Review paper

Principles and applications of dual source CT

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ABSTRACT

This article describes the technical principles and clinical applications of dual source CT. A dual source CT (DSCT) is a CT system with two x-ray tubes and two detectors at an angle of approximately 90°. Both measurement systems acquire CT scan data simultaneously at the same anatomical level of the patient (same z-position). DSCT provides temporal resolution of approximately a quarter of the gantry rotation time for cardiac, cardio-thoracic and pediatric imaging. Successful imaging of the heart and the coronary arteries at high and variable heart rates has been demonstrated. DSCT systems can be operated at twice the spiral pitch of single source CT systems (up to pitch 3.2). The resulting high table speed is beneficial for pediatric applications and fast CT angiographic scans, e. g. of the aorta or the extremities. Operating both X-ray tubes at different tube potential (kV) enables the acquisition of dual energy data and the corresponding applications such as monoenergetic imaging and computation of material maps. Spectral separation can be improved by different filtration of the X-ray beams of both X-ray tubes. As a downside, DSCT systems have to cope with some challenges, among them the limited size of the second measurement system, and cross-scattered radiation.

1. Technical principles and setup of dual source CT systems

Cardiac imaging with computed tomography requires short exposure times of the axial slices and the corresponding dedicated scan and image reconstruction techniques to avoid motion-artifacts in the images. Increased gantry rotation speed is a prerequisite for clinically robust improvements of the temporal resolution with 3rd generation multi-detector row CT (MDCT) systems. An alternative scanner concept that provides considerably enhanced temporal resolution but does not require faster gantry rotation is a CT with multiple X-ray tubes and corresponding detectors [1,2].

In 2006, a dual-source CT (DSCT) was commercially introduced. It was equipped with two x-ray tubes and two detectors (see Fig. 1). Both measurement systems operate simultaneously and acquire CT scan data at the same anatomical level of the patient. Since then, three generations of DSCT systems have been commercially introduced. The 1st DSCT, the SOMATOM Definition (Siemens Healthineers, Forchheim, Germany) [3], has two measurement systems mounted onto the rotating gantry with an angular offset of exactly 90°. Detector (A) covers the full 50 cm diameter scan field of view (SFOV), while detector (B) covers a smaller 26 cm SFOV as a consequence of space limitations on the gantry. Both detectors simultaneously acquire 64 overlapping 0.6 mm slices by means of a z-flying focal spot technique [4]. The gantry rotation time

(t_{rot}) is 0.33 s. The 2nd DSCT, the SOMATOM Definition Flash (Siemens Healthineers, Forchheim, Germany) was introduced in 2009. The angular offset of both measurement systems was increased to 95° to provide a larger 33 cm SFOV for the (B)-detector. Both detectors simultaneously acquire 128 overlapping 0.6 mm slices at a gantry rotation time as fast as 0.28 s. The 3rd DSCT, the SOMATOM Force (Siemens Healthineers, Forchheim, Germany) has been on the market since 2014. It provides a further increased 35.5 cm SFOV for the (B)-detector and simultaneous acquisition of 196 overlapping 0.6 mm slices per detector at a minimum gantry rotation time of 0.25 s.

2. Improved temporal resolution and high-pitch scanning

The key benefit of DSCT is its improved temporal resolution for ECG-synchronized cardiac scanning. A DSCT scanner provides temporal resolution of approximately a quarter of the gantry rotation time, independent of the patient's heart rate and without the need for multi-segment reconstruction techniques [1].

Partial scans are used for ECG-synchronized image reconstruction with single source CT, with a scan data segment of 180° plus the detector fan angle (about 50°–60° depending on system geometry). This is the minimum data necessary for image reconstruction of the full SFOV of usually 50 cm in diameter, but not needed for any image pixel within

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Fig. 1. DSCT with two independent measurement systems. a) 1st generation. The system angle between both measurement systems is 90° . b) 2nd generation. To increase the SFOV of detector (B), a larger system angle of 95° was chosen. With the 3rd generation DSCT (c), the SFOV of detector B was further increased to 35.5 cm. Drawings provided by Junia Hagenauer, Siemens Healthineers, Germany.

