1 Introduction and Technical Principles

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Technological advances continuously increase the clinical applications and diagnostic value of CT.

CT was born in the late 1960s, and spiral CT was introduced just in the early 1990s; it marked one of the important steps in the evolution of CT imaging techniques. This technology allowed to acquire volume data without the risk of miss- or double-registration, enabling the reconstruction of images at any position along the patient's length axis as well as reconstructions of overlapping images to improve the longitudinal resolution. It also reduced scan times significantly.

The introduction of multi-detector row computed tomography in 1998 (MDCT) allowed other improvements. Larger anatomical volumes could be acquired with a single acquisition. The first generation of MDCT systems offered simultaneous acquisition of 4 slices, which provided considerably improved scan speed and longitudinal resolution as well as better utilization of the available X-ray power. MDCT also made longer scan ranges with substantially reduced slice width feasible, which is essential, for example, in CT angiography. The introduction of 16-slice CT scanner enabled isotropic sub-millimeter spatial resolution.

The generation of 64-slice CT systems was introduced in 2004. Two different technologies were launched in the market.

Scanners introduced by GE, Philips, and Toshiba increased the volume coverage speed by using 64 detector rows instead of 16, with a body coverage along the Z-axis of about 4 cm. Scanner introduced by Siemens used 32 physical detector rows (body coverage about 2 cm) in combination with "double Z sampling," enabled by periodically moving the focal point in the *Z* direction, in order to simultaneously acquire 64 overlapping slices. The objective of this system was to increase the longitudinal resolution and reduce spiral artifacts independent of pitch. Acquisition is faster with 64 detector rows but *Z*-axis resolution is more accurate with double *Z* sampling (0.38-mm instead of 0.6 mm).

A recent development in CT has been the introduction of dual-energy technology (2005). Dual-energy CT implies simultaneously acquiring data sets at two different photon spectra [1].

Dual-energy CT is limited to two energies. Multi-energy imaging is a challenge today and has the potential to greatly expand the clinical application of tissue differentiation.

In this chapter, technical aspects of dual- and multi-energy CT will be analyzed.

1.1 Dual-Energy CT

The first experiments with dual-energy CT date back to the late 1970s. However, the poor spatial resolution of early computed tomography and the long scan durations at that time prevented the application of the technique. Another major problem was that tube technology did not provide sufficient tube currents at low tube voltages to achieve a sufficient output of quanta relative to the higher voltage tube. Dual-energy

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technology aims to better distinguish different materials in body tissues, thus improving diagnostic power of CT.

Recently, three dual-energy CT systems have been developed and are commercially available: Dual Source, Rapid kV switching, Sequential Dual Energy [2]. Regardless the kind of CT system, material differentiation is based on the same physical X-rays interactions with materials.

The differentiation of material in computed tomography is based on their X-ray attenuation as quantified in Hounsfield Units and displayed in shades of gray at different window levels in normal CT scans. Attenuation is caused by absorption and scattering of radiation by the material under investigation. The two main mechanisms responsible for these effects in the photon energy range used in CT are the Compton scatter and the photoelectric effect.

Photoelectric effect is produced when a low-energy photon collides with an electron; the energy is completely absorbed by the electron, which moves from its position; an ionized atom is obtained and the photon disappears.

Compton effect is produced when a high-energy photon collides with an electron. In this case, the electron absorbs a part of the energy and moves inducing an ionized atom; a new photon, with lower energy and a different direction, is emitted. If the scattered photon still has enough energy left, the process may be repeated (Fig. 1.1).

Another important physical phenomenon should be taken into account while thinking of X-rays interactions: k edge.

K edge describes a sudden increase in the attenuation coefficient of photons occurring at a photon energy just above the binding energy of the K shell electron of the atoms interacting with the photons. The sudden increase in attenuation is due to photoelectric absorption of the photons. For this interaction to occur, the photons must have more energy than the binding energy of the K-shell electrons. A photon having an energy just above the binding energy of the electron is therefore more likely to be absorbed than a photon having an energy just below this binding energy.

The two X-ray contrast media iodine and barium have ideal K-shell binding energies for absorption of X-rays, 33.2 and 37.4 keV, respectively, which is close to the mean energy of most diagnostic X-ray beams.

X-ray absorption depends on the inner electron shells: dual-energy CT is sensitive to atomic number and density, but it is not sensitive to chemical binding.

