Chapter 4

Techniques to Acquire Spectral CT Data

There are several ways to acquire CT data with spectral information. Today's medical CT systems are typically equipped with solid state detectors. A scintillation crystal absorbs the x rays and converts them into visible light, which is then detected by an attached Si photodiode. These detectors integrate the x-ray flux over a certain period of time (the reading time), with a weighting factor proportional to the energy of the absorbed x-ray quanta (see Section 2.3 for a detailed description). As a consequence, the detectors do not provide energy resolution, and the use of different x-ray spectra is needed to acquire spectral CT data. In CT, different x-ray spectra can be realized by using different kV settings of the x-ray tube. In a standard CT system, the kV setting of the x-ray tube can be changed either between different CT scans (slow kV switching), or more rapidly, ideally, between the different projections of a CT scan (fast kV switching). An alternative is the use of DSCT systems with two x-ray tubes and two corresponding detectors offset by about 90 deg. These systems have the potential to acquire dual-energy data by operating both x-ray tubes at different kV settings.

Energy-resolving detectors enable the acquisition of spectral CT data with a single polychromatic x-ray spectrum. Pertinent examples are dual-layer detectors consisting of two conventional scintillation detectors on top of each other, and direct converting photon-counting detectors. So far, only prototype CT systems relying on both detector technologies have been realized. However, in particular, photon-counting detectors are a promising technology for future CT systems. In this chapter we will discuss the different technologies for acquiring spectrally resolved CT data.

4.1 Use of Different X-Ray Spectra

4.1.1 Slow kV switching

The most straightforward approach to acquire dual-energy CT data is the use of different kV settings for the x-ray tube of the respective CT scanner.

In a simple technical realization, two consecutive axial scans of the same anatomy are performed, one with low tube potential and the other with high tube potential. To reduce the time shift between both scans, the scans can be performed as partial scans, with an angular coverage of 180 deg plus the total detector fan angle (see Fig. 4.1).

Another simple method uses two consecutive spiral/helical scans of the same anatomy at different kV settings and preferably at high pitch for short overall acquisition times. Most commonly, 80 and 140 kV are used because these are typically the lowest and highest CT x-ray tube kV settings that provide best spectral separation. The mean energy of the 80(140) kV spectrum is typically 52(69) keV. If an integrating scintillation detector is used, higher-energy photons dominate the signal, and the mean energy of the power spectrum shifts to 54(76) keV. The energy spectrum at both kV settings is rather broad (Fig. 4.2) and ranges from about 35 keV to electron charge *e* times the tube voltage (kV). The photon energy is thus above the K-edge of all common elements in a human body, including iodinated contrast agent (the K-edge of iodine is at 33 keV).

Using the slow kV switching approach, spectral optimization is possible, e.g., by moving an additional filter into the beam when it is switched to the high kV setting. Suitable materials will be discussed in Section 4.1.3. Spectrally resolved CT data can be acquired with standard CT systems in the full scan field of view [(SFOV), which is typically 50 cm in diameter)] of the respective CT detectors. This allows the use of dual-energy scanning for larger patients and off-center anatomy. Equal dose at high and low kV, a dose that improves the results of dual-energy evaluation algorithms, can be obtained by an adaptation of the mA settings at 80 and 140 kV. For equal dose, the mA at 80 kV needs to be a factor of about 3 higher than the mA at 140 kV. Another method uses longer acquisition times at 80 kV, e.g., by reducing the spiral pitch to values well below 1 to accumulate radiation dose in the images by overlapping data acquisition.

As a downside, the long time interval of at least one-half second between the two scans hampers the evaluation of moving organs.

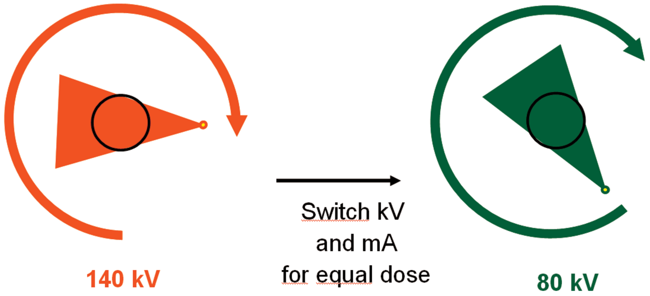

Figure 4.1 Principle of slow kV switching to acquire dual-energy CT data.