Fig. 2. In DSCT, the 180° half-scan segment in parallel geometry necessary to reconstruct an image at a pre-defined phase within the patient's cardiac cycle (e. g. at a time T_{del} after the R-peak) is split into two 90° data segments (indicated green and orange) that are acquired by both measurement systems simultaneously at the same anatomical level. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

this SFOV. In a conventional approach the entire partial scan data segment is used for image reconstruction in any point of the SFOV, even if less data were sufficient. Redundant data are weighted using algorithms such as the one described by Parker [5]. The resulting temporal resolution is about two thirds of the gantry rotation time ($2t_{rot}/3$) and constant within the SFOV. To improve temporal resolution, modified reconstruction approaches for partial scan data have been proposed [6,7] that use only the minimum data for the reconstruction of each image point, a half-scan sinogram (180° of scan data) in parallel geometry after re-binning from fan-beam geometry. The temporal resolution then depends on the position of the image pixel within the SFOV. At the iso-center of the scanner where the heart is usually positioned, 180° of the acquired fan-beam data are sufficient for image reconstruction, and the temporal resolution is $t_{rot}/2$.

DSCT systems provide significantly improved temporal resolution for

Fig. 3. Temporal resolution for ECG-gated spiral/helical scanning with MDCT- and DSCT-systems at 0.33 s gantry rotation, both for single-segment (1-seg) reconstruction (green and blue line) and for two-segment (2-seg) reconstruction (black and red line). The DSCT-system provides 83 ms temporal resolution independent of the patient's heart rate by means of single-segment reconstruction. In case of a two-segment reconstruction, depending on the heart rate temporal resolution varies for both systems – MDCT and DSCT. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

cardio-thoracic imaging. Due to the 90° angle between both detectors, the half-scan sinogram can be split up into two 90° data segments which are simultaneously acquired by the two measurement systems in the same relative phase of the patient's cardiac cycle and at the same anatomical level. The two quarter scan segments are appended by means of a smooth transition function to avoid streaking or other artifacts from potential discontinuities at the respective start- and end-projections. With this approach, temporal resolution equivalent to a quarter of the gantry rotation time $t_{rot}/4$ is achieved in a sufficiently centered region of the SFOV (Fig. 2).

For the 1st generation DSCT with $t_{rot} = 0.33$ s the temporal resolution is $t_{rot}/4 = 83$ ms. For the 2nd generation DSCT with $t_{rot} = 0.28$ s it is 75 ms, slightly more than a quarter of the gantry rotation time because of the increased angle between both measurement systems (95°). With the 3rd generation DSCT with $t_{rot} = 0.25$ s a temporal resolution of 66 ms is achieved. With the dual source approach, temporal resolution is independent of the patient's heart rate, because data from only one cardiac cycle are used to reconstruct the image. This is a major difference to single source MDCT-systems, which can provide similar temporal resolution by combining data from several heart cycles into one image in a

Fig. 4. Images of a moving coronary artery phantom. a) DSCT at 70 bpm, b) DSCT at 90 bpm, c) single source CT at 70 bpm, d) single source CT at 90 bpm. Both systems had a gantry rotation time of 0.33 s. Clinical images at similar heart rate from a single source CT e) and a 1st generation DSCT f), both at 0.33 s gantry rotation time. Since data are from different patients and thus a direct comparison is challenging, images indicate the opportunity for improved image quality as a result of the better temporal resolution (courtesy of S. Achenbach, University Erlangen, Germany).

multi-segment reconstruction. With this approach, however, temporal resolution strongly depends on the relation of heart rate and gantry rotation time (see Fig. 3).

Fig. 4 shows axial slices and MPRs of a moving coronary artery phantom at 70 bpm and at 90 bpm both for a 1st generation DSCT system and for a comparable 64-slice single source CT system, both at 0.33 s gantry rotation time. The phantom simulates realistic coronary artery motion. Note the degradation of image quality of the single source CT images at 90 bpm as a consequence of the insufficient temporal resolution of 160 ms at this heart rate, while the DSCT images are nearly free of motion artifacts. At the bottom of Fig. 4 a clinical example is shown, demonstrating the clinical advantage of improved temporal resolution for vessel assessment.

It is interesting to note that multi-segment approaches can also be applied to DSCT systems. In a two-segment reconstruction, the quarter scan segments acquired by each of the two detectors are independently divided into smaller sub-segments acquired in subsequent cardiac cycles of the patient – similar to two segment reconstruction in MDCT. Using a multi-segment approach, temporal resolution again varies as a function of the patient's heart rate, and a mean temporal resolution of about 60 ms can be established at 0.33 s gantry rotation time, see Fig. 3 (minimum

temporal resolution 42 ms). The clinical benefit of a two-segment reconstruction in case of DSCT systems was investigated in detail by Leschka and colleagues in 2008 [9], concluding that despite higher temporal resolution, a gain in overall image quality was not significant. So, while this mode is generally not recommended for coronary angiography examinations, it still may be beneficial for advanced functional evaluations such as the detection of wall motion abnormalities, or the determination of parameters such as peak ejection fraction.

