Principles of Multi-slice Cardiac CT Imaging

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4.1 Basic Performance Requirements for CT Imaging of the Heart

To virtually freeze cardiac motion and avoid motion-artifacts, very short exposure times are needed for the acquisition of transaxial slices. High temporal resolution is particularly important for imaging the coronary arteries, as they are located very close to the heart muscles and show strong movement during the cardiac cycle. Due to the very complex 3D motion pattern of the heart, the intensity of movement varies for different cardiac anatomies and different coronary vessels, and within the cardiac cycle. The strongest movement is present during contraction of the atria and ventricles in systole. The short end-systolic rest phase is followed by a continuous-filling phase of the ventricles during diastole that slows down towards mid- and end-diastole. This motion pattern can be measured by the dislocation of the left ventricular wall, the aortic valve flaps, and the different segments of the coronary arteries in representative transaxial planes (Fig. 4.1a) (Achenbach 2000). The least amount of movement of the major cardiac anatomy and the coronary arteries, and thus the least amount of dislocation over time, is observed in end-systole and mid- to end-diastole of the cardiac cycle (Fig. 4.1a). While the duration of the end-systolic phase (approximately 100–150 ms) is more or less independent of the present heart rate (and the related RR-interval time), the duration of the phase with the least cardiac motion during diastole narrows with increasing heart rate (Fig. 4.1b). For low heart rates, the end-systolic phase is much shorter than the diastolic rest phase, but the difference decreases with increasing heart rate. For
higher heart rates, the diastolic rest phase can be even shorter than the end-systolic rest phase.

The cardiac anatomy should be imaged during those phases of cardiac cycle in which there is the least movement; that is, when the least blurring due to motion artifacts is to be expected. Thus, image acquisition and reconstruction need to be synchronized as accurately as possible to the movement of the heart, i.e., by using ECG information that is recorded in parallel to the CT scan acquisition. Since the fastest gantry rotation speeds of today’s multislice CT scanners are between 0.5 s up to 0.33 s per rotation, imaging of the heart is usually performed during the diastolic rest phase of the cardiac cycle, as the achievable temporal resolution is usually not sufficient to reliably eliminate cardiac motion during other phases of the cardiac cycle. However, with CT gantry rotation speeds becoming even faster, imaging in the end-systolic rest phase may also become feasible.

According to a rough estimation, a temporal resolution of about 250 ms is appropriate for motion-free imaging in the diastolic rest phase up to a heart rate of about 60 bpm, about 200 ms up to a heart rate of 70 bpm, and approximately 150 ms for clinically usual heart rates up to 90 bpm (Fig. 4.1b). It can be expected that a temporal resolution of about 100 ms is sufficient for imaging the heart during the diastolic or end-systolic rest phase also at high heart rates (Achenbach 2000). Image acquisition during other phases of the cardiac cycle (e.g., systole) with rapid cardiac motion is needed for evaluation of cardiac function. Motion-free imaging during phases other than the diastolic and end-systolic rest phases requires a temporal resolution of about 50 ms (Stehling 1991) and is usually not possible with state-of-the-art CT systems. Thus, only the larger cardiac anatomy or cardiac anatomy with low motion amplitudes should be assessed during phases of high cardiac motion.

Many structures of the cardiac morphology, especially the coronary arteries and the cardiac valves including the valve flaps, represent small and complex 3D structures that require very high and, at best, sub-millimeter isotropic spatial resolution with longitudinal resolution close or equal to in-plane resolution. The most proximal coronary segments and the distal segments of the right coronary artery (RCA) are directed parallel to the image plane, while the middle segments are directed
perpendicular to the image plane. The lumen diameter of the main segments of the coronary artery tree ranges from 4 mm (left main coronary artery) down to about 1 mm (peripheral left main coronary artery and circumflex). Detection and quantification of coronary stenosis with the ability to differentiate a 20% change in the diameter of the vessel lumen for the larger-caliber vessels represents a viable goal for cardiac CT imaging. To achieve this, CT systems need to provide a spatial resolution in all three dimensions (isotropic) of at least 0.5 mm for visualization of the main coronary vessels and of smaller branches. Thus, in-plane and through-plane resolution significantly below 1 mm are needed to assess the main coronary segments, including narrowing and plaques. Less spatial resolution is sufficient for assessment of the larger cardiac anatomy, such as the myocardium and the cardiac chambers. However, scans should a priori be acquired at best possible spatial resolution, as lower spatial resolution can be generated via data post-processing.

In addition to high spatial resolution, sufficient contrast-to-noise ratio is important to resolve small and low-contrast structures, such as atherosclerotic coronary plaques, with different attenuation properties. Appropriate low-contrast resolution has to be provided with limited radiation exposure at the shortest possible exposure time. For most cardiac applications, appropriate contrast enhancement of the cardiac vessels and the cardiac anatomy is achieved with peripheral injection of contrast agent and optimized timing of the bolus. Sufficient contrast enhancement is particularly important for imaging the distal coronary segments, as they are located very close to the myocardium and are separated from it only by a thin layer of epicardial fat. Only the assessment of cardiac and coronary calcification is possible without administration of contrast agents.

Scan acquisition within a single and short breath-hold time is mandatory for minimizing the amount of contrast material needed for vascular enhancement and to avoid respiratory artifacts. Breath-hold times of 20 s or less are appropriate for stable patients but 10 s or less are preferable for less stable, dyspneic patients. Ideally, a complete data set of the entire heart anatomy would be acquired within a single phase of the cardiac cycle without patient movement, but CT technology today, and presumably in the near future, does not provide sufficient detector width to cover the entire heart volume within a single heartbeat. Instead, images at consecutive z-positions that continuously cover the heart volume need to be generated from data acquired in different cardiac cycles. A virtually «frozen» cardiac volume can only be produced by phase-consistent synchronization of the acquisition to the movement of the heart using simultaneously recorded ECG information. A reasonably stable sinus rhythm without rapid arrhythmic changes provides the best 3D images of the cardiac anatomy. Functional information can be derived if cardiac volume images can be generated in different phases of the cardiac cycle, e.g., as a basis for quantitative evaluation.

Cardiac imaging is obviously a highly demanding application for CT, since the temporal, spatial, contrast resolutions as well as scan time have to be optimized simultaneously, and radiation exposure has to be limited to levels of related imaging modalities, such as invasive diagnostic coronary angiography or nuclear imaging studies. These conflicting performance requirements have to be fulfilled and optimized for the particular application at the same time and within the same cardiac scan protocol. Optimization of one performance cornerstone alone (e.g., high temporal resolution, low radiation exposure) by trading-off other important parameters (e.g., spatial and contrast resolutions) may not lead to clinically useful results.

4.2 CT Imaging with Optimized Temporal Resolution: The Principle of Half-Scan Reconstruction

For motion-free imaging of the heart, data have to be acquired during phases of the cardiac cycle with little cardiac motion and with as high as possible temporal resolution. In the first place, temporal resolution of an axial CT slice is determined by the exposure time associated with the scan data used for reconstruction of that CT slice. Therefore, scan techniques and reconstruction algorithms need
to be developed that use a minimum amount of scan data while maintaining high image quality. In modern 3rd-generation CT systems with fan-beam geometry, the minimum amount of scan data needed for reconstruction of axial slices is a so-called partial scan (Parker 1982). Depending on the exact system geometry and its dimensions, a partial-scan fan-beam data set has to cover a projection-angle interval $\alpha_p$ (angle interval between tube positions at the start and end points of tube rotation) of 180° plus the breadth of the X-ray fan: $\alpha_p = \pi + \beta_f$. The breadth of the X-ray fan-beam $\beta_f$ depends strongly on the diameter of the scan field of view (usually 50 cm) and the distances of the focal spot and detector from the center of the scan field of view. The equation $\alpha_p = \pi + \beta_f$ states that a minimum data segment of 180° has to be available for every fan angle $\beta$. As a consequence, a partial rotation usually covers about two-thirds of a rotation ($\approx 240°$) (Fig. 4.2a). Conventional partial-scan reconstruction techniques based on fan-beam geometry (Parker 1982) make use of all acquired data even if more data than the minimum angle of 180° are available for a fan angle $\beta$. These techniques produce a temporal resolution equal to the partial-scan acquisition time ($\approx$ two-thirds of the rotation time).

Better temporal resolution can be achieved with special reconstruction algorithms that use the minimum required amount of scan data. These algorithms, referred to as "half-scan" reconstruction, can be best explained using parallel-beam geometry. To this end, the fan-beam geometry of the partial-scan data set is transformed to parallel-beam

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**Fig. 4.2a, b.** Principle of “half-scan” image reconstruction based on parallel-beam scan data. Scan data acquired in fan-beam geometry (a) with fan-beam projection angle $\alpha$ and fan-ray angle $\beta$ is transformed into parallel-beam geometry (b) with parallel-beam projection angle $\Theta$ and parallel-ray position $p$ using the “rebinning” technique and 2D interpolations. Acquired fan-beam data that form incomplete parallel projections are omitted during reconstruction. A data transition range enables smooth transition weighing for reduction of artifacts due to data inconsistencies.
geometry using "rebinning" techniques (Kak 1988). The rebinning of a partial-scan fan-beam data set provides 180° of complete parallel projections, including chunks of incomplete parallel projections that consist of redundant data. Fan-beam data are described by angle Θ of the tube position and angle β of a certain ray within the fan-beam projection (Fig. 4.2a). Correspondingly, parallel data can be identified with angle Θ for the projection angle and position p of the rays within the parallel-beam projection (Fig. 4.2b). Fan-beam data \( P_\beta(\alpha, \beta) \) are transformed into parallel-beam data \( P_p(\Theta, p) \) based on:

\[
P_p(\Theta, p) = P_\beta(\alpha + \beta - \pi/2, RF \sin \beta) \text{ for } \Theta = 0...\pi,
p = -D/2...D/2
\]

(4.1)

\( RF \) represents the distance of the X-ray source to the center of rotation of the measurement system, and D the diameter of the scan field of view. As all projection data are digitally processed and therefore digitally sampled, continuous transformation requires simple interpolation procedures. The geometrical transformation is illustrated in Figure 4.2. An image can be reconstructed from parallel projections that cover a parallel-projection angle range of 180°. If the extra data sectors with incomplete parallel projections at the start and end positions of the X-ray source vary with the rotation time in a centered region of the scan field of view (e.g. 250 ms for 500-ms rotation time, 200 ms for 400-ms rotation time, and 165 ms for 330-ms rotation time). It shows an approximately linear decrease between the edges of the scan field of view, with a gradient that is perpendicular to the connection line of the start and end position of the source (Fig. 4.3a). Different temporal data windows visualized by the paths of different image points within the scan field. The temporal resolution equals half the rotation time in a centered region of the scan field of view (e.g. 250 ms for 500-ms rotation time, 200 ms for 400-ms rotation time, and 165 ms for 330-ms rotation time). It shows an approximately linear decrease between the edges of the scan field of view, with a gradient that is perpendicular to the connection line of the start and end position of the source (Fig. 4.3a).

