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Dual-Layer Computed Tomography in Cardiovascular Imaging

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Despite the established theoretical potential of dual-energy CT (DECT), the growing prevalence of dual-source CT, and rapid peak kilovoltage switching CT, it is not frequently used in routine clinical cardiovascular imaging, especially outside academic institutions. The recently introduced dual-layer CT offers several advantages over DECT analysis for routine clinical use. This review aims to illustrate the basics of DECT, including dual-layer CT, and its applications for cardiovascular imaging.

Key words Multi-slice computed tomography · Dual energy · Cardiovascular.

INTRODUCTION

Since the introduction of dual-energy CT (DECT) in 1976 [1,2], several studies have reported its advantages in various examinations, including the quantification of calcium [3], iron [4], and fat [5]. However, the dual-energy system remained limited to research applications because of its prolonged scan time. Approximately three decades later, Siemens introduced the dual-source DECT scanner (DSCT, SOMATOM Definition; Siemens, Erlangen, Germany) in 2005 [6], and GE launched the rapid peak kilovoltage (kVp) switching CT (KVSCT, Discovery CT750 HD; GE Healthcare, Waukesha, WI, USA) scanner in 2010 [7], thereby drastically reducing the misregistration attributed to the motion in DECT and facilitating the procurement of dual-energy data in any examination during a single breath hold. However, to date, DECT has not been extensively used in clinical practice, especially cardiovascular imaging. In 2015, Philips introduced the dual-layer DECT scanner (DLCT, IQon Spectral CT; Philips Healthcare, Best, Netherlands), which is a new DECT system in 2018. This review aims to summarize the principle and clinical efficacy of DLCT compared to other DECT systems in cardiovascular imaging.

BASIC PRINCIPLES OF DUAL-ENERGY COMPUTED TOMOGRAPHY

DECT is defined as the use of attenuation measurements attained with various energy spectra, along with the use of known variations in attenuation between two spectra to distinguish and quantify material composition based on energy-dependent attenuation profiles of specific materials. Typically, DECT acquires two image datasets from the same anatomic location with different X-ray energies [1,2]. In the early days of CT, successive single-slice attainments with different kVp were performed as a dual-energy technique. In contrast, the DSCT system is equipped with two data measurement systems, each comprised of one X-ray tube and one corresponding detector array. In DECT mode, two X-ray tubes are set at different energies (such as 80 kVp and 140 kVp), and two different energy image sets are obtained simultaneously. In addition, the KVSCT system can switch the X-ray energy between the low- and high-energy settings within microseconds, which facilitates nearly simultaneous temporal and spatial registration of the two energy datasets. Fig. 1 summarizes the mechanism of three DECT systems.

Despite variable acquisition mechanisms with CT scanners, the fundamental principle of DECT remains nearly the same. Two critical physical processes accountable for attenuation of X-rays passing through a material (at X-ray energies of clinical CT) are the photoelectric effect and Compton scattering. Although Compton scattering is relatively similar across the majority of materials, the photoelectric effect is proportional to the cube of

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Fig. 1. Schema of three different dual-energy CT systems. (A) Dual-source CT. (B) Rapid peak kilovoltage (kVp) switching CT. (C) Dual-layer CT. CT: computed tomography.

Fig. 2. The graph of linear attenuation coefficients for iodine ($Z=53$), calcium ($Z=20$), and nitrogen ($Z=7$). The plotted linear attenuation coefficients were obtained using the energy-dependent mass attenuation coefficients from the National Institute of Standards. Linear attenuation coefficients of high atomic number objects increased at low keV images.

Fig. 3. Schema of the two material decomposition methods. The algorithm is based on the difference in densities of two materials in HU. Higher-atomic number objects exhibit higher attenuation changes. HU: Hounsfield units, CT: computed tomography.

the atomic number and inversely proportional to the incident photon energy (Z^3/E^3). Thus, different materials possess different energy-dependent attenuation profiles (Fig. 2). Theoretically, two different X-ray energy images enable material characterization. Apparently, small differences are suggestive of low atomic number objects, and substantial differences are suggestive of high atomic number objects; this difference constitutes the foundation of all DECT analyses.