Meanwhile, several clinical studies have demonstrated the potential of DSCT to reliably perform coronary CT angiography in patients with high and irregular heart rates (e. g. [8,10,11]). DSCT is sufficiently accurate to diagnose clinically significant coronary artery disease in some or all difficult-to-image patients [12,13]. In a meta-analysis of 33 studies which compare the diagnostic accuracy of 1st and 2nd generation DSCT coronary angiography for the detection of >50% stenosis with invasive catheter angiography as a reference standard [14], the authors found a pooled sensitivity of 98% and a pooled specificity of 88% on a per-patient basis, with no significant differences in sensitivity or specificity in the subgroup of studies with and without heart rate control. The median radiation dose, however, was smaller in the studies with heart rate control (1.6 mSv) than in the studies without heart rate control (8

Fig. 5. Data sampling scheme along the z-axis (along the body of the patient) for a 64-row DSCT scanner operated at a pitch of 3. Sampling gaps (yellow arrow) of detector A (blue) caused by the high pitch are filled with data acquired by detector B (red). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

mSv).

DSCT systems offer a way to scan the heart within one heartbeat without the need for area detectors covering the entire heart volume. With a single-source CT, the spiral pitch p is limited to $p \leq 1.5$ to ensure gapless volume coverage along the z-axis. If the pitch is increased to $p > 1.5$, sampling gaps occur that hamper the reconstruction of images with well-defined narrow slice sensitivity profiles and without excessive image artifacts. With DSCT systems, however, data acquired with the second tube-detector system – mounted on the gantry with ~ 90 degree offset – a quarter rotation later can be used to fill these gaps, see Fig. 5 [15,16], and the pitch can be increased up to $p = 3.2$.

At maximum pitch, no redundant data are acquired, and a quarter rotation of data per measurement system is used for image reconstruction. Temporal resolution is approximately a quarter of the gantry rotation time. At decreasing pitch, temporal resolution worsens because of the increasing angular data segment that corresponds to an image. At a pitch of 2, for example, temporal resolution is roughly 0.4 times the rotation time, resulting in a temporal resolution of 100 ms with the 3rd generation DSCT [17].

With the high-pitch scan mode, very high scan speed is achieved – up to 450 mm/s with the 2nd generation DSCT (38.4 mm detector coverage, 0.28 s gantry rotation time) and up to 737 mm/s with the 3rd

generation DSCT (57.6 mm detector coverage, 0.25 s gantry rotation time). This enables the examination of larger anatomical ranges in very short scan times, such as chest CTA at high temporal resolution [18], the evaluation of pulmonary embolisms, visualization of most cardiac structures and proximal coronary arteries [19] and fast CTA scans of the aorta at low radiation and contrast dose [20]. High scan speed and correspondingly short scan times are also helpful in examinations of patients with limited ability to co-operate, such as in pediatric radiology [21,22]. Fig. 6 demonstrates the reduction of motion artifacts in a phantom experiment with a moving doll scanned both in standard and high-pitch mode.

The high-pitch scan mode can be combined with ECG-triggering – the patient's ECG triggers both table motion and data acquisition. The patient table is positioned, and table acceleration is started so that the table arrives at the prescribed start z-position (such as the base or the apex of the heart) at the prescribed cardiac phase after full table speed has been reached, see Fig. 7. Then data acquisition begins. The temporal resolution per image is approximately $t_{rot}/4$. The scan data for images at adjacent z-positions, however, are acquired at slightly different phases of the cardiac cycle. Meanwhile, several clinical studies have demonstrated the successful use of the high-pitch scanning technique for coronary CT angiography in patients with sufficiently low and stable heart rates (<65 bpm with the 2nd generation DSCT, <73 –75 bpm with the 3rd generation DSCT), with the potential to scan the entire heart in one beat at a very low radiation dose [23–27].

ECG-triggered high-pitch scans have been used for comprehensive thorax examinations in the emergency room and in the planning and/or checking of TAVR procedures, because they provide adequate visualization of the coronary arteries, the aorta and the iliac arteries in one scan at low radiation dose, see Fig. 8. The very short total scan time may allow for a reduction of the amount of contrast agent administered, see [28,29].