However, half-scan reconstruction is prone to streak-type artifacts due to inconsistencies of the projections \( P_p(0, p) \) and \( P_p(\pi, p) \) at the beginning and end of the data set that may be produced by object motion between the acquisition of these projections. This effect can be substantially reduced by acquisition of a slightly extended range of fan-beam projections \( \alpha_p = \pi + \beta_1 + \alpha_T \). The additional projection range \( \alpha_T \) transforms into an extra range of parallel-beam projections \( \Theta_T = \alpha_T \) for smooth transition weighting of the projections at the start and end of the parallel-beam data set:

\[
P'_p(\Theta, p) = W(\Theta) P_p(\Theta, p) + (1-W(\Theta)) P_p(\Theta+\pi, p') \text{ for } 
\]

\[
\Theta = 0...\Theta_T, \quad p = -D/2...D/2
\]

(4.2)

\[
P'_p(\Theta, p) = P_p(\Theta, p) \text{ for } 
\]

\[
\Theta = \Theta_T...\pi, \quad p = -D/2...D/2
\]

\( \alpha_T = -p \) represents the so-called complementary ray.

The function \( W(\Theta) \) (e.g. \( W(\Theta) = \sin^2(\pi\Theta/2\Theta_T) \)) represents a smooth, monotone transition between 0 and 1 within the angle range 0...\( \Theta_T \). We used a transition range of \( \Theta_T = 28° \) with a total scan angle of \( \alpha_T = 260° \). This type of transition weighting does not affect temporal resolution but can effectively suppress streak-artifacts in a reconstruction of the parallel-beam data set \( P'_p(\Theta, p) \).

The temporal resolution that is present at a particular position within the scan field of view is determined by the temporal window of the data that contributes to the reconstruction of that particular image point. The temporal resolution is represented by time sensitivity profiles (Hu 2000) that vary with position in the scan field of view (similar to slice sensitivity profiles for spiral reconstruction). We used the full-width at half maximum (FWHM) of the time sensitivity profile as a measure to determine the distribution of the temporal resolution within the scan field of view produced by our reconstruction algorithm. The temporal resolution equals half the rotation time in a centered region of the scan field of view (e.g. 250 ms for 500-ms rotation time, 200 ms for 400-ms rotation time, and 165 ms for 330-ms rotation time). It shows an approximately linear decrease between the edges of the scan field of view, with a gradient that is perpendicular to the connection line of the start and end position of the source (Fig. 4.3a). Different temporal data windows visualized by the paths of different image points within the scan field.

In all modern multi-slice CT scanners, the half-scan reconstruction technique is the method of choice for image reconstruction in cardiac applications. In clinical applications, the heart should be sufficiently centered within the scan field of view in order to maintain a consistent and stable temporal resolution of about half the rotation time. It is not possible to make clinical use of the areas with better temporal resolution within the scan field of view because the start and end positions of the X-ray source vary strongly for individual slices during ECG-synchronized acquisition and, as they depend on the patient’s heart rate, can therefore not be fixed.
Prospectively ECG-Triggered Multi-slice CT

Prospectively ECG triggering is an established method to synchronize sequential CT scanning using the partial-scan technique to the motion of the heart in order to acquire data in a certain phase of the cardiac cycle, preferably in the diastolic phase, when cardiac motion is minimal.

For ECG-triggered sequential imaging, a prospective trigger is derived from the ECG trace to initiate the CT scan with a certain, user-selectable delay time after the R-wave. For best temporal resolution, a partial scan is acquired that is reconstructed with the half-scan algorithm described above. After every scan, the table moves by the width of the acquired scan range in the z-direction towards the next scan position in order to provide gap-less volume coverage. The delay time for scan acquisition after an R-wave is calculated from a given phase parameter (e.g., a percentage of the RR-interval time as delay after an R-wave) for each cardiac cycle and is individually based on a prospective estimation of the RR-intervals. Usually, the delay is defined such that the scans are acquired during the diastolic phase of the heart.

Prospectively ECG-triggered sequential scanning has already been used with electron-beam (EBCT) (Stanford 1992) and mechanical single-slice CT (Becker 1999). In both techniques, single slices are acquired in consecutive heartbeats within equivalent phases of the cardiac cycle. The total scan time is related to the heart rate of the patient and is often too long for volume coverage with thin slices within a single breath-hold. Multi-slice CT scanners can
acquire multiple slices within one heartbeat. Thus, they offer shorter scan times that can be beneficial for clinical applications.

The first generation of multi-slice CT scanners, introduced in 1998, allowed for simultaneous acquisition of up to four adjacent slices per prospective ECG trigger for sequential coverage of the heart volume (Fig. 4.4a). Newer, multi-slice CT scanners acquire up to 16 slices simultaneously per heart beat and provide shorter rotation times and shorter acquisition windows per heart cycle (Fig. 4.4b). The latest generation of multi-slice CT scanners, with up to 64 slices per rotation, often do not use the maximum possible amount of slices. The maximum number of simultaneously acquired slices in sequential scan mode of such scanners is usually limited to 24–32 slices, due to increasing cone-beam artifacts that cannot easily be compensated by sequential scanning. With an increasing number of slices, more scan volume can be covered with every individual scan. This translates either into a significantly shorter acquisition time for the total scan range or the ability to acquire a given scan range with thinner slices in comfortable breathhold times (Fig. 4.4b).

The concept of coverage of the entire heart with sequential ECG-triggered scans is illustrated in Figure 4.5. The exact scan time for such coverage heart depends on the patient’s individual heart rate, as the RR-interval time determines the time between the individual scans. In this regard, an additional technical effect has to be considered. As the patient table has to move by the distance covered by multiple slices in between the scans, a certain technical delay time (so-called scan-cycle time) has to be taken into consideration before a following scan can be initiated. The minimum scan-cycle time between two consecutive ECG-triggered scans depends on the table-feed between the two scans and on the acquisition time of one scan. Usual scan-cycle times of modern multi-slice CT scanners are in the range of 0.8–1.5 s; thus, one heart beat has to be skipped in between every scan for usual clinical examinations at heart rates between 50 and 90 bpm with RR-interval times between 0.7 and 1.2 s. In the example of a 16-slice scanner presented in Figure 4.4, the entire heart is scanned within the time of 15 heart beats and reconstructed from data that are acquired during eight heart beats.

The two different filter techniques that are commonly used for prospective estimation of the position of the following R-wave are mean filtering (e.g., 3 previous RR-intervals) and median filtering (e.g., 5 previous RR-intervals). The median filter approach shows increased robustness for patients with moderate arrhythmia due to the fact that single extra beats are eliminated.

To diagnose dynamic processes, ECG-triggered acquisition can also be done without table-feed in between the scans. The same volume is then acquired in corresponding phases of consecutive heartbeats. As no table-feed is needed, the scan-cycle time is reduced and scans can usually be acquired within every heart beat for normal heart rates (scan-cycle time 0.5–0.8 s).

Prospectively ECG-triggered multi-slice CT acquisition results in significantly faster volume coverage than obtained with ECG-triggered acquisition with single-slice mechanical CT or EBCT. The acquisition of a small volume with each triggered scan reduces the probability of misregistration of lesions that can occur due to significant motion of the heart in the z-direction. With 4-slice CT and a collimated slice-width of $4 \times 2.5$ mm, the complete heart can be scanned within a comfortable single breath-hold of 15–20 s. With this performance, prospective ECG triggering is feasible for quantification of coronary calcification (Fig. 4.6a) and CT angiographic imaging of larger cardiac and cardio-thoracic anatomy, such as the myocardium, cardiac valves, thoracic aorta, and proximal segments of coronary bypass grafts (Fig. 4.6b). With modern 16- and 64-slice CT, breath-hold times of ECG-triggered acquisition can be further reduced and the use of thin collimation becomes feasible. The increased rotation speed of up to 0.33 s per rotation provides increased temporal resolution up to 165 ms for significantly improved robustness compared to 4- and 8-slice CT, also at higher heart rates. Furthermore, with 16- and 64-slice CT scanners prospective ECG-triggered scanning can be used for quantification of coronary calcification (Fig. 4.6c) as well as examination of the general cardiac anatomy and coronary bypass grafts (Fig. 4.6d). However, the quality of contrast-enhanced imaging of smaller cardiac anatomy and the coronary arteries with ECG triggering is limited, since the longitudinal spatial resolution
Fig. 4.4a, b. Sequential volume coverage with prospectively ECG-triggered 4-slice (a) and 16-slice (b) acquisition. Four and 16 adjacent images, respectively, with a slice thickness depending on the collimation (hatched blocks) are acquired at a time with $T_{rot}/2$ temporal resolution. Due to the limitation of the scan-cycle time, a scan can be acquired every other heart cycle for usual heart rates. Faster volume coverage and thinner slices are provided by 16-slice CT.
is compromised by the purely sequential, non-overlapping slice acquisition. In addition, small changes in heart rate during the scan can cause acquisition in inconsistent heart phases and thus inconsistent volume coverage, resulting in artifacts at the intersections of adjacent image stacks.