The primary applications of DECT analyses involve material decomposition and quantification. Of note, the definition of the optimal cut-off line in the X–Y plane of two different energy images facilitates the performance of two-material decomposition (Fig. 3). In addition, such separation can offer paired material datasets (e.g., water/iodine, uric acid/calcium, etc.), enabling material quantification. For instance, the iodine-density map can be obtained by iodine images with suppressed water information when the paired datasets of water/iodine are

generated (Fig. 4). Conversely, water images provide the density map of objects with suppressed iodine information; this method is called virtual non-contrast imaging.

Another broadly used application of DECT is the virtual-monochromatic energy image (VMI), which synchronizes monoenergetic images based on material decomposition by DECT in the projection domain. In addition, VMI finds utility in routine diagnosis methods analogous to conventional polychromatic images attained at a single X-ray tube potential; this method offers several advantages, including a beam-hardening correction. Reportedly, the beam-hardening effect is attributed to the preferential absorption of low-energy photons of a polychromatic X-ray beam when it passes through an object [8]. Consequently, the filtered beam contains high-energy photons (thereby a harder beam) and results in attenuation error between various angles, causing artifacts on the reconstructed images, including streaks, dark bands, and cupping artifacts. Theo-

Fig. 4. An example of material decomposition. The conventional CT images (A) can be separated into two materials as follows: (B) iodine concentration map and (C) water image-virtual non-contrast image by DECT. CT: computed tomography, DECT: dual-energy CT.

retically, beam-hardening artifacts are not observed in CT images acquired by a monochromatic X-ray. Previous studies investigating DECT have illustrated that “virtual” monochromatic images are valuable in decreasing beam-hardening artifacts [9]. Moreover, the VMI of low keV has been shown to increase the contrast of iodine parallel to a low kVp scan, suggesting the possibility to compensate for image contrast deficiency [10-12].

As mentioned earlier, although DECT offers several advantages compared to single energy CT (SECT), it has not been extensively used in routine clinical imaging for two primary reasons. First, DECT systems are based on SECT, which cannot perform DECT analyses with image data scanned in SECT mode, necessitating performance of the examination in DECT mode. Second, the image quality of DECT usually is worse than SECT, because the mean beam energies of low- and high-tube voltages are not significantly different, and the evaluated attenuation coefficient values of low- and high-energy images are similar and unstable. The instability of the attenuation coefficients at different energies accounts for severe image noise in VMI, especially at high or low keV settings, because these evaluated errors were amplified with distance from the assessed energies. Instead, these measured errors were averaged and compensated at a moderate keV (Fig. 5). In addition, DECT requires two different images with nearly the same radiation dose compared to SECT and affects the image quality of DECT. A study reported that the image noise of the conventional VMI at approximately 70 keV was marginally lower than that of SECT; however, the image noise was exceptionally high in both low and high keV images [13]. In addition, the CNR of iodine was shown to decrease at low keV because the increment of image noise at low keV was higher than that observed with contrast enhancement of iodine. Thus, DECT has been primarily used for material quantification, and low kVp SECT was used to enhance the image contrast during contrast material dose reduction.

Fig. 5. Details of noise amplification by DECT. If the energy of the VMI is intermediate between the low- and the high-energy images, estimation errors are compensated by averaging. Conversely, in the low or high VMI, estimation errors are amplified. Therefore, the image noise of the VMI was significantly increased at the low-energy setting, which also increased in the high-energy setting. HU: Hounsfield units, CT: computed tomography, DECT: dual-energy CT, VMI: virtual-monochromatic energy image.

ADVANTAGES OF DUAL-LAYER COMPUTED TOMOGRAPHY

In 2015, the DLCT system was introduced for clinical practice to overcome these disadvantages. All DECT systems attained two X-ray energy images by differentiating the X-ray tube voltage. Of note, DLCT utilizes a unique dual-layer scintillator system, allowing the attainment of high- and low-energy data simultaneously in time and space at the detector level. Fig. 6 details the dual-layer scintillator system; the top-layer yttrium-based garnet scintillator is designed to be sensitive to low-energy photons, and the bottom-layer gadolinium oxysulfide scintillator is designed to be sensitive to high-energy photons.

The DLCT system can use these two images for material decomposition, material quantification, and to create a virtual monochromatic image analogous to other DECT systems. This system offers three significant advantages compared to X-ray

tube-based DECT systems. First, the DLCT system provides conventional images for all examinations. The DECT mode of the X-ray tube-based DECT systems uses low- and high-tube voltage to provide two energy image sets, which cannot be produced in conventional tube voltage images. Thus, with DECT mode, these images and previous images with similar image quality cannot be compared. In addition, waiting for the reconstruction and blending of the two energy image sets is very time-consuming. In contrast, the DLCT system uses the conventional tube voltage (120 kVp) for all examinations, with no misregistration between low- and high-energy datasets. Hence, the DLCT system can instantly offer conventional tube voltage images by integrating the low- and high-energy projection data as fast as a single energy scan, which is why conventional energy images are observed immediately, facilitating comparison with previous studies using the same image quality.