Considering that any substance has a different photoelectric effect, Compton effect and k-edge at different energy levels, it is possible to understand how dual-energy CT works.

In particular at 80 kV, iodine has its maximum absorption; applying different X-ray spectra and analyzing the differences in attenuation, as we can see in dual-energy acquisition, iodine can be easily differentiated from other materials that do not show its behavior at 80 kV. Iodine is the unique material that doubles its HU values from 140 to 80 kV. At 80 kV, some other materials have higher CT values such as bone, metal, but do not double; at 140 kV, some materials have higher CT values, such as fat, plastic, uric acid; moreover, some materials have the same CT values at 80 and 140 kV, such as water, soft tissues, blood. Since these material behavior is known, tissues can be differentiated [3].

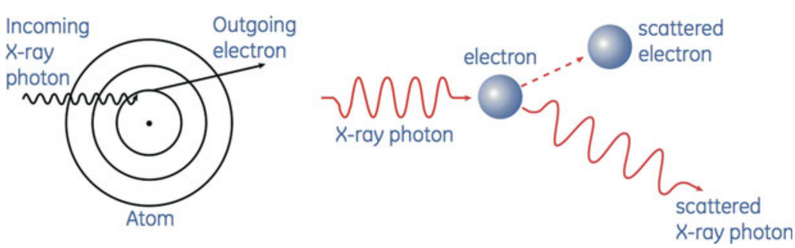

Fig. 1.1 Photoelectric effect (*left*) and compton effect (*right*) are the physical basis of X-ray attenuation (Courtesy of GE Healthcare, Waukesha, U.S.A.)

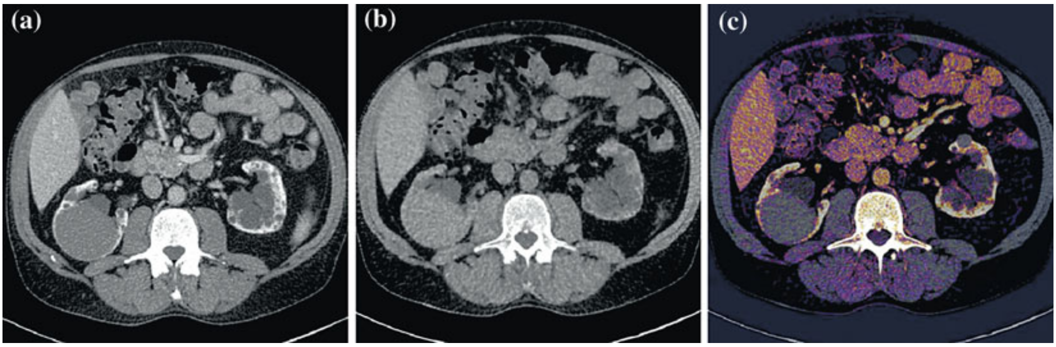

Fig. 1.2 An hyperdense mass can be seen in the diagnostic image (**a**), in the Virtual Non-Contrast (VNC) image (**b**), obtained with subtraction of iodine data, the mass results already hyperdense without contrast; moreover, in iodine map image (**c**) the mass has not iodine enhancement. Therefore, in a single dual-energy acquisition, the diagnosis can be obtained (hyperdense cyst. Courtesy of Siemens AG, Muenchen, Germany)

Considering iodine properties at 80 kV, in dual-energy CT a virtual-unenhanced image can be generated, and this data may be used for baseline density measurements, thereby making true unenhanced imaging unnecessary and saving radiation.

Using dual-energy post-processing software, the contrast agent can be digitally subtracted from the image. This can be done because the dual-energy index of iodine is significantly different from the dual-energy index of soft tissue and fat. The dual-energy data can also be used to generate a color-coded image that shows the distribution of iodine within the volume of tissue examined by CT. This color-coded display is very sensitive to subtle enhancement.

In abdominal imaging, for example, this can be useful in cases of incidentally detected renal lesions with high attenuation on unenhanced CT, the main differential diagnostic considerations being hyperdense cysts and renal masses (Fig. 1.2).

Immediately after reconstruction, the images can be semi-automatically post-processed, offering the radiologist a fast diagnosis.

In clinical application, dual energy helps to visualize and segmentate vessels, contrast media, tendons, ligaments, kidney stones, gout deposits, perfused blood volume, hard plaque, iodine content in small vessels. Any of these structure often has its dedicated software application for analysis. Moreover, post-processing applications can give additional information about body tissues composition.