4.4 · Retrospectively ECG-Gated Multi-slice CT

Spiral CT acquisition with continuous table movement represented a very important step towards true volumetric imaging with mechanical CT. With spiral scanning, a true 3D data set can be acquired that consists of overlapping transaxial image slices (Kalender 1995). Additional advantages of spiral CT acquisition compared to sequential CT scanning are considerably improved spatial z-resolution, faster scan speed, and larger volume coverage. ECG-triggered acquisition techniques are limited to sequential scan modes and cannot be applied to spiral scanning.

Retrospectively ECG-gated spiral scanning is an attempt to synchronize the reconstruction of a continuous spiral scan to the movement of the heart by using an ECG trace that is recorded simultaneously. The acquired scan data are selected for image reconstruction with respect to a predefined cardiac phase with a certain temporal relation to the onset of the R-waves that defines the start point of data used for image reconstruction.

Although retrospectively ECG-gated spiral scanning was introduced for sub-second single-slice spiral CT systems, its feasibility in clinical routine could not be demonstrated due to various limitations. To obtain the best temporal resolution, partial-scan-based reconstruction techniques are applied to the spiral scan data (Kachelriess 1998, Bahner 1999). However, no spiral interpolation algorithms could be used for retrospectively ECG-gated single-slice CT and data inconsistencies due to table movement and spiral artifacts were very common in transaxial slices. Slow table-feed is a precondition...
Fig. 4.6a–d. Case examples acquired with prospectively ECG-triggered 4-slice CT with a 0.5-s rotation time (a, b) and 64-slice CT with a 0.33-s rotation time (c, d). With ECG-triggered 4-slice CT, $4 \times 2.5$-mm collimation, 120 kV, and 100 mA tube current, coronary calcifications in the circumflex and right coronary arteries can be detected with high sensitivity and high signal-to-noise ratio (arrows in a). Cardiac morphology and bypass patency can be evaluated with contrast-enhanced scanning using $4 \times 2.5$-mm collimation, 120 kV, and 200 mA tube current. In the example, a coronary bypass occlusion can be appreciated in the transaxial slices (arrow in b). With ECG-triggered 64-slice CT, 3- and 1.2-mm-thick slices can be generated from a $30 \times 0.6$-mm collimation setting. ECG-triggered 64-slice CT with 3-mm slices, 120 kV, and 126 mA tube current is used for coronary calcium quantification (arrow in c). For contrast enhanced imaging of the cardiothoracic morphology, ECG-triggered 64-slice CT with 1.2-mm slices, 120 kV, and 250 mA can be used (d). Based on the fast rotation of 0.33 s and the underlying thin-slice scan acquisition with 0.6-mm collimation, 64-slice CT scanners have advantages in image resolution and image quality at higher heart rates. (Images courtesy of (a, d) University of Erlangen, Germany and (b, c) Klinikum Grosshadern, Munich, Germany)

for continuous and consistent volume coverage of the beating heart in all phases of the cardiac cycle, including diastole. Thus, single-slice systems hardly allow for continuous ECG-gated volume coverage with overlapping slice acquisition of the entire heart and reasonable longitudinal resolution within reasonable scan times. Retrospective ECG gating can improve the robustness of ECG synchroniza-

tion against arrhythmia during the scan. However, single-slice acquisition produces a direct relation of the positions of the reconstructed slices and heart rate, including arrhythmia. For a given spiral table-feed, images with high overlap can be reconstructed for high heart rates; for low heart rates, gaps may be produced such that no fix reconstruction increment can be used.
Retrospectively ECG-gated multi-slice spiral scanning has the potential to provide an isotropic, 3D image data set of the complete cardiac volume without gaps and which can be acquired with a single breath-hold spiral acquisition. For ECG-gated reconstruction of scan data with continuous table-feed, dedicated multi-slice spiral reconstruction algorithms are needed that are optimized with respect to temporal resolution, spiral artifact reduction, and volume coverage. Such algorithms require certain pre-conditions of the scan data acquisition. Therefore, we will explain the technical principles of cardiac image reconstruction algorithms in the first section of this chapter before introducing the related acquisition techniques.

4.4.1 Multi-slice Cardiac Spiral Reconstruction

Multi-slice cardiac spiral reconstruction techniques allow for the reconstruction of overlapping images with fixed and heart-rate-independent image increments at arbitrary z-positions and during any given phase of the cardiac cycle. These algorithms consist of two major steps and combine the previously described half-scan reconstruction for optimized temporal resolution with a multi-slice spiral weighting algorithm that compensates for table movement and provides well-defined slice sensitivity profiles. A representative example of these algorithms that also served as the basis for today’s algorithms is the so-called multi-slice cardiac volume reconstruction (MSCV) algorithm; (Ohnesorge 2000a), which is explained in detail below.

During the first step of the MSCV algorithm, so-called multi-slice spiral weighting, a “single-slice” partial-scan data segment is generated for each image using a partial rotation of the multi-slice spiral scan that covers the pertinent z-position. For each projection angle \( \alpha \) within the multi-slice data segment, a linear interpolation is performed between the data of those two detector slices that are in closest proximity to the desired image plane \( z_{ima} \). In contrast to standard multi-slice interpolation techniques (Klingenbeck 1999, Taguchi 1998, Hu 1999, Schaller 2000), each projection is treated independently. As a representative example, the spiral interpolation scheme for a 4-slice CT system, including the calculation of the spiral interpolation weights for some representative projection angles, is illustrated in Figure 4.7. The presented approach is also used for today’s CT scanners with more than four slices and can easily be extended to 16- and 64-slice scanner geometry. With increasing scan time \( t \) and increasing projection angle \( \alpha \), the detector slices travel along the z-axis relative to the patient table. The z-position is normalized to the collimated slice width of one detector slice \( (SW_{coll}) \). Each multi-slice fan-beam projection \( P_{fM}(\alpha_{q}, \beta_{m}, n) \) consists of \( N \) sub-projections corresponding to the \( N \) detector slices that are measured at the same focus (source) position. \( (\alpha_{q}, \beta_{m}) \): projection angle of fan-beam projection \( q, \beta_{m} \): the angle of a ray \( m \) within the fan relative to the central ray, \( n \): detector slice \( n = 0,1,2,3 \) for the example of \( N = 4 \) detector slices). For reconstructing an image at a given z-position \( z_{ima} \), the single-slice projections \( P_{f}(\alpha_{q}, \beta_{m}) \) are calculated by linear interpolation of those sub-projections within the multi-slice projection \( P_{fM}(\alpha_{q}, \beta_{m}, n) \) that are closest to the image z-position \( z_{ima} \) for a given projection angle \( \alpha_{q} \) (Eq. 4.3a). The interpolation weights are determined according to the distances \( d(\alpha_{q}, n) \) of the sub-projections that are considered for reconstruction with z-positions \( z_{i}(\alpha_{q}, n) \) to the image position \( z_{ima} \) (Eqs. 4.3b,c).

\[
P_{f}(\alpha_{q}, \beta_{m}) = \sum_{n=0}^{N-1} w(\alpha_{q}, n) \cdot P_{f,sl}(\alpha_{q}, \beta_{m}, n)
\]

\[
w(\alpha_{q}, n) = \begin{cases} 1 - |d(\alpha_{q}, n)|, & |d(\alpha_{q}, n)| \leq SW_{coll} \\ 0, & |d(\alpha_{q}, n)| > SW_{coll} \end{cases}
\]

\[
d(\alpha_{q}, n) = z_{ima} - z_{i}(\alpha_{q}, n)
\]
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Of course, nearest-neighbor interpolation may not be performed if the distances of the image z-position and the nearest measured sub-projection are too large. A reasonable limitation is to allow that a maximum of 50% of the projections within a single-slice partial-scan data segment may be generated by using the projection of the nearest detector slice only. This situation is shown for image position \( z_{ima,2} \) in Figure 4.7.

Special attention has to be paid to the slice sensitivity profiles (SSPs) as an important parameter of image quality in spiral CT. SSPs produced by the spiral weighting scheme of the MSCV algorithm have been evaluated according to measurements with a gold-plate phantom and based on theoretical calculations in Flohr (1999). In this study, the SSPs and resulting slice widths were assessed for different collimated slice widths and different image positions. In a first-order approximation, MSCV provides a constant, pitch-independent relation (Eq. 4.5) of the collimated slice width \( SW_{coll} \) of one detector slice to the FWHM of the SSP representing the slice width (SW) of the reconstructed image.

\[
SW \approx 1.3 \, SW_{coll} \tag{4.5}
\]

Dependent on the weighting scheme that is present for a particular slice, the FWHM may vary between 1.1 \( SW_{coll} \) and 1.5 \( SW_{coll} \) (Flohr 2001). Slices that are generated in the center of the multi-slice partial scan (Fig. 4.7, \( z_{ima,1} \)) show symmetric profiles. However, at boundary positions (Fig. 4.7, \( z_{ima,2} \)), where projections may be generated with nearest-neighbor interpolation, slices show moderate shifts of the center position on the order of 10%.
MSCV also allows for retrospective generation of thicker slices for a certain collimated slice width $SW_{\text{coll}}$ than given by Eq. 4.5, and it can be performed in a separate reconstruction using the same scan data. During the multi-slice spiral weighting step, up to three single-slice partial-scan data segments are generated for each image $z$-position $z_{\text{ima}}$ at closely adjacent $z$-positions $z_{\text{ima}} - \delta z$, $z_{\text{ima}}$, and $z_{\text{ima}} + \delta z$ ($\delta z$ is a small distance in the z-direction) (Fig. 4.8). The weighted sum of these data segments before performing the partial-scan reconstruction step results in the reconstruction of a thicker slice at the desired position $z_{\text{ima}}$. Table 4.1 contains the appropriate parameters for reconstruction of the slice widths $SW = 1.5SW_{\text{coll}}$, $SW = 2SW_{\text{coll}}$, and $SW = 3SW_{\text{coll}}$. This technique is suited for reconstruction of images with reduced image noise for improved low-contrast resolution at the expense of reduced $z$-resolution.