Second, the DLCT system offers retrospective DECT analysis for all examinations without any specific settings. In the X-

ray tube-based DECT system, users need to decide whether to use DECT mode or not. This scanning workflow functions accurately only in cases in which it can be ascertained that the examination might require DECT analysis before scanning. However, to anticipate whether SECT is adequate for the image diagnosis before scanning is typically challenging. In the DLCT system, however, the data from the two layers are documented at all times, facilitating dual-energy analysis on every dataset acquired. Of note, DECT analysis need not be performed if SECT is adequate for image diagnosis, and DECT analysis can be performed in all examinations in which SECT is not adequate for the image diagnosis.

Third, currently, the image quality of the DLCT system is better than that of the X-ray tube-based DECT system. Fig. 7 presents the relative image noise of 70 keV images from conventional DECT and DLCT. Apparently, the DECT mode of the DSCT method triggers substantial cross-scatter; as both X-ray tubes function simultaneously, photons from one X-ray tube can be scattered into the un-paired detector. Reportedly, the described effect decreases the signal-to-noise ratio [14]. In addition, the virtual monochromatic images in the projection domain offer an advantage of being capable of correcting beam-hardening artifacts [15]; however, this method requires the angular matching of projections, which is challenging on DSCT [16]. Furthermore, the DECT mode of the KVSCT systems poses other concerns. The ideal system for energy switching must utilize a square wave variation in the tube voltage and tube current to offer distinct high-quality X-ray energy images. However, real KVSCT systems alter tube voltages following a sinusoidal wave, and effective X-ray energies fluctuate imperfectly [17], resulting in substantial spectral overlap relative to the DSCT system. In addition, the present KVSCT systems cannot modulate the tube current among various tube voltage pro-

Fig. 6. An illustration of the dual-layer detection system. The photodiodes are parallel to the X-ray direction, attached to the sides of the two types of scintillator elements.

Fig. 7. A graph of relative image noise within 70 keV images of conventional DECT and DLCT. The image noise of conventional DECT increased significantly in the 40 keV setting. In contrast, the image noise of DLCT was relatively constant at all keV settings. DECT: dual-energy CT, DLCT: dual-layer CT, CT: computed tomography.

jections, leading to noisier low-energy images. The DLCT facilitates perfect angular matching of projections because high- and low-energy detectors are in the same positions, enabling pro-

jection-based beam-hardening correction and offering exact material decomposition in the semi-anthropomorphic abdomen phantom that requires accurate beam-hardening correla-

Fig. 8. A 79-year-old female patient with deep vein thrombosis. It is difficult to detect pulmonary thrombosis at 120 kVp (A and C); however, the 40 keV image (B and D) clearly depicted the thrombosis (arrow). The iodine concentration map and effective Z map clearly depicted the peripheral hypo-perfusion area (C). (E) Effective Z map, (F) iodine concentration map.

tion [18]. Moreover, X-ray photons are distributed into an optimal rate for the bottom and top layers in the DLCT system, allowing the usage of more X-ray photons than in KVSCT. Previously, some studies have reported that the image noise values of two material images created by DECT, especially pixels, are highly correlated, and that filtering and iterative reconstruction can decrease the correlated noise [19-21]. Further, iterative reconstruction with material decomposition, including an anti-correlated noise filter, is another advantage of DLCT. A recent phantom study comparing the VMI of DSCT, KVSCT, and DLCT reported that the increased proportion of image noise from 140 keV to 40 keV was minimal with DLCT (14%) compared to DSCT (265%) and KVSCT (363%) [18]. Consequently,

the low-energy VMI of DLCT enhances the CNR of iodine similar to the low kVp CT and can be used to enhance image contrast for contrast material dose reduction.

The disadvantages of DLCT are the narrow detector width (40 mm) and small detector rows (64) compared with other CT scanners. Future improvements are needed in detector width, especially for electrocardiogram-gated CT.