Groups of two or three materials are usually analyzed; for example, iodine, soft tissue, and water, or water, calcium, and uric acid (kidney stones application); for the visualization of calcifications in the setting of a contrast-enhanced examination, the algorithm will analyze iodine, calcium, and soft-tissue densities [4].

1.2 Dual-Source CT

In dual-source CT system, released by Siemens in 2005, two X-ray tubes with two corresponding detectors are mounted onto a rotating gantry with a 90 angular offset (Fig. 1.3).

For each tube, a 64-slice design (Siemens Somatom Definition) is present: the “A” detector, which is equal in size to a standard detector (50 cm), and the “B” detector, which also has a 64-slice design, but with a reduced field of view of 27 cm. This detector array provides high spatial resolution of isotropic 0.38 mm edge-length voxels and allows a rapid acquisition of a *Z*-axis volume.

In 2008, the second-generation DSCT “Siemens Somatom Definition Flash” was introduced. It features even faster gantry rotation (0.28 s), twice the number of detector slices and a larger field of view (332 mm) [5].

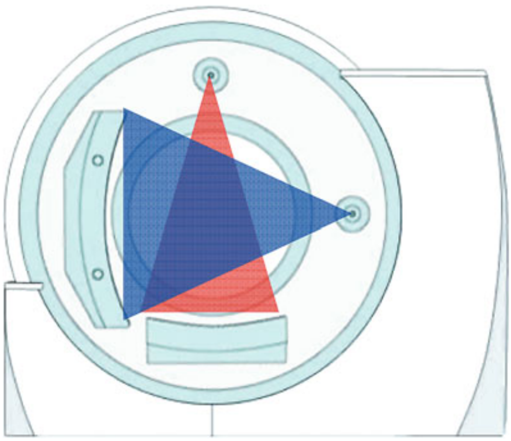

Fig. 1.3 Geometry of dual-source CT (Courtesy of Siemens AG, Muenchen, Germany)

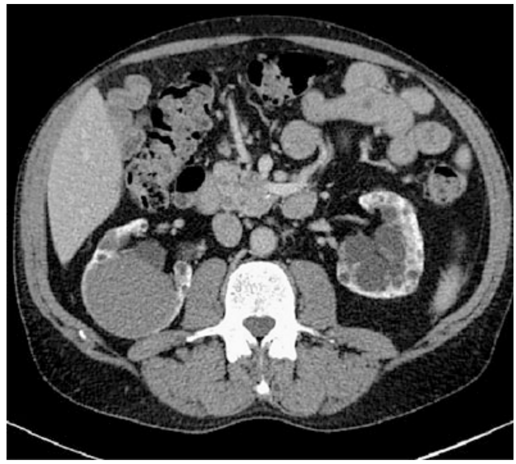

Fig. 1.4 Example of a mixed 140–80 kV image, resulting in a common 120 kV diagnostic image (Courtesy of Siemens AG, Muenchen, Germany)

This CT system allows almost simultaneous acquisition of two energy data sets, with very few artifacts due to image misregistration.

The acquired projection data primarily have to be reconstructed by standard filtered back-projection, separately for the two simultaneously acquired spiral data sets (80 and 140 kV). Subsequently, materials can be analyzed; this data analysis is called post-reconstruction analysis, meaning that different materials are analyzed after standard reconstruction for each tube. This system differs from material decomposition obtained by GE.

For abdominal imaging, dual-energy CT acquisitions should employ a collimation of 14 × 1.2 mm rather than 64 × 0.6 mm as the latter configuration will cause increased image noise on the B detector images. Since a reconstructed slice thickness below 1.2 mm is usually not required for most applications in the abdomen, this typically does not represent a significant limitation in terms of spatial resolution. However, the data acquired are not isotropic.

Each dual-energy acquisition can generate the following types of data: pure 80 kVp data, pure 140 kVp data, and a weighted average 120 kVp data set that usually is a composition of 70 % from the A (high kV) and 30 % from the B (low kV) tube (Fig. 1.4). This relation can be manually adjusted on a dual-energy workstation.