Further advanced weighting schemes for multi-slice cardiac spiral reconstruction using 16- and 64-slice CT scanners have recently been developed (Flohr 2003). In those generalized weighting schemes, spiral interpolation functions are used that involve more than only the two detector slices in closest proximity to the desired $z$-position $z_{\text{ima}}$. The slice width can be controlled by the shape of the interpolation function (Fig. 4.9). For reconstruction of wide slices relative to the collimation $SW_{\text{coll}}$, trapezoidal weighting functions are used that provide well-defined SSPs almost rectangular in shape, which help to reduce partial-volume artifacts. The parameters $a=0.4, 0.5, \text{ and } 0.6$ determine the width of the weighting function (Flohr 2003). With edge-enhancing weighting functions, the usual slice-broadening by linear trapezoidal weighting functions can be avoided and slice widths can be generated that are equal to the collimated slice width $SW_{\text{coll}}$. Figure 4.10a, b shows the SSPs that are generated for a 16-slice CT scanner for different heart rates and using different weighting functions. Different slice widths can be provided, and it can be demonstrated that the SSPs are largely independent from the heart rate. The smallest reconstructed slice width, measured by means of FWHM, using an edge-enhancing weighting function shows slice broadening of less than 10% compared to the collimated slice width. Based on the generalized weighting scheme, equivalent reconstructed slice widths can also be generated based on different collimation settings (Fig. 4.10c). However, SSPs that are based on thinner collimation are better defined and closer to the ideal rectangular shape, thus minimizing spiral artifacts.

Multi-slice cardiac spiral weighting generates partial-scan data segments for all required image $z$-positions $z_{\text{ima}}$ in arbitrary image increments. Usually, the selected increments are smaller that the reconstructed slice width in order to enhance $z$-axis resolution. The following, second step of the MSCV algorithm performs the previously described single-slice half-scan reconstruction (see Sect. 4.2) of the partial-scan fan-beam data at each image position $z_{\text{ima}}$. According to the sequential half-scan reconstruction, the temporal resolution in the center of the scan field of view equals half the rotation time. The time sensitivity profiles are spatially variable depending on the start and end positions of the tube for acquisition of the considered multi-slice partial-scan data set.

### 4.4.2 ECG-Gated Multi-slice Spiral Acquisition

For retrospectively ECG-gated reconstruction, each image is reconstructed using a multi-slice partial-scan data segment with an arbitrary temporal rela-

<table>
<thead>
<tr>
<th>Slice width</th>
<th>$SW = 1.3SW_{\text{coll}}$</th>
<th>$SW = 1.5SW_{\text{coll}}$</th>
<th>$SW = 2SW_{\text{coll}}$</th>
<th>$SW = 3SW_{\text{coll}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$z$-positions for segments</td>
<td>$z_{\text{ima}}$</td>
<td>$z_{\text{ima}} + 0.25 SW_{\text{coll}}$</td>
<td>$z_{\text{ima}} + 0.5 SW_{\text{coll}}$</td>
<td>$z_{\text{ima}} + SW_{\text{coll}}$</td>
</tr>
<tr>
<td></td>
<td>$z_{\text{ima}} - 0.25 SW_{\text{coll}}$</td>
<td>$z_{\text{ima}} - 0.5 SW_{\text{coll}}$</td>
<td>$z_{\text{ima}}$</td>
<td>$z_{\text{ima}} - SW_{\text{coll}}$</td>
</tr>
</tbody>
</table>
Fig. 4.8. Reconstruction of thicker slice-width with the MSCV algorithm by combining closely adjacent single-slice partial-scan fan-beam data sets. The ratio between the collimation and slice thickness determines the distance and number of adjacent data sets that have been used.

Fig. 4.9. Interpolation functions for a generalized cardiac spiral weighting approach. Trapezoidal weighting functions are used to generate wider slices with well-defined, rectangular-shaped slice sensitivity profiles (SSPs). Edge-enhancing weighting functions are used to generate thin slices with minimal slice-broadening compared to the collimated slice width.
the pitch, which needs to be properly limited to allow for continuous volume coverage. In the shown example, an ECG gating approach was used with a delay that results in image reconstruction during the diastolic phase (shaded stacks). The hatched bars represent image stacks reconstructed from the same spiral data set, but in a different heart phase with a different delay parameter that, in turn, results in image reconstruction during the systolic phase. The entire heart volume can be reconstructed in both heart phases and in any other selected heart phase without gaps. Thus, a multi-phase reconstruction for true functional volume imaging of the moving heart can be generated from various 3D images with incrementally shifted delay parameters.

In each stack, single-slice partial-scan data segments are generated equidistantly spaced in the z-direction depending on the selected image reconstruction increment. Continuous volume coverage can only be achieved when the spiral pitch is appropriately limited by the heart rate. In order to achieve full-volume coverage, the image stacks reconstructed in subsequent heart cycles must cover all z-positions. If the pitch is too high, volume gaps between image stacks that are reconstructed using data from different heart cycles are present (Fig. 4.12). The pitch is limited by the patient’s heart cycle time (RR-interval time), as every z-position of the heart has to be covered by a detector slice at every time during one entire heart cycle. Consequently, the table may not move more than approximately the width of the multi-slice detector within one heartbeat.

According to Eq. 4.6, the table feed is restricted to N-1 single slice widths of a detector with N detector slices within the time interval TRR + TQ, where TRR represents the RR-interval time for the present heart rate, Ttot the full rotation time (between 330 ms and 500 ms), and TQ the partial scan time of a 240–260°
Fig. 4.11. a A scan with continuous table feed and continuous exposure is acquired for retrospectively ECG-gated multi-slice spiral scanning. b Stacks of overlapping images can be reconstructed with $T_{rot}/2$ temporal resolution in every cardiac cycle with the MSCV algorithm. The individual stacks cover certain portions of the cardiac anatomy. Continuous 3D images can be reconstructed in different phases of the cardiac cycle by selection of the data ranges with certain phase relations to the R-waves.
Fig. 4.12. a ECG-gated cardiac spiral examination with a multi-slice CT scanner at a pitch that is too high for the given heart rate. Continuous coverage of the heart volume is no longer possible. Volume gaps are present (hatched segments) between the image stacks that can be reconstructed in each cardiac cycle. b The gaps result in missing cardiac anatomy that could not be covered by the scan.
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partial rotation (e.g., \(T_Q = 360\) ms for \(T_{rot} = 500\) ms and \(T_Q = 238\) ms for \(T_{rot} = 330\) ms).

\[
\text{pitch} \leq [(N-1) \frac{T_{rot}}{(T_{RR}+T_Q)}] / N \quad (4.6)
\]

\(N\) represents the number of illuminated detector slices, each with collimated slice width \(SW_{coll}\). Based on Eq. 4.6, spiral weighting can be performed with two interpolation partners for all projections within the partial-scan data set if the interpolation function covers two detector widths (Fig. 4.9). If the interpolation function covers four detector widths (Fig. 4.9), at least three interpolation partners are available for all projections within the partial-scan data set. For example, with a heart rate of 60 bpm (\(T_{RR} = 1000\) ms), \(N = 16, T_{rot} = 370\) ms and \(T_Q = 267\) ms (e.g., SOMATOM Sensation 16, Siemens), Eq. 4.6 reveals a pitch limit of about 0.27. The maximum pitch depends on the number of illuminated detector slices and rotation time. With \(N = 32, T_{rot} = 330\) ms and \(T_Q = 238\) ms (e.g., SOMATOM Sensation 64, Siemens) the maximum pitch for a heart rate of 60 bpm is reduced to 0.258. However, due to the larger width of the detector with \(N = 32\) detector rows, a significantly increased volume coverage speed is still present compared to the scanner with \(N = 16\) detector rows.

Scan speed can be increased for faster volume coverage by allowing that up to 50% of the interpolated projections are generated with nearest-neighbor interpolation at the edges of the image stacks. With this approach, the heart-rate-dependent spiral pitch is independent of the number of detector rows and is restricted according to Eq. 4.7. For example, with a heart rate of 60 bpm (\(T_{RR} = 1000\) ms), \(N = 16, T_{rot} = 370\) ms, and \(T_Q = 267\) ms, the pitch can be increased from pitch = 0.27, according to Eq. 4.6, to pitch = 0.37, according to Eq. 4.7.

\[
\text{pitch} \leq \frac{T_{rot}}{T_{RR}} \quad (4.7)
\]

Scan protocols with pitch limitations according to Eq. 4.7 have been in use with 4-slice CT systems due to their limited volume coverage speed. With an increasing number of available detector rows (16 and even up to 64), most modern CT scanners use scan protocols with pitch limitations according to Eq. 4.6, due to significant image-quality advantages related to higher-quality SSPs.

The spiral pitch has to be selected prior to the scan and cannot be modified during the scan if the heart rate changes. Substantial drops in heart rate during the scan can produce local volume gaps. Therefore, the maximum heart cycle time (for the minimum heart rate) that is expected during the spiral scan should be used for calculation of the pitch for an individual patient. For modern 16- and 64-slice CT scanners, a spiral pitch of about 0.25 is feasible for most clinical applications, as the entire range of clinically relevant heart rates \(\geq 45\) bpm can be covered and the heart volume can be scanned with the thinnest available slices within a single breath-hold (e.g., 12 cm in 15 s with \(N = 16, T_{rot} = 370\) ms, and \(SW_{coll} = 0.75\) mm).

Retrospectively ECG-gated spiral scanning with 4-slice CT scanners (Ohnesorge 2000a) offered, for the first time, continuous volume imaging for quantification of coronary calcification (Fig. 4.13a) and high-resolution CT angiographic scanning of the coronary arteries (Fig. 4.13b). Moreover, assessment of cardiac function became possible by reusing the same scan data that had been acquired for evaluation of cardiac and coronary morphologies (Fig. 4.13c). The introduction of 16- and 64-slice CT technology and rotation times down to 0.33 s have significantly enhanced spatial and temporal resolution as well as volume-coverage speed (Fig. 4.14), but the basic principles of cardiac scan acquisition and image reconstruction have remained the same since the appearance of the first 4-slice CT scanners. The significant improvement of spatial resolution from 4- to 16-slice and from 16- to 64-slice CT scanners is demonstrated in Figure 4.15. The application spectrum and clinical experience in cardiac and coronary imaging using the presented ECG-gated multi-slice spiral CT acquisition and reconstruction techniques will be presented separately, in detail, in the clinical application section.

