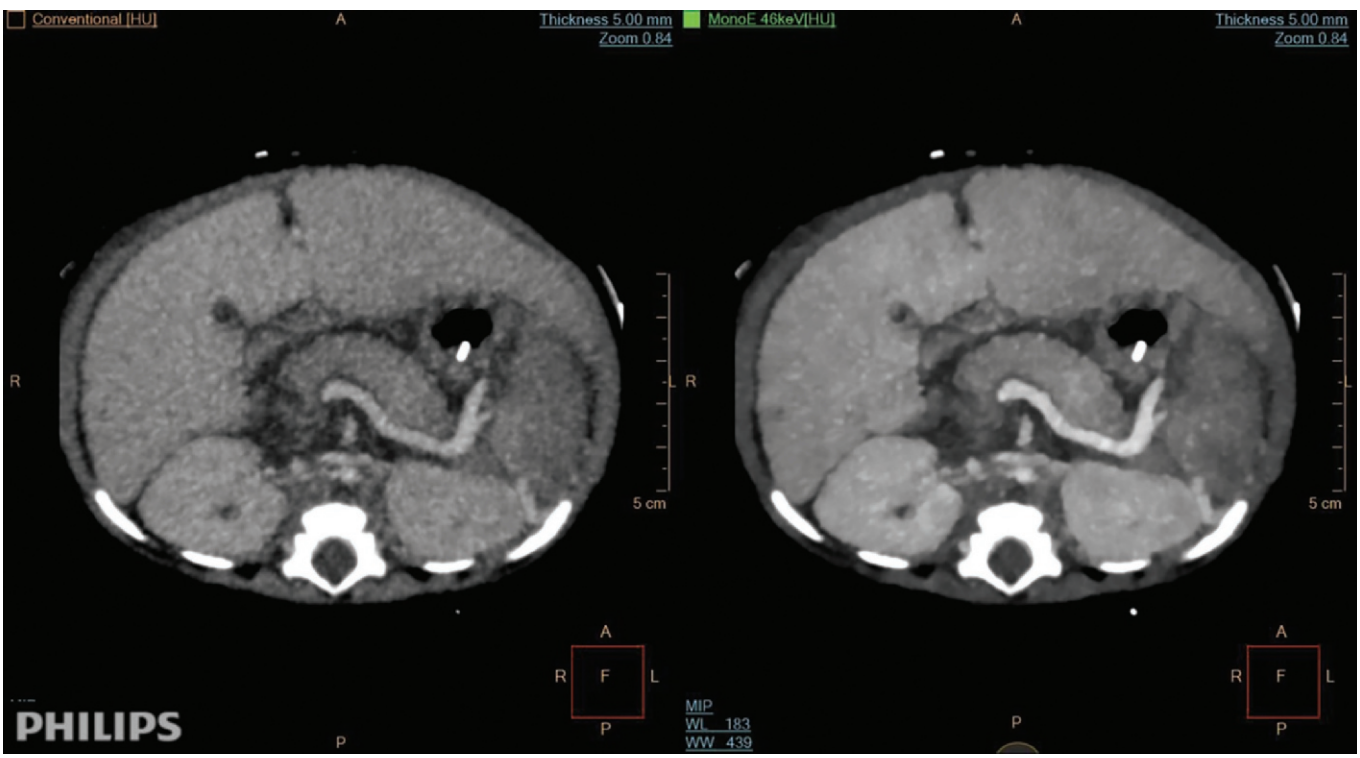
True conventional images

Because IQon Spectral CT is a detection-based spectral CT system, it generates true, uncompromised, conventional images that are essentially identical to single-energy CT. One simply adds the data from the low- and high-energy layers into a combined signal similar to a conventional (single-energy) detector. This is a unique capability of the dual-layer spectral CT scanner. Moreover, spectral information is enabled for scans at 120 kVp, which is the tube voltage that is used in the majority of the scans. Source-based dual-energy systems scan at two different kVps (70 kVp to 150 kVp), and they then create a 120 kVp-like image using image reconstruction and processing techniques or only provide the low- and high-energy images.



Conventional image | **MonoE 55 keV image**

Figure 6. Spectral detector CT of an abdomen and pelvis (68 BMI). *Courtesy of LSU Health Sciences Center.*



Conventional image | **MonoE 46 keV image**

Figure 7. Spectral detector CT of a 6-month-old pediatric abdomen. *Courtesy of Phoenix Children's Hospital.*

Patient size

IQon Spectral CT can scan the entire range of patient populations from pediatric to bariatric without any limitations or compromises.

Bariatric

There are several limitations when imaging obese patients using conventional CT scanners. The tradeoff between achieving high image quality and maintaining radiation dose is a challenge for radiologists. Early introduction of dual-energy CT scanners did not address these obstacles, since source-based solutions (kVp switching, dual-source and dual-spin scanners) rely on acquiring the images at high and low kVp settings or dual-tube filtration. For dual-source systems, the limited FOV on the smaller detector exclusively affects imaging of obese patients. For a rapid kVp-switching system, high and low voltages have to be rapidly alternated. Therefore, current cannot be correspondingly alternated to provide adequate numbers of photons at low energy, resulting in higher current settings and higher radiation dose. Additionally, automated current modulation algorithms cannot be employed, which leads to increased radiation exposure. As a result, these systems limit imaging to patients with BMI less than 30 kg/m^2, or weighing less than 260 lbs. For dual-helical, dual-spin or slow kV-switching, mAs can be adapted to each kVp, but low kVp still suffers from very high attenuation and low image quality with obese patients. In twin-beam technique, the tube filtration used to create two different spectrums reduces the maximum X-ray flux available, and the low energy part of the beam also suffers from high attenuation.