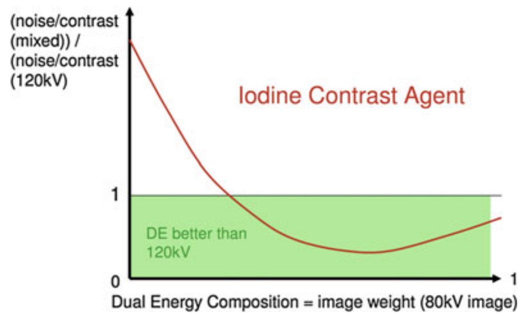

Fig. 1.5 When the contribution of 80 kV data in mixed images is correct (*left part*), high response of iodine to 80 kV tube current leads to better contrast to noise ratio than simple 120 kV standard images (Courtesy of Siemens AG, Muenchen, Germany)

Note that at same dose a dual energy mixed image has better contrast/noise ratio than 120 kV routine images, due to the contribution of 80 kV data, where the absorption peak of iodine is at its maximum (Fig. 1.5) [3].

Dual-source CT, when compared with a single-source system, presents two other possible applications. First, the tubes can be used at equal tube potentials, permitting increased photon flux in larger patients. Second, since an image can be acquired using 2 × 90 degrees of gantry rotation rather than 180, temporal resolution can be increased by a factor of two when using the two tubes at identical kVp levels [4]. Using this technique, a temporal resolution of 70 ms is possible; this is most useful for cardiac imaging.

1.3 Dose

To minimize patient exposure to ionizing radiation, abdominal dual-energy CT protocols operate using an online dose modulation system (CareDOSE 4D, Siemens) that adapts the tube current to the patient's anatomy [6]. The image quality reference mAs values are usually set to 400 mAs on the B tube and 96 mAs on the A tube, thereby splitting the energy between the two tubes. These settings take into account that higher mAs values on the A tube would lead to increased image noise on the B detector due to scatter radiation. The calculated effective patient doses for abdominal scans will range from 4.5–12.5 mSv, which is similar to the effective dose of a standard abdominal CT acquisition using 120 kVp with 250 mAs [4].

Dual-energy CT is not recommended for patients whose body mass index is >30. In morbidly obese patients, the two tubes can both be operated at 120 kVp, which will help to decrease image noise in these very large patients. The system can be used to scan patients with a body weight of up to 500 lbs (220 kg).

1.4 Limitations

A limitation of first-generation dual-source CT in the abdomen and pelvis is that the smaller size of the B detector will prevent imaging of the entire FOV in larger patients. Therefore, patients may have to be positioned off center if the location of the lesion is in the periphery of the FOV (for example renal masses). Therefore, it is mandatory to acquire two topograms. Moreover, objects at the outer periphery of the B detector may be unable to undergo optimal post-processing due to the technical specifications of the post-processing algorithm. In detail, adjacent voxels have to be used for calculation of the DE properties of any voxel within the field of view. The reconstructed DE field of view is 5 mm smaller than the actual B detector FOV.

Grosjean et al. have reported a significant impairment of the efficiency of DE analysis by motion during CT [7]. Thus, they conclude a perfect breath-hold during CT is essential [8].

1.5 Rapid kVp Switching or Gemstone Spectral Imaging

The CT system released by GE is a single X-ray source system that employs fast kVp switching for dual-energy acquisitions. It is called Gemstone spectral imaging (GSI), and it is based on projection-based material decomposition.

This technique enables precise temporal registration of views, freezing motion as the alternating spectrums penetrate the patient, thereby significantly reducing motion artifacts.

The generator and tube are capable of reliably switching between 80 and 140 kVp targets and have the capability to support sampling as quickly as every 140 microseconds.

The Gemstone detector is a key contributor to fast kVp switching acquisitions through its scintillator and data acquisition system (DAS). It is a complex rare earth based oxide, which has a chemically replicated garnet crystal structure. This lends itself to imaging that requires high light output, fast primary speed, very low afterglow, and almost undetectable radiation damage. Gemstone has a primary decay time of only 30 ns, making it 100 times faster than GOS (Gd2O2S), while also having afterglow levels that reach only 25 % of GOS levels making it ideal for fast sampling. The capabilities of the scintillator are matched with a fast sampling capability DAS, enabling simultaneous acquisition of low and high kVp sinograms.

In order to combat the traditional flux issues that have challenged fast kV switching, low-kVp and high-kVp acquisitions are flux balanced through advances in the DAS, which allow for dynamically changing view integration times. Additional time is allocated to the low-kVp acquisition relative to the high-kV acquisition in order to reduce photon starvation conditions.

Coupled with the appropriate rotation speed, a more balanced flux condition between the two kVp scans is achieved and serves to minimize patient dose.