4.4.3 Segmented Cardiac Reconstruction Algorithms

Conventional cardiac spiral reconstruction algorithms provide a continuous volume image of the heart with a temporal resolution equal to half the rotation time \(T_{rot}/2\) of the individual slices. How-
Fig. 4.13a–c. Case examples obtained with retrospectively ECG-gated 4-slice spiral CT scans with 0.5-s rotation time. a Coronary calcification of the left main and left descending coronary arteries can be identified with high longitudinal resolution. Image reconstruction with overlapping increments enables more accurate evaluation of the calcified plaque morphology in the z-direction. b The main coronary segments can be evaluated in a 3D reconstruction of a contrast-enhanced coronary CT angiography examination using volume-rendering technique. c Reconstruction of the same data set in end-diastole and end-systole allows for functional evaluation, e.g., of the left atrium (LA) and left ventricle (LV). Volume changes in the LA and LV and in mitral valve movement can be assessed. (Images courtesy of (a) University of Tübingen, Germany, and (b, c) Klinikum Grosshadern, Munich, Germany)

However, the resulting temporal resolution with today’s available minimum rotation times down to 330 ms may not be sufficient for motion-free imaging of the heart at high heart rates (Fig. 4.1) or for imaging the heart in phases with rapid cardiac motion. With unchanged rotation time, the temporal resolution can be improved by using scan data from more than one heart cycle for reconstruction of an image ("segmented reconstruction") (Lackner 1981, Kachelriess 1998, Bruder 1999, Hu 2000, Kachelriess 2000, Flohr 2001). The temporal resolution can be improved up to $T_{rot}/(2S)$ by using scan data of $S$ subsequent heart cycles for image reconstruction, but at the expense of reduced volume-coverage speed or loss of resolution. To maintain good longitudinal resolution and thin-slice images, every z-position of
the heart has to be seen by a detector slice at every
time during the S heart cycles. As a consequence, the
larger the value of S and the lower the patient’s heart
rate, the more the spiral pitch has to be reduced. If
the pitch is too high, there will be z-positions that
are not covered by a detector slice in the desired
phase of the cardiac cycle. To obtain images at these
z-positions, far-reaching interpolations have to be
performed, which may degrade the SSP and reduce
z-resolution (Kachelriess 1998, Kachelriess 2000). If the selected spiral pitch is sufficiently small
for continuous and gap-less volume coverage, the
resulting increase in scan time usually needs to be
compensated by thicker slice collimation and thus
reduced spatial z-resolution. Moreover, a reduction
of the spiral pitch also correlates with a significant
increase of radiation exposure. Improved temporal
resolution alone, at the expense of reduced longitudi
nal resolution and increased radiation exposure,
may degrade the overall diagnostic quality of stud
dies at low heart rates due to blurring of small ana
tomical details.

Heart-rate-adaptive algorithms were developed
that provide adequately improved temporal reso
lution by using segmented reconstruction tech
niques that maintain high z-resolution at the same

**Fig. 4.14a–c.** Case examples obtained with retrospectively ECG-gated 64-slice spiral examinations. a The
coronary artery tree can be visualized with significantly increased spatial resolution and examined with sig
ificantly reduced breath-hold time and increased volume coverage compared to 4-slice CT. Reconstruction
of the same data set in end-diastole and end-systole allows for functional evaluation of the cardiac chambers
and of the cardiac valves. In the example, aortic-valve replacement can be visualized in the open (b, systole)
and closed (c, diastole) positions. The reduced rotation time of 0.33 s allows for improved image quality
also in systolic phase, when there is rapid cardiac motion. (Images courtesy of Jankharia Heart Scan Center,
Bombay, India)
Fig. 4.15. a z-resolution phantom measurements for 4-, 16- and 64-slice CT scanners with ECG-gated spiral acquisition and reconstruction. The resting resolution phantom includes air-filled spheres 0.4–3.0 mm in diameter that can be visualized with multi-planar (MPR) cuts along the scan direction. The thinner collimation of 16-slice CT than of 4-slice CT provides an increase in z-resolution from about 0.9 mm with 4-slice CT to about 0.6 mm with 16-slice CT. Further reduced slice collimation combined with double z-sampling technique (using a z-flying focal spot, z-FFS) enables a further increase of z-resolution for 64-slice CT to about 0.4 mm. Direct comparison of an examination of the same patient with 4-slice (b) and 64-slice (c) CT demonstrates the enhanced spatial resolution of 64-slice CT and its improved visualization of small cardiac structures, such as the flaps of the aortic valve (arrowheads in b and c), and of complex coronary lesions (double arrows in b and c). Also, small-caliber distal coronary artery segments that could not be visualized with 4-slice CT (arrow in b) can be visualized with 64-slice CT (arrow in c).
time. A representative example of these algorithms is the adaptive cardiac volume (ACV) reconstruction technique, which can use data from up to three consecutive heart cycles (Ohnesorge 2001, Flohr 2001). Depending on the patient’s heart rate during the scan, a variable number of heart cycles is used for image reconstruction, ranging from \( S = 1 \) heart cycle at low heart rates (temporal resolution \( T_{\text{rot}}/2 \)) up to \( S = 3 \) heart cycles at high pulse rates [temporal resolution heart-rate-dependent up to \( T_{\text{rot}}/(2S) \)]. With this adaptive approach, narrow SSPs may be obtained at adequate temporal resolution for a wide range of clinically relevant heart rates. A spiral pitch can be used that is sufficient to scan the complete heart with thin slices within a short single breath-hold and with only a moderate increase of radiation exposure.

If scan data are used for the reconstruction of an image that is acquired in \( S > 1 \) subsequent heart cycles, every z-position of the heart has to be covered by a detector slice at every time during the \( S \) heart cycles. Thus, spiral pitch is not only limited by the heart rate but also by the number of heart cycles \( S \) used for reconstruction. If the table moves too quickly, interpolation between data acquired at a larger distance from the image plane can degrade the SSPs (Kachelriess 2000). As an extension of Eq. 4.7, the spiral pitch is limited by the minimum RR-interval time \( T_{\text{RR}} \) during the scan, according to Eq. 4.8, in order to maintain a reasonable quality of the SSPs.

\[
pitch \leq \frac{1}{N} \left( \frac{N - 1}{S} + 1 \right) \frac{T_{\text{rot}}}{T_{\text{RR}}} \tag{4.8}
\]

\( S \) is the number of subsequent heart cycles that are used for image reconstruction. The maximum pitch as a function of heart rate and number \( S \) of consecutive heart cycles used for reconstruction (\( S = 1, 2, 3 \)) is shown in Fig. 4.16 for the example of a 16-slice CT scanner with 0.37-s rotation time. Reasonably fast volume coverage of the heart within one breath-hold and thin slices can be achieved with \( \text{pitch} = 0.2 \). According to Eq. 4.8, \( \text{pitch} = 0.2 \) implies that image reconstruction should be restricted to using data from \( S = 1 \) heart cycle only for moderate heart rates \(< 63 \) bpm. For heart rates \( \geq 3 \) bpm, \( S = 2 \) heart cycles may be used and \( S = 3 \) heart cycles for heart rates \( \geq 95 \) bpm. For other pitch values, different heart-rate thresholds are implied.

Adaptive-segmented reconstruction algorithms automatically adapt the number of consecutive heart cycles used for image reconstruction according to the momentary heart rate of the patient during the scan (Fig. 4.17). These algorithms are extensions to single-segment methods that are limited to the use of one heart cycle only. Both major processing steps of single-segment algorithms, multi-slice spiral weighting and half-scan reconstruction, are also used for adaptive-segmented algorithms. During multi-slice spiral weighting, single-slice partial-scan data segments are generated from partial-scan data segments that are divided into \( S = 1, 2 \) or 3 sub-segments (depending on the heart rate), resulting in a temporal resolution of up to \( T_{\text{rot}}/(2S) \). \( S = 1 \) sub-segment is used for low heart rates and reconstruction is performed with the conventional single-segment algorithm. At higher heart rates, the partial-scan data segment is divided into \( S = 2 \) or 3 sub-segments to improve temporal resolution. Thus, multi-slice spiral data from two or three consecutive heart cycles, respectively, contributes to the single-slice partial-scan data segment. Each sub-segment is generated using data from one heart cycle only. Equivalent to single-segment algorithms, a linear interpolation is performed for each projection angle \( \alpha_n \) within each sub-segment \( j (j = 0...S-1) \) between the data of those two detector slices that are in closest proximity to the desired image plane. The interpolation produces \( S \) single-slice sub-segments located at the same z-position \( z_{\text{ima}} \) (Fig. 4.18) that can be assembled to a complete single-slice partial-scan data segment for image reconstruction. A certain overlap between the sub-segments can be taken into account to allow for smooth transition weighting, which reduces image artifacts due to data inconsistencies. For data consistency during image reconstruction, the sub-segments have to be picked at identical time points within the heart cycle, when the beating heart is exactly in the same relative phase position. The time points are determined by a certain distance relation to the R-waves. Data inconsistency due to arrhythmic heart rates or irregular heart movement in consecutive heart cycles will result in degraded temporal resolution and substantial spatial blurring of moving structures.

Similar to the single-segment algorithms, the second step of adaptive-segmented algorithms per-
forms the previously described single-slice half-scan reconstruction of the partial-scan fan-beam data that are generated for each image position $z_{ima}$. The different data sub-segments from $S$ individual heart cycles are transformed independent of the parallel-beam geometry and are appended as parallel-beam projections.