3. Dual energy imaging

With a DSCT system, dual energy data are acquired by simultaneously operating both X-ray tubes at different kV-settings, e.g., 80 kV and 140 kV [3,30,31]. The scan parameters can be individually adjusted for both measurement systems, resulting in a flexible choice of scan protocols with no restrictions in spiral pitch or available tube current (mA) per x-ray tube. In combination with on-line anatomical dose modulation the radiation dose to the patient can be adjusted to the patient anatomy and the planned examination.

The quality of dual energy (DE) CT examinations depends on the separation of both energy spectra. Significant spectral overlap and bad energy separation result in inefficient and unprecise tissue

Fig. 6. CT scans of a moving doll phantom (left) simulating the motion of a child without sedation. a) and b) MPR and VRT of the phantom scanned with a standard spiral (pitch 1, 0.33 s rotation time) show significant motion artifacts. c) and d) Using the high-pitch spiral (pitch 3.2, 0.33 s rotation time) motion artifacts are significantly reduced because of the very short scan time and the optimized temporal resolution per image. Courtesy of C. McCollough, Clinical Innovation Center, Mayo Clinic Rochester, MN, USA.

Fig. 7. ECG-triggered start of table movement and data acquisition for the high-pitch DSCT spiral.

differentiation, which in turn has to be compensated for by increased radiation dose. DSCT systems allow for improved spectral separation by additional pre-filtration of the high-kV beam, e. g. by means of a filter that can be moved into the beam when needed and moved out for non-DE applications. The 2nd generation DSCT uses a tin filter (Sn) with a thickness of 0.4 mm to shift the mean energy (after 20 cm water) of the 140 kV spectrum from 86 keV to 97 keV, see Fig. 9. The mean energy of the 80 kV spectrum is 60 keV. The tin filter has several benefits. It increases the spectral separation between the low- and the high-energy spectrum, it narrows the 140 kV spectrum (which results in better dose efficiency and less beam hardening artifacts), and it reduces cross-scattering. Primak et al. [32] found that adding tin filtration to the high-kV tube improved the dual-energy contrast between iodine and calcium by as much as 290%.

The 3rd generation DSCT further improves spectral separation by providing 150 kV x-ray tube voltage with more aggressive tin pre-filtration (0.6 mm) to acquire the high-energy CT data (see Fig. 9). Fig. 10 shows clinical examples of DE scanning with DSCT.

One method to objectively quantify the performance of a DE CT acquisition technique with regard to energy separation and material differentiation capability is the use of DE ratios [33]. The DE ratio of a material is defined as its CT-number (in HU) at low kV divided by its CT-number (in HU) at high kV. Soft tissue, for example, has a DE ratio close to 1, fat has a DE ratio < 1, bone and iodine have DE ratios > 1.

As a representative clinical example we focus on the DE ratio of iodine. Performing a two-material decomposition into a soft-tissue image and an iodine image in a contrast-enhanced scan is a frequently used clinical application of DE CT. The iodine image visualizes the iodine uptake in different tissues as a surrogate parameter for local perfusion, while the soft-tissue image serves as a “virtual non-enhanced image”, corresponding to a true non-enhanced CT image without administration of contrast agent (see Fig. 12d). Virtual non-enhanced images and iodine maps have been used to characterize renal masses as benign or malignant [34], or to visualize perfusion defects in the lung parenchyma [35] or in the myocardium [36]. The DE iodine ratio is independent of mAs and other reconstruction parameters; it depends only marginally on the iodine concentration for reasonably low amounts

of iodine. The better the spectral separation of the high-energy and the low-energy spectrum, the higher is the DE iodine ratio. The higher the DE iodine ratio, the lower is the image noise of the virtual non-enhanced images, and the more precise is the quantification of iodine in the corresponding iodine maps [31].

The DE iodine ratios for different DE CT acquisition techniques and different CT scanner generations are shown in Fig. 11. They were measured using water-equivalent circular phantoms of different diameters (10/20/30/40 cm), representing different patient sizes. The phantoms had a small tube (diameter 2.0 cm) inserted at the center which was filled with diluted iodinated contrast agent (15 mg iodine/ml - Ultravist 300 diluted by saline, Bayer Healthcare, Germany) representing the typical attenuation of a contrast-enhanced aorta.