1.6 Reconstruction

The overall spectrum is decomposed into a superposition of several known kVp spectra through the measurement of the detector response to attenuation.

Once a GSI acquisition is completed, the imaging reconstruction/processing chain, in combination with the GSI viewer, provides the user different kinds of images, such as conventional low-/high-kVp Hounsfield (HU) attenuation, material density and monochromatic keV HU representations of the data.

We commonly define the X-ray beam quality in terms of kVp (kilovoltage peak) denoting the maximum photon energy, since the X-ray beam is comprised of a mixture of X-ray photon energies (Fig. 1.6).

If we measure the X-ray attenuation of an object at two different spectrums, low kVp and high kVp, we can mathematically transform the attenuation measurements into the density of two materials that would be needed to produce the measured attenuation. This process is referred to as material decomposition or material separation.

Material decomposition does not identify materials. Rather, given two selected basis materials, material decomposition determines how much of each material would be needed to produce the observed low- and high-kVp measurements. Generally, low- and high-attenuating materials are selected as the basis pair. For diagnostic imaging, water and iodine are often used, since they span the atomic number range of materials generally found in medical imaging and approximate soft tissue and iodinated contrast, resulting in material density images that are intuitive to interpret.

Monochromatic images may be synthesized from the material density images. The monochromatic image depicts how the imaged object would look if the X-ray source produced only

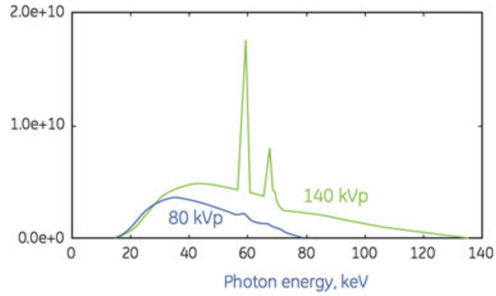

Fig. 1.6 Polychromatic X-ray spectrums for 80 and 140 kVp (Courtesy of GE Healthcare, Waukesha, U.S.A.)

X-ray photons at a single energy. Through this calculation, material decomposition enables the representation of the data as though it came from a monochromatic source. Monochromatic images are free of beam hardening and are a significant step toward quantitative imaging. These images remain subject to errors induced by scatter, aliasing and partial volume. Generally, the monochromatic image is similar to that of a conventional HU image with fewer artifacts. The monochromatic energy is selectable with higher energies yielding less contrast between materials and more contrast with low energies. GSI viewer enables selection of 101 keV energy levels virtual images (Figs. 1.7 and 1.8) [9].

1.7 Sequential Dual Energy

Sequential Dual Energy (SDE) released by Philips is a technique using two consecutive scans with different energies (iCT version 3.2). Technological requirements are more simple for this kind of acquisition, compared with Dual Source and GSI.

There are no factory default scanning protocols for Sequential Dual Energy; the user should create a tailored protocol; a dedicated software is available for reconstructions (EBW Spectral Analysis). Protocols include two Coupled Acquisitions with a minimal delay between them (Currently 2.1 s). There is a single axial rotation (<=8 cm coverage), without couch movement; there must be identical scan Length, FOV and scan parameters except for kV and mAs. Any

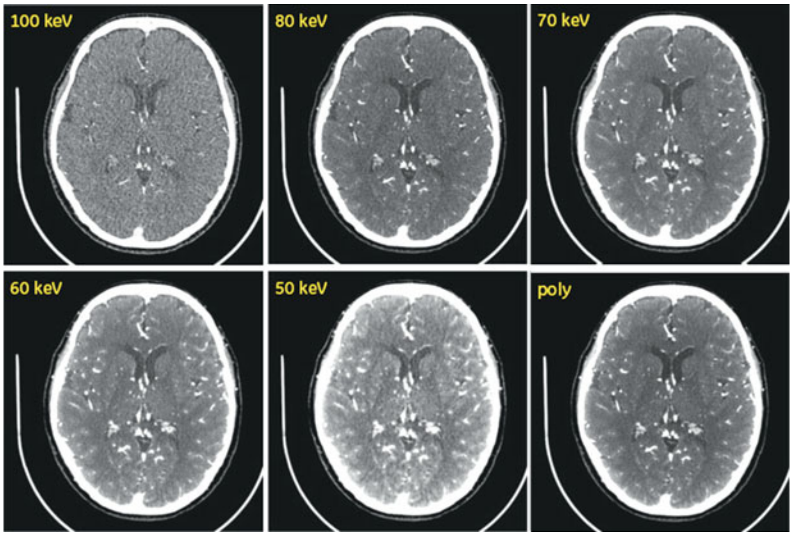

Fig. 1.7 Monochromatic renderings of a head image. Changes in noise and contrast can be seen (Courtesy of GE Healthcare, Waukesha, U.S.A.)