Adaptive-segmented reconstruction with $S > 1$ sub-segments only allows for improved temporal resolution if the patient’s heart rate and the rotation time of the scanner are appropriately de-synchronized. In a situation of optimal de-synchronization, the projection angles of the start- and end projections of the sub-segments fit together and form a complete partial-scan data segment that contains $180^\circ$ parallel projections after rebinning to parallel geometry (Fig. 4.19a). The partial-scan interval may then be divided into $S$ sub-segments of equal size, and each sub-segment covers a temporal data interval $T_{rot}/4$ for $S = 2$ or $T_{rot}/6$ for $S = 3$ within the same relative heart phase. However, if heart rate and scanner rotation are synchronous, the same heart phase always corresponds to the same projection-angle segment, and a partial-scan interval cannot be divided into smaller sub-segments. As a conse-

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**Fig. 4.16.** Maximum spiral pitch as a function of heart rate and the number $S$ ($S = 1, 2, 3$) of used data segments from consecutive cardiac cycles for the example of a 16-slice CT scanner with 0.37-s rotation time. Exceeding the spiral pitch limitation for a certain heart rate results in volume gaps. Minimum heart rates for use of $S = 2$ and $S = 3$ segments are shown for a pitch of 0.2, which allows for reasonably fast volume coverage with thin-slice collimation. Other pitch values imply different heart-rate thresholds. A pitch of 0.2 enables gap-less volume reconstruction for 16- and 64-slice CT scanners for all heart rates $> 40$ bpm

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**Fig. 4.17.** Retrospectively ECG-gated multi-slice spiral reconstruction using adaptive-segmented reconstruction with up to $S = 2$ segments. Reconstruction with $S = 1$ segment ($T_{rot}/2$ temporal resolution) is performed for low heart rates (e.g., $\leq 63$ bpm). $S = 2$ segments (up to $T_{rot}/4$ temporal resolution) are used for higher heart rates (e.g., $> 63$ bpm)
Fig. 4.18. Generation of a single-slice partial-scan fan-beam data set from multi-slice fan-beam projections acquired in a spiral scan with adaptive-segmented reconstruction. • Projections used for interpolation, ◊ interpolated projections. Single-slice data segments are generated in consecutive cardiac cycles by linear interpolation at specific image z-positions. The individual data segments are combined into a single-slice partial-scan data set for image reconstruction at the considered z-position.

sequence, no better temporal resolution than $T_{\text{rot}}/2$ is possible. In a situation with partially de-synchronized heart rate and rotation, the partial-scan interval may be divided into $S$ sub-segments of different size, each covering temporal data intervals between $T_{\text{rot}}/2$ and $T_{\text{rot}}/2S$. The sub-segment with the longest temporal data interval then determines the temporal resolution of the image (Fig. 4.19b). Segmented reconstruction techniques assume that all anatomy is exactly located in the same position in equivalent phases during the cardiac cycle. Slight shifts of the cardiac anatomy from one heart cycle to the next produce inconsistent scan data and may result in image artifacts (Fig. 4.20). Generally, adaptive-segmented reconstruction algorithms generate images with temporal resolution in the interval $[T_{\text{rot}}/2, T_{\text{rot}}/2S]$, depending on the relation of rotation time and patient heart rate. During the same scan, temporal resolution can vary with changing heart rate as the rotation time is fixed. The relationship between temporal resolution and rotation time and heart rate is demonstrated in Figure 4.21 for different rotation times between 0.5 s and 0.33 s and up to $S = 2$ segments. Figure 4.22 shows the temporal resolution as a function of heart rate for a 64-slice CT scanner with 0.33-s rotation time and up to $S = 4$ segments. The temporal resolution is calculated as the FWHM of the time sensitivity profiles in the center of the scan field of view. Although excellent temporal resolution can be achieved for certain heart rates, for others the heart cycle and scanner rotation are perfectly synchronized and the temporal resolution
For segmented reconstruction with \( S = 2 \) segments and optimal de-synchronization of heart rate and system rotation, the sub-segments form a complete partial-scan data segment without overlap and with temporal resolution \( T_{\text{rot}}/4 \). For only partially de-synchronized heart rate and system rotation, the sub-segment with the longest temporal data interval determines the temporal resolution of the image.

For segmented reconstruction with \( S = 3 \) segments. The segments have different temporal widths for the given rotation time and heart rate. Sub-segment 1, with the longest temporal data interval, determines the temporal resolution of the image. However, the coronary anatomy may be located at slightly shifted positions during consecutive heart cycles, which may result in data inconsistencies and blurring artifacts. In the example, coronary angiography shows the slightly shifted location of the center line of the right coronary artery in three consecutive heart cycles.
Fig. 4.21. Demonstration of temporal resolution as a function of heart rate and rotation time for adaptive-segmented reconstruction using up to $S = 2$ segments. The heart rate threshold for switching from $S = 1$ to $S = 2$ segments is set to 63 bpm. Temporal resolution is strongly heart rate dependent within the interval $[T_{rot}/2, T_{rot}/2S]$. The very different locations of maxima and minima for different rotation speeds suggest that heart-rate-dependent selection of rotation speed might be feasible only in patients with stable and predictable heart rate. Reconstruction with $S = 1$ segment can also be used for higher heart rates. The temporal resolution then remains constant at $T_{rot}/2$, independent of heart rate.

Fig. 4.22. Temporal resolution as a function of heart rate for adaptive-segmented reconstruction with up to $S = 4$ segments using a 0.33-s rotation time. With $S = 4$, temporal resolution can be increased to $T_{rot}/8$ only for certain heart rates. With $S = 4$ segments, temporal resolution changes significantly with only small changes in heart rate. The locations of maxima and minima are strongly dependent on rotation time. Temporal resolution may be difficult to predict in clinical conditions – given the very strong variation of temporal resolution with heart rate and rotation speed – when using $S = 3$, or $S = 4$ segments for reconstruction of $T_{rot}/2$ may not be improved by using data from multiple heart cycles. Image quality improvements with adaptive-segmented reconstruction techniques can be demonstrated in patients with higher heart rates. A direct comparison of single-segment reconstruction and reconstruction with $S = 2$ segments can reveal significantly improved delineation of coronary calcifications in a patient with a heart rate of 76 bpm (Fig. 4.23).

Image quality and resolution along the scan direction provided by segmented reconstruction algorithms have been assessed with a longitudinal resolution phantom that included air-filled spheres with diameters ranging from 0.4 to 3 mm (Flohr 2001, Flohr 2003). It could be demonstrated that adaptive-segmented reconstruction can provide stable longitudinal resolution without a considerable influence of heart rate (Fig. 4.24).

The possible image quality improvements of segmented reconstruction algorithms compared to single-segment reconstruction can be demonstrated in examinations of patients with high heart rates.
Fig. 4.23. Case example of a 16-slice CT cardiac examination that demonstrates the improvement of image quality with adaptive segmented vs. single-segment reconstruction at higher heart rate. A coronary calcium scan was performed in a patient with 76 bpm during the scan with: 16 × 1.5-mm collimation, 0.37-s rotation, reconstruction in diastole at 55% of the RR-interval. Coronary calcifications in the left anterior descending and right coronary arteries ([arrows]) are depicted with considerably less motion using adaptive-segmented reconstruction (right, S = 2 segments, 130-ms temporal resolution) compared to single-segment reconstruction (left, S = 1 segment, 185-ms temporal resolution). (Case courtesy of Tübingen University, Germany)

(Fig. 4.25) or for reconstruction of images in phases of the cardiac cycle with rapid cardiac motion (Fig. 4.26). However, cardiac anatomy can be located at slightly shifted positions even in equivalent phases during the cardiac cycle. Due to these inconsistencies, segmented reconstruction techniques may produce degraded image quality compared to single-segment reconstruction, despite higher temporal resolution (Fig. 4.27). To achieve most robust cardiac image quality with the lowest possible radiation exposure, single-segment reconstruction techniques combined with fast rotation speeds are usually preferred in clinical practice. The use of segmented reconstruction techniques is limited to special clinical situations, such as patients with either very high heart rates or stable and predictable heart rates, and for reconstruction of the larger cardiac anatomy during phases with rapid cardiac motion in order to analyze cardiac function.

4.4.4 Cardiac Cone-Beam Reconstruction Algorithms

The cardiac image reconstruction algorithms introduced thus far were designed for multi-slice CT systems with only a limited number of detector rows, and did not take into account the conical shape of the X-ray beam in the axial direction. As demonstrated in Chapter 3, cone-beam reconstruction algorithms are mandatory for general-purpose CT scanning with eight and more detector slices to avoid severe image artifacts. The severity of cone-beam-induced artifacts depends on the number of simultaneously acquired slices, on the width of each slice, and on the distance of a considered object from the center of the scan field of view. Cone-beam artifacts are most pronounced at high contrast boundaries; thus, typical sources of cone-beam artifacts are the ribs, pelvic bones, and bronchi. Since the heart is
Fig. 4.25a–c. Case examples of a 64-slice CT scanner with 0.6-mm collimation and 0.33-s rotation time. The figure shows the improved image quality of cardiac and coronary CT angiography examinations with adaptive-segmented vs. single-segment reconstruction at higher heart rate. The patient’s heart rate was between 106 and 110 bpm during the scan. Images were reconstructed at 35% of the RR-interval. Motion artifacts are present in the images based on single-segment reconstruction with 165-ms temporal resolution (a, arrows). Two segments can increase temporal resolution from 165 ms to about 90 ms using adaptive segmented reconstruction, and motion artifacts can be largely eliminated (b). Three segments allow a further improvement in temporal resolution to 70 ms, but image quality is compromised due to the combination of data from three consecutive heart beats (c). (Cases courtesy of Klinikum Grosshadern, University of Munich, Germany)

Fig. 4.24a, b. Investigation of the influence of heart rate on image quality and resolution in the scan direction with 16-slice (a) and 64-slice (b) CT using a resting longitudinal resolution phantom (see Figs. 4.3 and 4.15) and simulated ECG signals with different heart rates that control the adaptive-segmented reconstruction. The 16-slice CT scans were acquired with 0.75-mm collimation, 0.37-s rotation, pitch 0.2, 0.75-mm slice width, and 0.5-mm increments. A longitudinal resolution of 0.6 mm can be achieved for all anticipated heart rates. The 64-slice CT scans were acquired with 0.6-mm collimation (using double z-sampling technique), 0.33-s rotation, pitch 0.2, 0.6-mm slice width, and 0.3-mm increments. A longitudinal resolution of 0.4-mm can be achieved for all anticipated heart rates. In 16-slice and 64-slice CT phantom tests, image quality is largely independent of both heart rate and the number of segments used for reconstruction.

usually centered and does not contain large high-contrast structures, cone-beam artifacts are largely negligible for CT scanners up to 16 slices (Flohr 2003). However, cardiac cone-beam reconstruction algorithms gain importance for CT scanners with a higher number (32 or 64) of detector slices as well as for ECG-gated examinations of the chest and diagnosis of anatomy that is located at a larger distance from the center of the scan field of view. The cardiac cone-beam reconstruction algorithms in use today (Bruder 2001, Bruder 2002) represent extensions of those in use for general-purpose multi-slice spiral reconstruction and have been successfully tested for the latest multi-slice CT scanners with 32 and up to 64 detector slices.