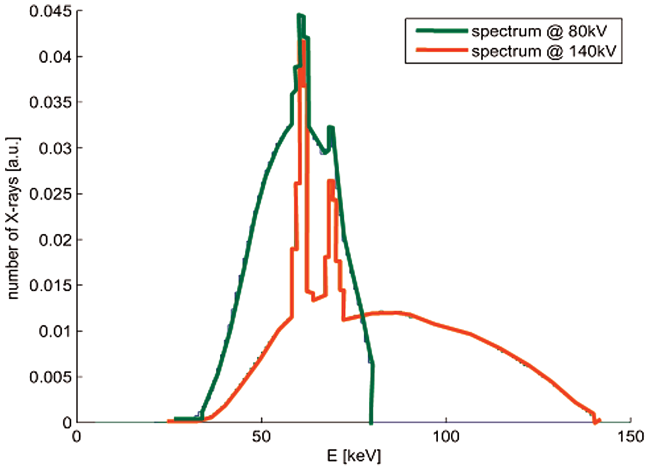

Figure 4.2 Typical CT x-ray spectra at 80- and 140-kV x-ray tube voltage, scaled to equal tube output at both kilovolt settings. Note that there is no spectral overlap in the energy range from 80 to 140 keV.

Figure 4.3 shows two axial scans of a pig at 80 and 140 kV after iodinated contrast agent was applied. The simple difference image of both scans (Fig. 4.3) shows severe motion artifacts, which can be reduced but not eliminated by means of nonrigid anatomical registration algorithms. These algorithms warp the contours of one image (e.g., the low-kV image) to match the contours of the other (e.g., the high-kV image).

In CT scans using iodinated contrast agent, the rapid blood flow dynamics in early contrast phases can cause local variations of the contrast agent densities in the two CT scans at different x-ray tube voltages, possibly leading to inconsistent differences in CT numbers (HUs) and wrong interpretation by dual-energy evaluation algorithms (see Fig. 4.4). Meaningful contrast-enhanced scans using slow kV switching therefore need to be restricted to late contrast phases, when contrast equilibrium has been reached.

In clinical practice, slow kV switching can be used for dual-energy imaging in “static” situations, e.g., for the characterization of different types of kidney stones, for the differential diagnosis of gout, or for the calculation of monoenergetic images to reduce metal artifacts at a metal-specific high energy (see Fig. 4.5).

In scans with contrast agent, differentiation between bone and iodine-filled vessels for an automatic removal of bones in CT angiographic studies seems to be feasible if the first low-kV scan is performed as fast as possible in an early contrast phase, while the second high-kV scan is performed at a later phase. As an additional application, the computation of virtual unenhanced images or iodine maps in later venous contrast phases may be

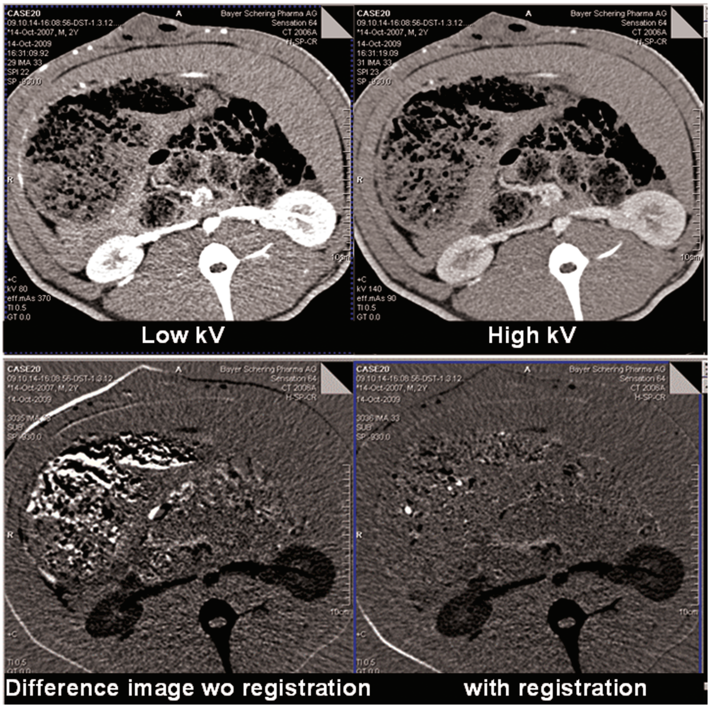

Figure 4.3 Slow kV switching. Two consecutive contrast-enhanced axial scans of a pig, at 80 kV (top left) and at 140 kV (top right). Simple difference image (high–low kV) shows severe artifacts caused by motion (bottom left); difference image after nonrigid registration shows reduced artifacts (bottom right).