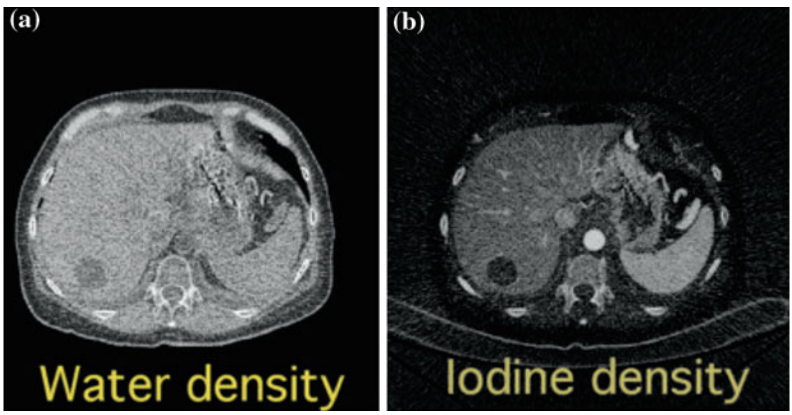

Fig. 1.8 Example of material decomposition with water (**a**) and iodine (**b**) images (Courtesy of GE Healthcare, Waukesha, U.S.A.)

two kV can be selected for SDE, but the best tissue differentiation can be achieved with a large kV difference (i.e., 140 and 80 kV), as well as for Dual Source and GSI. In order to have same image quality despite differences in kV, CTDI ratio should be 1; that means higher mAs with lower kV. After acquisition, the two scans must be aligned before analysis. A virtual mean kV image can be easily obtained from the two series. The workstation starts the analysis computing pixel values. Each pixel has two different values at high and low energy; the ratio of the two values corresponds to a specific material in an energy map.

These dual-energy series are easily achievable without a technological expense; they are also easily clinically feasible; but compared with Dual Source and GSI, they have some limitations: they are more subject to motion artifacts and can suffer from different iodine distribution at different time acquisition.

Philips implemented another experimental dual-energy system: a dual-layer detector scanner. In a single source, dual-layer detector scanner configuration, one X-ray tube is used to expose a detector consisting of two layers of scintillators. The two layers are directly on top of one another, and CT scan is performed at a high kVp. The first layer encountered by the X-ray photons absorbs most of the low-energy spectrum (approximately 50 % of the beam), while the bottom detector layer absorbs the remaining high-energy photons. Images are reconstructed separately from the data of the upper and lower layers. Since the spectral energy separation is intrinsic to the detection System, this approach eliminates the time lag of sequential techniques, making it ideal for imaging moving organs.

As in Dual Source and GSI systems, low- and high-energy images, water–iodine images can be easily obtained from data sets.

1.8 Multi-Energy CT

Multi-energy CT is not clinically applied yet, but there are some research studies giving interesting results.

In contrast to the scintillator detectors used in conventional and dual-energy CT systems, whose signal output is almost exclusively dependent on the energy flux integrated over the entire X-ray spectrum, spectral CT is based on new detector designs that may possess energy-sensitive photon-counting capability. This system allows the compartmentalization of detected X-ray photons into energy bins. Soft-tissue characterization and selective contrast material detection are subsequently achieved by recognizing tissue- and material-specific energy distributions [10].

As said before, the attenuation of X-rays is energy dependent and each substance has a specific attenuation curve [11]. Tissue characterization using X-ray is possible provided the absorption characteristics can be either measured or differentiated [12].

In dual-energy CT, materials can be differentiated by exposing the tissue to two different X-ray spectra or using a combination detector with two different energy ranges.

The disadvantages of this technique are that materials with similar attenuation curves such as iodine and barium cannot be distinguished, due to partially overlapping spectra [12].

In spectral CT imaging, a fixed broad spectrum X-ray source is used (such as used in conventional CT) combined with an energy discriminating detector that can differentiate the attenuation spectrum of all materials within the voxel [13, 14].