Similar to cone-beam reconstruction techniques in multi-slice body CT imaging, cardiac cone-beam reconstruction techniques generate double-oblique image stacks (so-called booklets) that are individually adapted and optimally fitted to the spiral path. In a second step, these booklets are reformatted to a set of overlapping trans-axial images (Fig. 4.28). For single-segment reconstruction, one booklet is used for reconstruction and reformation in each heart cycle. Data segments that cover 180° of parallel-beam projections are employed for the generation of these booklets, leading to a temporal resolution of the images within the booklet of half of the rotation time of the scanner. In the case of segmented
Fig. 4.26a–d. Case example of a cardiac CT angiography examination for a patient with a heart rate of 100–107 bpm using a 64-slice CT scanner with $64 \times 0.6$-mm slices and 0.33-s rotation time. Images are reconstructed during the systolic phase with a relative delay of 25% of the RR-interval, and MPRs are generated in the short heart axis. With single-segment reconstruction and a temporal resolution of 165-ms, motion artifacts are present (a, arrows). Adaptive-segmented reconstruction with 2 (b), 3 (c), and 4 (d) segments improves temporal resolution up to 60 ms, and cardiac anatomy can be visualized during phases of rapid cardiac motion. A substantial image quality improvement takes place when using two segments instead of one. Further image quality improvements are only minor when using more than two segments. (Cases courtesy of Klinikum Grosshadern, University of Munich, Germany)
Fig. 4.27a–e. Case example of a cardiac CT angiography using a 64-slice CT scanner with 64 × 0.6-mm slices and 0.33-s rotation time. a The patient had a varying heart rate between 69 and 95 bpm during the scan. The arrows in a point to the scan start and to the scan end, and illustrate the scan time interval of about 11 s. b, d Images were reconstructed with a relative delay of 45% of the RR-interval using single-segment reconstruction with 165-ms temporal resolution. c, e Three-segment reconstruction with increased temporal resolution between 55 and 165 ms depending on heart rate. Despite the higher effective temporal resolution of 3-segment reconstruction, motion artifacts are still present (arrows). In some segments, even better image quality is achieved with single-segment reconstruction (double arrows, triple arrows) due to beat-to-beat data inconsistencies with 3-segment reconstruction in the presence of heart-rate changes. (Cases courtesy of Klinikum Grosshadern, University of Munich, Germany)
cardiac reconstruction data, segments from consecutive heart cycles that cover less than 180° of parallel-beam projections are used for image formation. Segmented cardiac cone-beam reconstruction algorithms permit reconstruction of image booklets from data segments covering less than 180°; however, these do not represent complete CT images due to truncation of the projection segments. The incomplete CT images from individual heart cycles are separately reformatted and subsequently combined to complete axial CT images containing contributions from multiple consecutive heart cycles that add up to a total parallel projection interval of 180°.

The elimination of cone-beam artifacts in cardiac CT images has been demonstrated in phantom studies using data from multi-slice CT scanners with up to 64 individual detector rows (Bruder 2002) (Fig. 4.29). Cardiac cone-beam reconstruction algorithms have also demonstrated considerable image quality improvements in ECG-gated cardiac studies with latest 64-slice CT scanners, in particular for the diagnosis of peripheral thoracic anatomy located further from the center of the scan field of view (Fig. 4.30).

**4.4.5 ECG-Gated Spiral Scanning with Increased Volume Speed**

Both ECG-triggered sequential scanning and ECG-gated spiral scanning require substantially slower scan speeds than conventional multi-slice spiral scanning without ECG-gating. Normal ECG-triggered or ECG-gated scan techniques using the present 4- and 16-slice CT scanners cannot provide thin-slice coverage of the entire chest anatomy with a scan range of about 350 mm within a short single breath-hold. However, thoracic CT studies are frequently degraded by motion artifacts caused by transmitted cardiac pulsation (Loubeyre 1997, Schöpf 2000) and elimination of motion artifacts via ECG synchronization may substantially improve diagnostic quality. Although the latest 32- and 64-slice CT scanners are capable of covering the entire chest with ECG-gated scan protocols in about 15–20 s, a further reduction of breath-hold time remains feasible.

Scan speed can be increased with a modified ECG-gated spiral acquisition technique that enables image reconstruction with high temporal resolution by the use of higher pitch values (Flohr 2002).
limitation of the spiral pitch for ECG-gated multi-slice spiral scanning results is a consequence of the phase-consistent coverage of the heart volume, i.e., data may only be used during identical relative time points within the cardiac cycle. The spiral pitch can be increased by allowing images that are part of the same continuous volume to be reconstructed with half-scan reconstruction algorithms but in different phases of the cardiac cycle. In this approach, data have to be excluded from image reconstruction acquired during phases of high cardiac motion (i.e., systole). The data windows that are excluded from reconstruction during the different cardiac cycles may be positioned relative to the onset of the R-waves with a certain temporal relation (Fig. 4.31). A stack of images with a given increment is reconstructed during each cardiac cycle and combined to form a volume image data set. For each image, an interpolated partial-scan data segment is generated with the previously described single-segment cardiac spiral reconstruction approach. Thus, the temporal resolution of each individual image is equal to half of the rotation time. The data segments selected for reconstruction of individual slices are shifted equidistantly in time within the heart cycle. Pulsation artifacts are usually most severe during the systolic phase of cardiac contraction. Therefore, to eliminate image degradation due to systolic pulsation, scan

Fig. 4.29a–d. Evaluation of cardiac cone-beam reconstruction algorithms using an anthropomorphic heart phantom. CT data were generated for a multi-slice CT scanner with $64 \times 0.625$-mm collimation (40-mm coverage per rotation, no $z$-flying focal spot) and reconstructed using cardiac reconstruction algorithms without (a, c) and with (b, d) cone-beam correction. Due to the larger volume coverage per rotation, images without cardiac cone-beam reconstruction demonstrate significant cone-beam artifacts in the heart and in the periphery. These can be eliminated using cardiac cone-beam reconstruction.
Fig. 4.30a–f. Demonstration of image-quality improvements using cardiac cone-beam reconstruction techniques with a 64-slice CT scanner. CT data were acquired in a chest phantom for a 64-slice CT scanner with 32 × 0.6-mm collimation (19.2-mm coverage per rotation, with z-flying focal spot) and reconstructed using cardiac reconstruction algorithms without (a, c, e) and with (b, d, f) cone-beam correction. Due to the limited volume coverage per rotation, images can be reconstructed without cone-beam reconstruction to display the cardiac anatomy, and cone-beam artifacts are negligible (arrows in a and b). However, such artifacts are present in the periphery of the scan field of view (arrows in c and e) and cardiac cone-beam reconstruction is required to reduce artifacts, in particular for the peripheral thoracic anatomy (arrows in d and f).
data acquired during this phase of cardiac contraction is omitted within a temporal window of fixed width $\Delta T_S$. This window is defined within the cardiac cycle with a fixed temporal relation to the onset of the R-wave. Continuous volume coverage requires that the image stacks that are reconstructed in consecutive heart cycles overlap in the z-direction. Thus, the spiral pitch is restricted to a value given by the fixed temporal width $\Delta T_S$ of the data window that is omitted during image reconstruction (Eq. 4.9). Here, the pitch limitation is independent of heart rate.

$$\text{pitch} \leq \frac{N-1}{N} \cdot \frac{T_{\text{rot}}}{\Delta T_S + T_{\text{rot}}/2}$$  \hspace{1cm} (4.9)$$

$N$ represents the number of slices and $T_{\text{rot}}$ is the scanner rotation time. It has been shown that values between $\Delta T_S$ 400 ms and $\Delta T_S$ 500 ms are a reasonable approximation for the duration of the phase of strongest cardiac pulsation. The resulting pitch values between 0.5 and 0.6 allow the entire thorax to be covered with thin slices within a single breath-hold. As an example, a modern 16-slice CT scanner can cover a 350-mm scan range with 16 × 0.75-mm collimation, 0.37-s rotation, and a pitch of 0.5 in a breath-hold time of 22 s. Examples of examinations of the thoracic aorta are shown in Figs. 4.32 and 4.33. Direct comparison of conventional non-gated spiral reconstruction and ECG-gated reconstruction of the same scan data demonstrates a substantial improvement with the latter technique, since motion artifacts can be largely eliminated.