With the introduction of the IQon Spectral CT system, signal separation happens at the detector level, which allows the user to scan patients with typical peak voltages of 120 kVp or 140 kVp. IQon Spectral CT allows scanning for patients who weigh over 260 lbs., making the spectral benefits available to the obese patient population.

In a study published by Atwi et al. in Abdominal Radiology[7], researchers evaluated the image quality of spectral results in patients weighing >270 lbs. using IQon Spectral CT. They demonstrated that radiologists preferred the image quality of MonoE 70 keV results as compared to conventional CT. Noise was significantly lower; SNR and CNR were significantly higher in MonoE 70 images as compared to conventional images. Evaluation of iodine-based results showed that iodine maps were homogenous and complete in all abdominal organs of interest. The researchers concluded that IQon Spectral CT provides viable, high-quality imaging for patients weighing >270 lbs. (bariatric patients).

Pediatric

With the IQon Spectral CT scanner, the user can scan at 80-140 kVp, in order to obtain conventional images. The user has to scan at 120 kVp or 140 kVp to obtain spectral data. Since IQon Spectral CT is a detector-based spectral CT scanner, all the dose management tools are available.

Sites that use low kVp settings such as 80 kVp or 100 kVp can keep their current practice and use the true conventional images (as in a single-energy CT), or move to 120 kVp and adjust the mAs to match the CTDI of 80 kVp or 100 kVp protocol. The latter method enables full spectral results. In particular, MonoE images at low keV have superb image quality compared with conventional low kVp images. Pediatric patients can therefore be scanned in spectral mode without any impact on radiation dose.

Researchers of Phoenix Children's Hospital[8] published a study comparing the radiation dose and image quality for pediatric CT body protocols between IQon Spectral CT and a conventional CT scanner (Brilliance iCT). A phantom was scanned on both the scanners using standard body protocols that were matched to obtain the same $CTDI_{vol}$. The kVps on the conventional scanner ranged from 80-120 kVp. Radiation dose and CNR were compared between IQon Spectral CT and the conventional scanner. CNR was higher for IQon Spectral CT scans, showing an improvement in image quality without any significant increase in radiation dose.

Spectral suite (spectral CT viewer and spectral applications, Magic Glass on PACS)

Spectral CT Viewer

The Philips Spectral CT Viewer is designed to enable spectral quantification through proprietary spectral tools. By offering unique capabilities across clinical areas, spectral applications provide additional anatomical and functional information to enhance diagnostic confidence.

Spectral Magic Glass

In addition to conventional CT images, Philips Spectral Magic Glass enables on-demand simultaneous viewing and quick comparison of up to five different spectral results for a region of interest, including monoenergetic, iodine density, virtual non-contrast, Iodine no Water and Z effective maps. The Spectral Magic Glass tool is superimposed on CT images to provide a color view of an area of special interest. Materials such as iodine, calcium, water or fat can then be visually distinguished.

Spectral Magic Glass on PACS

The Spectral Magic Glass on PACS application conveniently launches directly in the user's PACS viewing setup, offering a simple interface that integrates into hospital workflow with little or no training required.

Spectral Advanced Vessel Analysis (sAVA)

The Spectral Advanced Vessel Analysis application offers a set of tools for general vascular analysis. It allows the user to remove bone, extract and edit vessel wall and lumen based on spectral data, perform lesion analysis based on spectral data and compare the extracted vessels using various spectral results.

Highlights of the application:

- Bone removal on different energy levels
- Reduced calcified plaque artifacts
- Comparison of different energy results

Spectral-enhanced Comprehensive Cardiac Analysis (sCCA)

The Spectral Comprehensive Cardiac Analysis provides the ability to run on-demand cardiac segmentation on different energy levels, compare vessel curves with various spectral data types and enhance the visual assessment of coronary vessel patency.

Highlights of the application:

- Automatic chamber and coronary segmentation using monoenergetic images
- Beam hardening reduction for improved visualization of perfusion deficits and calcified plaque visualization

Spectral-enhanced Multi-Modality Tumor Tracking (sMMTT)

This application provides tools to help clinicians monitor disease progression or assessment of therapy response.

Highlights of the application:

- Tumor viewing with different spectral data types (VNC, iodine map)
- Images at different energy levels (40-200 keV)
- Iodine uptake measurements

Future: Photon counting spectral CT as a possible next step for detector-based spectral CT

Photon counting spectral CT seems to be the next generation of high-end CT systems. There is a wide effort in academia and in the industry to advance this technology. Several prototypes are available and have demonstrated promising results on phantoms, animals and even humans. Advantages include ultra-high spatial resolution combined with high-quality spectral results and low-dose capabilities. Photon counting enables unique k-edge imaging capabilities on top of improved spectral performance of dual-energy-like applications. While prototypes do exist, as listed above, there are quite a few challenges that need to be addressed before commercial systems for general clinical use will be available in the market.

Note, however, that photon counting is a detection-based spectral technology and has the same advantages as dual-layer spectral CT - full spatial and temporal coherence of spectral data, projection domain spectral analysis and always spectral, always on. The future is therefore in detection-based spectral CT.

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