The DE iodine ratio for the 2nd generation DSCT at the vendor-recommended x-ray tube voltage combination 100 kV/140 Sn kV is 2.25; the DE iodine ratio for the 3rd generation DSCT at the vendor-recommended x-ray tube voltage combination 90 kV/150 Sn kV is 3.0. This is an improvement by 58% compared to the use of 80 kV/140 kV without spectral shaping.

The DE iodine ratios at other phantom diameters are shown in Fig. 12, together with the measured image noise in virtual non-enhanced CT images at equal radiation doses. The CTDI_{vol} values for the 10, 20, 30 and 40 cm phantoms (1.2, 2.5, 7.2 and 21.2 mGy) were applied to all voltage combinations used to scan the respective phantoms.

Image noise in virtual non-enhanced images is a good indicator for the radiation dose efficiency of a DE technique: the lower the image noise in the virtual non-enhanced images after material decomposition, the lower is the radiation dose needed to obtain virtual non-enhanced images of sufficient quality to replace true non-contrast images. At equal radiation dose and equal phantom size, image noise is highest at 80/140 kV. For the 40 cm phantom, image noise at 80 kV/140 kV without spectral shaping is twice as high as at 90 kV/150 Sn kV.

Not only iodine and virtual non-contrast images, but also so called virtual monoenergetic images (VMI) benefit from improved spectral separation. VMI images are based on a two-material decomposition into iodine and soft tissue. By applying appropriate scaling factors, the CT numbers of iodine and soft tissue in the VMI images correspond to a

Fig. 8. ECG-triggered high-pitch spiral scan of the aorta of a 80-year-old male patient (92 kg) early after TAVR, oblique sagittal thick MIP (30 mm). Acquisition performed with 120 kVp, CTDIvol = 4.98 mGy and DLP = 355 mGyxc. Courtesy of Universitätsspital Zürich, Switzerland.

fictitious measurement with a monoenergetic x-ray beam at a specific energy (in keV). The CT numbers of other materials may not reflect their actual enhancement at the desired energy. VMI images based on textbook two-material decomposition suffer from increasing image noise at low and high keVs. However, refined processing techniques have been introduced that mitigate this noise increase, such as the Mono+ algorithm [37]. Using Mono+, improved iodine contrast to noise ratios (CNR) were demonstrated in phantom scans, comparable to or even surpassing low kV scanning [38]. Best results were obtained for high DE ratios, for example 80/140 Sn kV. Husarik et al. [38] confirmed these findings in a phantom and patient study. They found added value for imaging of liver lesions with Mono+ by decreased noise, increased CNR, and higher lesion conspicuity, although with limitations in very large body sizes. Other studies reported the ability to reduce metal artifacts by

Fig. 9. Typical 80 kV and 140 kV spectra (after 20 cm water), normalized to equal areas under the curves (top). 80 kV spectrum and 140 kV spectrum with additional 0.4 mm tin pre-filtration (center), and 80 kV spectrum and 150 kV spectrum with additional 0.6 mm tin pre-filtration (bottom). Note the shift of the mean energy of the high energy spectrum to higher values (arrow) and the reduced spectral overlap.

the use of Mono+ at high keV levels [39,40].

Spectral pre-filtration of the high-kV beam is a key technique to enable DE CT at low radiation dose. Meanwhile, several clinical studies have demonstrated the feasibility of dual source DECT at similar or reduced radiation dose compared with single energy CT (SECT). Schenzle et al. [41] report the feasibility of dual source DE CT without increasing radiation dose in chest CT. The authors claim that contrast-to-noise ratios (CNR) can be doubled with optimized DE CT reconstructions. They conclude that CT can be performed routinely in DE mode without an increase in radiation dose or compromises in image quality. Bauer et al. [42] compare radiation dose and image quality of 64-slice CT and dual source DE CT for CT pulmonary angiography (CTPA). They observe significant dose reduction with 2nd generation DECT operated at 80/ 140 Sn kV with image quality similar to single source 120 kV CTPA. In a direct comparison of radiation dose and image quality of contrast-enhanced single- and dual-energy CT examinations of the chest in matched cohorts, Lengua et al. [53] observed that 2nd and 3rd generation dual source DECT examinations can be performed without increased radiation exposure or decreased image quality compared with single-energy CT (SECT) acquisitions. In a retrospective analysis of 200 abdominal CT scans of patients matched by gender and body mass index, Wichmann et al. [51] found similar effective radiation dose at similar subjective image quality and reader confidence for single-energy and dual-energy scans both for 2nd generation and for 3rd generation DSCT, at a systematically lower radiation dose for 3rd generation DSCT. In an emergency department setting, dual source DE CT of