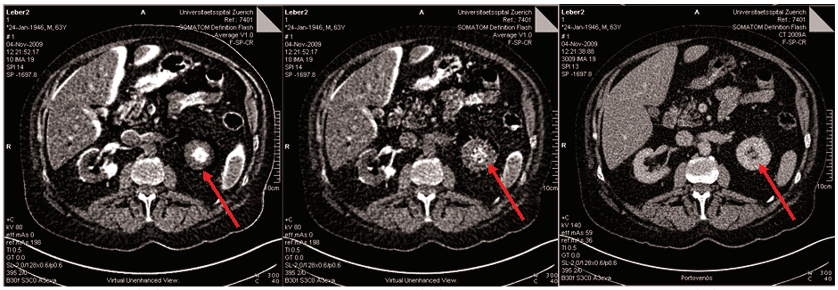
(a) (b) (c)

Figure 4.4 Slow kV switching. (a) Virtual noncontrast image obtained by dual-energy-based iodine removal from two consecutive contrast-enhanced axial scans of a pig at 80 and 140 kV, without registration. (b) Virtual noncontrast image after nonrigid registration. (c) True noncontrast image. Note the incompletely removed iodine in the virtual noncontrast images (red arrow) as a consequence of motion and rapidly changing contrast density.

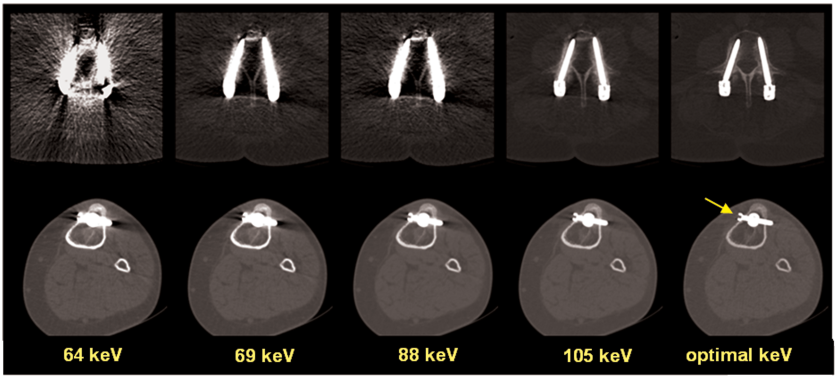

Figure 4.5 Example of a clinically relevant application of slow kV switching. Two consecutive axial scans at 80 and 140 kV are used to compute pseudo-monoenergetic images, with the goal of metal-artifact reduction at a metal-specific high energy.

feasible. Slow kV switching has meanwhile been commercially introduced by several vendors as a simple method to obtain dual-energy data for selected clinical applications (see Fig. 4.6).

4.1.2 Rapid kV switching

In a more refined approach, the kV setting of the x-ray tube is rapidly switched between consecutive projections (views) of the same axial or

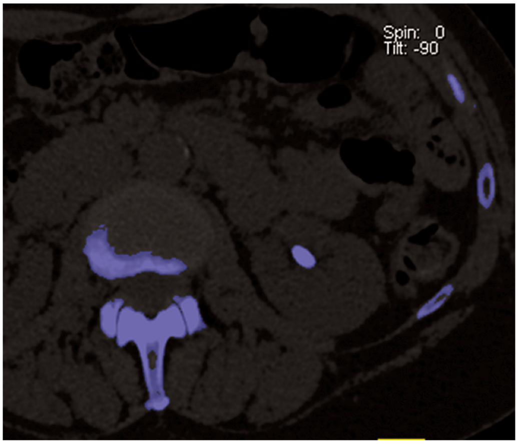

Figure 4.6 Slow kV switching for the characterization of kidney stones. Two consecutive spiral scans at 80 and 140 kV were performed with a 128-slice single-source CT scanner at 0.6-mm collimated slice width. Both low- and high-kV scans were nonrigidly registered. Calcium is highlighted in blue; the observed kidney stone is therefore considered to be calcified. (Courtesy of CIC, Mayo Clinic Rochester, MN, USA.)