New developments in direct-conversion semiconductor detectors based on CZT (Cadmium–Zinc–Telluride) or CdTe (Cadmium Telluride) may enable the application of this technique also for medical CT imaging.

A prerequisite for the spectral analysis is the knowledge of the spectral response of the detector.

X-ray detector is operated in photon-counting mode applying energy discrimination by threshold values for each individual pixel. Physically, the thresholds are realized by voltages that are fed into the pulse-height comparator circuits. The pulse height obtained from the detector is nearly proportional to the energy of the detected photon, but gain and offset vary from bin to bin and from pixel to pixel. Therefore, the relationship between the measured photon energy in keV and the threshold voltage (in mV), which triggers the corresponding event, had to be determined through a calibration process.

As reported by Anderson et al. the photon-counting Medipix2 [15] and Medipix3 [16] detectors developed by the European Centre for Nuclear Research (CERN) allow up to eight points of an attenuation spectrum to be reconstructed from a single exposure without overlap in the measurement.

Other research studies are conducted by Philips [17]. Spectral CT imaging is performed with an experimental single-section CT scanner (Multi-Energy Philips Photon-Counting CT; Research Europe) that allows differentiation of detected X-ray photons into six distinct energy bins . The X-ray beam passes through the beryllium exit window of the X-ray tube and enters the detector through a 2-mm aluminum window; no further spectral filtering is applied. The photon-counting detector is based on a cadmium telluride array (MEXC; Gamma Medica Ideas, Northridge, Calif) that allows the simultaneous differentiation of detected X-ray photons according to their individual energy levels by separating the X-ray photons into six energy bins at 25–34, 34–39, 39–44, 45–49, 49–55, and more than 55 keV. These threshold levels are set prior to the initiation of spectral

CT imaging. The energy separation is not sharply delineated; instead, a smooth transition between the energy bands exists. Gantry rotation time is 120 s because of the count rate limitations of the detector system and the need to enable good photon statistics (approximately 2,000 projections per gantry revolution). For each energy bin, 0.5-mm axial sections with a matrix of 512 × 512 pixels within a 20-mm field of view are reconstructed without the additional application of reconstruction filters. The attenuation units attained with this prototype spectral CT scanner are all calibrated with a water phantom as fixed, meaning, however (in contrast to clinical Hounsfield units), that they are directly dependent on detector counts [17].

In the absence of K-edge discontinuities, the energy dependence of the linear attenuation coefficient of materials of diagnostic interest can be described accurately by a linear combination of the photo-electric and the Compton effects.

In the presence of elements with a high atomic number Z, the description of the attenuation properties of matter has to be modified. In order to correctly describe the attenuation of a sample containing a single element with K-edge discontinuity inside the relevant energy range, the decomposition has to be extended by the energy dependent attenuation function of this particular element as a third component.

In order to observe the K-edge discontinuity, it is important that the transmission spectrum provides sufficient power at lower as well as at higher X-ray energies. In the case of strongly attenuating objects, the effects of beam hardening will eventually remove power in the low-energy part of the spectrum to the extent that the discontinuity is no longer observable due to photon starvation.

It is possible, in principle, to detect k edges of lower *Z* elements, but it has to be considered that photon starvation is a serious problem in the low-energy range. Therefore, gadolinium (*Z* 64; k edge, 50.2 keV) is more promising as a spectrally detectable contrast agent than, for example, iodine (*Z* 53; k edge, 33.2 keV) [18].

For effective material decomposition, the energy bins used for material decomposition should be sufficiently narrow and well separated. However, when narrow bins are used, a large fraction of the detected X-ray counts is lost and statistical noise is increased. Alternatively, the X-ray spectrum can be split into a few larger bins with no gap in between and all detected X-ray photons can be used for material decomposition. However, in this case, the energy bins are too wide and not well separated, which results in suboptimal material decomposition. The above contradictory requirements can be resolved if the X-ray photons are physically removed from the regions of the energy spectrum between the energy bins. Such a selective removal can be performed using filtration of the X-ray beam by high-Z filter materials with appropriate positions of K-edge energies [14].

In conclusion, dual energy is clinically feasible and it has brought some advance in material differentiation, image quality and diagnostic confidence.

There are several research studies investigating multi-energy CT, in order to achieve detailed tissue differentiation. Maybe some of these results will be applied to routine clinical practice, improving diagnostic performance.

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