VARIOUS APPLICATIONS OF DUAL-LAYER COMPUTED TOMOGRAPHY IN CARDIOVASCULAR IMAGING

This review demonstrates the efficacy of DLCT for various

Fig. 9. A 79-year-old female patient with deep vein thrombosis. The image contrast was increased at the low-energy VMI compared to the high-energy VMI: (A) 40 keV, (B) 50 keV, (C) 60 keV, (D) 70 keV, (E) 80 keV, and (F) 120 keV. However, the image noise was nearly constant at all energies. VMI: virtual-monochromatic energy image.

cardiovascular imaging applications. In routine clinical examination, iodine contrast increment by the low-energy VMI is widely used. Fig. 8 demonstrates a representative case of contrast dose reduction in the examination of a patient with pulmonary embolism with severe renal dysfunction, in whom we could not sufficiently insert a needle for rapid infusion of contrast media. Thus, low-energy images might be desirable to increase iodine contrast; however, conventional DECT suffers from severe image noise [13]. In contrast, the low-energy VMI of DLCT can fulfill both high-contrast and low-image noise requirements. Fig. 9 demonstrates another representative case of the efficacy of low-energy VMI for deep vein thrombosis. Image noise of low-energy VMI is significantly lower than that of conventional DECT systems, and it is useful to increase the image contrast of the contrast media.

The precise quantifications of CT attenuation and iodine dose

are advantages of DECT and DLCT. Fig. 10 demonstrates the effectiveness of VMI to suppress the beam-hardening artifact. Of note, the case presented in Fig. 10 has no coronary stenosis or perfusion defect. Although there must be no perfusion defect in the myocardium, a low-density area was observed in the posterior wall. The VMI and iodine-density map can efficiently correct beam-hardening artifacts from the chest wall because beam-hardening mainly arises from the polychromatic character of X-rays used in clinical CT scanners. Apparently, DECT is highly useful for late iodine enhancement (LIE) imaging of the myocardium, which could be an alternative to late gadolinium enhancement (LGE) imaging in cardiac MRI because the pharmacokinetics of iodinated contrast materials are parallel to those of gadolinium-containing contrast materials [22]. Fig. 11 illustrates the representative case of LIE imaging with DLCT. Some studies on LIE using SECT have reported that LIE has a weak

Fig. 10. A 50-year-old female with chest pain. The beam-hardening artifact (arrow) in 120 kVp CT (A) is suppressed in the 70 keV (B) image by dual-layer CT. CT: computed tomography.

Fig. 11. A 63-year-old female with hypertrophic cardiomyopathy. It is difficult to detect the enhanced lesion in the LIE image at 120 kVp (A); however, a 40 keV image (B) clearly depicts the LIE (arrows) in the septal and anterior wall, similar to late gadolinium enhancement (arrowheads) with MRI. (C) Late Gd Enhancement imaging in MRI. LIE: late iodine enhancement.

Fig. 12. A 65-year-old male with known left anterior descending stenosis. The computed tomography values of two plaques were identical (70.2 HU) in 120 kVp images (A and B). However, on the spectral curve (C), attenuation curves of these two plaques were not parallel. These findings suggest that the components of these plaques might be different. HU: Hounsfield units, ROI: regions of interest of coronary plaques.

point of relatively poor contrast resolution compared to LGE imaging; however, DLCT could potentially overcome this disadvantage [23]. Fig. 12 illustrates the representative case of DECT analysis of coronary plaque with DLCT. Theoretically, DECT has the potential to evaluate the effective atomic number of coronary plaque components and to increase the diagnostic performance of lipid-rich plaques. However, conventional DECT systems have a limitation in temporal and spatial resolution for the electrocardiogram-gated CT, which limits DECT application for coronary plaque component analysis. On the other hand, since DLCT does not compromise the temporal and spatial resolution of cardiac CT, it might be well-suited for coronary plaque analysis. DLCT could potentially improve the future diagnostic performance of coronary plaque components.

Some studies on LIE using SECT have reported that LIE has a weak point of relatively poor contrast resolution compared to LGE imaging; however, DLCT could potentially overcome this

disadvantage.

CONCLUSION

In all examinations, DLCT always operates in SECT and DECT modes, and the quality of the DECT analysis is higher than other DECT systems, primarily in cardiovascular imaging. This review infers that DLCT has revolutionized DECT analysis and rendered it independent. Nevertheless, further studies are warranted to elucidate the advantages of DLCT for cardiovascular imaging.

Conflicts of Interest

Shinichi Tokuyasu is an employee of Philips Medical, Japan.

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