Fig. 10. Left. pre-operative CTA of the chest for a 12-year old boy to assess the pulmonary vascular anatomy (a) with a superimposed iodine map of the lung parenchyma to assess lung perfusion (b); acquired with 80/140Sn kV on a 2nd generation DSCT (courtesy of M. Siegel, Washington University School of Medicine, St. Louis, USA). Right. 57-year old patient (BMI 36) with renal cell carcinoma. Virtual mono-energetic image at 70 keV (c) and virtual non-contrast image (d); acquired with 100/150Sn kV on a 3rd generation DSCT (courtesy of D. Marin, Duke University, Durham, USA).

Fig. 11. DE iodine ratios for different DE acquisition techniques at 20 cm phantom diameter. kV switching 80 kV/140 kV, 1st generation DSCT at 80 kV/140 kV, 2nd generation DSCT at 100 kV/140 Sn kV and at 80 kV/140 Sn kV (0.4 mm tin pre-filtration), 3rd generation DSCT at 100 kV/150Sn kV, at 90 kV/150 Sn kV, at 80 kV/150 Sn kV and at 70 kV/150 Sn kV (0.6 mm tin pre-filtration).

the abdomen and pelvis was performed with decreased radiation dose when compared with SECT, but demonstrated improved objective measurements of image quality, and equivalent subjective image quality [52]. In a retrospective study in a pediatric population (79 children who underwent thoracic or abdominal-pelvic CT or CT angiography), the use of dual source DECT resulted in radiation exposures comparable to or less than those of SECT while maintaining contrast and CNR [54]. A

comprehensive overview on radiation dose in DE CT can be found in [43].

4. Challenges for dual source CT systems

Despite their clinical benefits, DSCT systems have to address a number of challenges. One major challenge for image reconstruction is

Fig. 12. DE iodine ratios (left) and image noise in the virtual non-enhanced CT images (right) for different DE acquisition techniques at different phantom diameters. For each phantom diameter the radiation dose was kept constant for the different techniques. The lower the image noise in a virtual non-enhanced CT image, the better the radiation dose efficiency of the respective DE technique [32].

Fig. 13. Locations of interaction points for cross-scattered X-ray photons in a DSCT scanner. (a) X-ray photons coming from tube (A) at the bottom are scattered into detector (B). (b) X-ray photons coming from tube (B) at the top are scattered into detector (A). Water images without cross-scatter correction (c) and with cross-scatter correction (d).

data truncation: for a compact gantry design, one detector (A) covers the entire SFOV ($\varnothing$ 50 cm), while the other detector (B) is restricted to a smaller, central field of view ($\varnothing$ 26 cm resp. 33 cm), see Fig. 1. Consequently, the projection data of detector (B) are truncated if the scanned object extends beyond the central field of view, and the data will have to be extrapolated before reconstruction to avoid truncation artefacts in

the images. Data acquired with detector (A) are used to extrapolate the truncated projections of detector (B). The extrapolation is done in parallel geometry. Due to the mechanical assembly of the DSCT system, the corresponding (A)-data used to extrapolate (B)-data at a certain projection angle θ are acquired either a quarter rotation earlier (same half turn of the spiral), or a quarter rotation later (next half turn of the

Fig. 14. Images of an anthropomorphic thorax phantom with heart insert, scanned on a DSCT system. The X-ray beam width in the z direction was 38.4 mm at iso-center. FoV 420 mm, window width 300 HU, window center 40 HU. a) No scatter correction. The arrows indicate scatter artifacts due to direct scatter and cross-scatter. b) Measurement-based scatter correction. c) Model-based scatter correction.

spiral).

Another challenge is cross-scattered radiation, e.g. scattered radiation from X-ray tube (B) detected by detector (A) and vice versa, see Fig. 13. Cross-scattered radiation can produce artefacts and degrade the contrast-to-noise ratio of the images. It can result in incorrect material decomposition and material classification in DE scans.

Cross-scatter requires adequate correction [30]. The most straightforward correction approach is to directly measure the cross-scattered radiation in detectors (A) and (B) and to subtract it from the measured signal. This technique is implemented in the 2nd generation DSCT. It requires additional detector elements on each detector outside the direct beam. An alternative to direct measurement is a model-based cross-scatter correction. The primary source of cross-scattered radiation is Compton scatter at the object surface, knowledge of the surface is therefore sufficient to predict cross-scatter. The object surface, however, can be readily determined by analysing the outline of the raw data sinogram. This technique is realized in the 1st generation DSCT. Pre-stored cross-scatter tables for objects with similar surface shapes are used for an on-line correction of the cross-scattered radiation. The results of both measurement-based and model-based cross-scatter correction are shown in Fig. 14.

In the 3rd generation DSCT, a correction based on a simplified Monte-Carlo simulation of cross-scatter is implemented.

A 3rd challenge for dual source DE CT is the 90° offset of the projections acquired by both x-ray tubes at the same z-position. Even though high-energy and low-energy DE data are simultaneously acquired with regard to their z-position, their projection angles are different. Raw data-based DE algorithms are therefore difficult to

realize. Consequently, DE algorithms are image-based (see also [44], chapter 39). Beam hardening is often claimed to be a limiting factor for image-based DE methods. However, under certain pre-conditions typically fulfilled in modern CT-scanners, image-based methods are practically equivalent to raw data-based methods. One pre-requisite is the validity of the thin absorber model. If we use water and iodine as the base materials for image based dual energy decomposition, the maximum x-ray attenuation of the iodine along any measured ray path is expected to be so small that it is valid to assume a linear contribution to the total attenuation. The thin absorber model holds for iodine samples with up to 5000 HU·cm in water, corresponding to the clinical situation of an object with 200 HU iodine enhancement and 25 cm thickness. In clinical practice, this pre-requisite is only violated in extreme situations when very high concentrations of iodine are present, such as in CT nephrographic studies. As a second pre-requisite, both the CT-value of water and the CT-values of small iodine samples are expected to be independent from their position within the scanned object. DSCT scanners are therefore equipped with an optimized bowtie filter providing sufficient beam hardening, and the approximately cylindrical patient cross-section has to be centered within the SFOV. In practice, electronics noise, scanner calibration, stability of emitted spectra, cone beam effects, and scattered radiation can have a larger impact on the obtained results than the raw data or image-based analysis method. More specifically, DSCT scanners are equipped with an iterative algorithm to correct for iodine-related beam hardening, which is separately applied to low- and high-kV images prior to material decomposition. The algorithm has been evaluated and found to significantly reduce beam hardening artifacts and improve DE results [45], such as virtual non-

Fig. 15. Dual source DE iodine image of the myocardium, acquired on a 3rd generation DSCT (temporal resolution 125 ms). Left. without beam hardening correction. Beam hardening artefacts along a line between aorta and left ventricle (arrows) can mimic perfusion defects (darker zones with less iodine enhancement). Right. with iterative beam hardening correction. The beam hardening artefacts are significantly reduced.

contrast, iodine and/or monoenergetic images, see Fig. 15

Object motion is yet another challenge for dual source DE CT because of the 90° offset of the projections acquired by both measurement systems. Motion artefacts in the corresponding (A)- and (B)-images can be slightly different, which may affect material decomposition. In practice, however, this problem is not relevant thanks to the good temporal resolution of DSCT, and it can be further mitigated by non-rigid registration of A and B images.

5. Conclusion

Dual source CT, first introduced in 2006, has been refined in three scanner generations. Today, DSCT is widely used in academic and non-academic imaging centers. DSCT was primarily designed to achieve improved temporal resolution for cardiac CT, and clinically robust coronary CT angiography even at high and irregular heart rates could soon be demonstrated (see e. g. [14]). Other potential opportunities of the DSCT system architecture in addition to its cardiac capabilities were quickly discovered. The ability to acquire dual energy data by operating both X-ray tubes at different tube potential lead to a renaissance of dual energy CT, which had already been evaluated in the 1980's ([46,47]) but was abandoned because of technical limitations of the CT scanners at those times. Today, all major CT vendors offer DE CT scanners relying on different technical solutions [48]. Further approaches for dual energy scanning such as CT systems equipped with photon counting detectors allowing intrinsically for the combination of high resolution and spectral information are evaluated on first prototype systems [49]. The DSCT-inherent high-pitch scanning paved the way for fast CT angiographic studies and facilitated a broader acceptance of CT for paediatric examinations because of its potential to avoid sedation [50].

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