With the advent of 32- and 64-slice CT scanners, thin-slice ECG-gated coverage of the entire chest has also become feasible with regular ECG-gated cardiac scan protocols. Based on a usual pitch value of 0.2, which can provide gap-less volume coverage for a heart rate down to about 40 bpm, a 350-mm scan range can be covered within a breath-hold time of 10–20 s. For patients with heart rate of 60 bpm and above during the scan, the pitch can be increased to values of about 0.3, thus providing a 50% increase in volume coverage speed. Cardiac cone-beam reconstruction algorithms become increasingly important for ECG-gated scanning of the cardiothoracic anatomy with 32- and 64-slice CT scanners, since large portions of the thoracic anatomy are located...
Fig. 4.32a, b. Fast ECG-gated spiral acquisition with a 4-slice CT scanner for evaluation of the thoracic aorta: comparison with multi-slice spiral reconstruction using a standard algorithm. a With standard reconstruction, considerable pulsation artifacts that produce a double contour are visible. b The intimal flap of a type-A dissection can be evaluated free of pulsation artifacts on a corresponding transverse section using fast ECG gating. Scan and reconstruction parameters: 4 × 1-mm collimation, 0.5-s rotation, 120 kV, 300 mA/200 mAs, pitch 0.6, 1.25-mm slice-width, 0.8-mm image increments, window of omitted data has width $\Delta T_s = 400$ ms and starts 0 ms after onset of the R-wave. (Images courtesy of the University of Tübingen, Germany)

Fig. 4.33a, b. Case study of a CT angiography examination of the thoracic and abdominal aorta using a 16-slice CT scanner with 16 × 0.75-mm collimation and 0.37-s rotation time and fast ECG gating. In the conventional non-gated technique, the displayed scan range of 500 mm can be scanned in 10 s with a pitch of 1.5 but pulsation artifacts may be present. With a fast ECG-gated spiral protocol and pitch 0.5, the scan time is increased to 30 s but pulsation artifacts are largely eliminated. In the presented case, fast ECG-gated acquisition demonstrates an aortic dissection (type B) free of motion over its entire course in sagittal MPR (a) and volume-rendering technique (b) (Case by courtesy of University Hospital Graz, Austria)
at greater distance to the center of the scan field of view and thus are subject to cone-beam artifacts.

4.5 Synchronization with the ECG and Cardiac Motion

4.5.1 ECG-Based Phase Selection

With both prospective ECG triggering and retrospective ECG gating, the starting points of data acquisition or the start points of data selection for reconstruction have to be defined within each cardiac cycle during the acquisition. These start points are determined relative to the R-waves of the ECG signal by a phase parameter. The following phase-selection strategies can be used (Fig. 4.34).

- Relative delay: A temporal delay $T_{\text{del}}$ relative to the onset of the previous R-wave is used for determining the start point of the ECG-triggered acquisition or the start point of the reconstruction data interval. The delay time $T_{\text{del}}$ is determined individually for each heart cycle as a given percentage $\delta_{RR}$ of the RR-interval time $T_{RR}$. For ECG triggering, the RR-interval times have to be prospectively estimated based on the prior RR-interval times.

- Absolute delay: Fixed delay times $T_{\text{del}}$ after onset of the R-wave define the start point of the ECG-triggered acquisition or the start point of the reconstruction data interval.

- Absolute reverse: Fixed times $T_{\text{rev}}$ prior to the onset of the next R-wave define the start point of the ECG-triggered acquisition or the start point of the reconstruction data interval. For ECG triggering, the position of the next R-wave has to be prospectively estimated based on the prior RR-interval times.

Fig. 4.34a–c. Phase definition for ECG triggering and ECG gating by selection of the start point of the temporal data interval within every heart cycle. The following different phase-definition strategies are used. a Relative delay: delay time after the previous R-wave, determined as a fraction $\delta_{RR}$ of the RR-interval. b Absolute reverse: constant interval $T_{\text{rev}}$ prior to the next R-wave. c Absolute delay: constant delay time $T_{\text{del}}$ after the previous R-wave
Different approaches are in use in clinical practice today depending on the clinical application. For motion-free imaging of small anatomical structures (i.e., coronary arteries) in diastolic phase, with least cardiac motion, the relative delay and absolute reverse approaches are most frequently used. Cardiac image quality greatly depends on the cardiac phase selected for image reconstruction (Fig. 4.35), as cardiac motion varies widely during different phases within the cardiac cycle (Fig. 4.1). Cardiac image reconstruction is usually performed during phases of least cardiac motion – usually between mid-diastole and end-diastole of the cardiac cycle. Within the ECG trace, end-diastole is usually represented by the onset of the P-wave. For the latest 64-slice CT scanners, which provide higher temporal resolution, cardiac image reconstruction can also be feasible during end-systole. The latter represents a shorter phase of low cardiac motion and is marked within the ECG trace by the onset of the T-wave. Despite intensive research to standardize phase selection for cardiac image reconstruction, heart-rate-dependent and patient-individualized optimization usually remain necessary to obtain the best possible results (Hong 2001a, Köpp 2002).

For functional imaging with retrospective ECG gating, images need to be reconstructed in phases of maximum and minimum filling of the ventricles (end-diastole and end-systole). End-diastolic reconstruction is feasible with the absolute reverse approach, while the absolute delay approach allows for most consistent reconstruction in end-systolic phase.

A regular heart rate without significant and sudden heart rate changes during the scan is usually a pre-requisite for diagnostic image quality for examinations of small cardiac and coronary anatomy. A strongly irregular heart rate may result in substantial data mis-registration and image artifacts, thus leading to results that cannot be used diagnostically (Fig. 4.36). However, retrospective ECG gating allows for viewing and analysis of the ECG signal after the end of the scan, and data are available during all phases of the cardiac cycle. This offers the possibility of retrospectively modifying synchronization of the ECG trace and data reconstruction. Interactive editing of R-peak positions that are detected inappropriately or represent irregular heart beats can have a positive impact on phase consistency and image quality in patients with irregular

Fig. 4.35. Cardiac image reconstruction of data acquired with a 64-slice CT scanner with 0.33-s rotation time in different phases of the cardiac cycle using the relative-delay approach. Images were reconstructed with single-segment reconstruction, thus providing a temporal resolution of 165 ms. Image quality largely depends on the selected phase of the cardiac cycle. The best image quality can be obtained in mid-diastolic phase ($\delta_{RR} = 50\%$) and end-systolic phase ($\delta_{RR} = 30\%$).
heart beats (Fig. 4.37). Also, individual adjustment of the image-time interval positions – independent of the position that is determined by the ECG-gating parameter – may be useful in patients with substantial arrhythmias.

### 4.5.2
The Pros and Cons of ECG Gating and ECG Triggering

Retrospectively ECG-gated spiral scanning with single-slice CT systems featuring sub-second rotation has been tested for coronary artery and cardiac-function imaging in clinical trials but serious limitations have been discovered (Mochizuki 2000). With the advent of multi-slice acquisition, ECG-gated spiral scanning has become feasible, and with significant advantages over prospective ECG triggering that are important for clinical applications.

- **ECG-gated spiral scanning** provides continuous volume coverage and better spatial resolution in the patients’ longitudinal direction, as images can be reconstructed with arbitrary, overlapping slice increments. ECG-triggered sequential scanning is usually restricted to scanning with non-overlapping adjacent slices or slice increments with only small overlap. The scan time to cover the heart volume is thus directly proportional to the slice increment.
- **Retrospective analysis** of the ECG results in less sensitivity to heart-rate changes during the scan. The ECG trace can be retrospectively analyzed and extra-systolic beats can be eliminated for reconstruction. With prospective ECG triggering, estimation of the next RR-interval may be incorrect when heart-rate changes are present (e.g., arrhythmia, Vasalva maneuver) and scans may be placed in inconsistent heart phases. In that case, even slices with high temporal resolu-
tion that are free of motion artifacts cannot be used for continuous 3D image data sets.

- ECG-gated spiral scanning provides faster volume coverage than ECG-triggered sequential scanning because spiral scan data can be acquired continuously and images can be reconstructed in every cardiac cycle. Relatively long travel distances and travel times of the table are present for multi-slice acquisition in between two consecutive scans. This limits the scan-cycle time (minimum time between the start of two consecutive scans), and ECG-triggered scans can often be obtained only in every second heart beat for patients with higher heart rates.

- ECG-gated spiral acquisition allows for imaging in a complete cardiac cycle using the same scan data set, thus providing information on cardiac function. ECG-triggered acquisition targets only one specific phase of the cardiac cycle and requires additional examinations with new breath-hold levels and additional contrast agent to cover more phases of the cardiac cycle.

During ECG-gated spiral imaging of the heart, data are acquired with small spiral pitch (pitch << number of slices, i.e., overlapping acquisition) and continuous X-ray exposure. Thus, ECG-gated spiral acquisition requires a higher patient dose of radiation than ECG-triggered sequential acquisition for comparable signal-to-noise ratio. All spiral data can be used for image reconstruction in different cardiac phases and no data have to be omitted. However, if only one dedicated cardiac phase (i.e., diastolic phase) needs to be targeted by retrospec-
tive data selection, the specific requirements of the clinical application should indicate whether ECG-triggered sequential scanning with less radiation exposure could provide sufficient performance and image quality. Newer developments enable a reduction of radiation exposure during retrospectively ECG-gated scanning. The goal of these techniques is to maintain the important benefits of ECG-gated spiral scanning but to reduce X-ray radiation exposure to levels comparable to those of ECG-triggered sequential acquisition, as will be explained in a later chapter.

### 4.5.3 Alternative Cardiac-Motion Gating Approaches

Phase selection of ECG-synchronized CT scans is usually performed with respect to the temporal position of the R-waves. The R-wave can be easily detected from the patient's ECG due to its high signal amplitudes. Cardiac image reconstruction is frequently carried out during end-systole and end-diastole, which are the phases of the cardiac cycle with the least cardiac motion and which determine minimum and the maximum left ventricular volumes. For R-wave-based ECG-synchronization, end-systole and end-diastole have to be estimated with appropriate phase parameters relative to the R-wave. More advanced ECG-gating techniques that determine the location of the end-systolic phase based on the detection of the T-wave and the end-diastolic phase based on the detection of the P-wave are under investigation. However, given the small signal amplitudes, reliable detection of the T-waves and P-waves requires advanced software-based algorithms, which themselves are currently under development. Nonetheless, the first promising results were obtained in a clinical study that obtained better results with cardiac image reconstruction in end-diastole based on P-wave gating, compared to reconstruction in diastole with a relative delay in relation to the R-wave (SATO 2003) (Fig. 4.38).

It should be kept in mind that the ECG is only an indirect measure of cardiac motion. In some cases, such as in patients with atrial fibrillation or with extra-systoles, the electrical stimulation does not directly correspond to the real motion of the heart. Therefore, motion-detection algorithms have been investigated that have the potential to directly measure cardiac motion from the acquired scan data.

The first cardiac-motion detection algorithm to be clinically investigated (OHNESORGE 1999) used the opposite parallel-beam projections $P_p(\Theta, \pi)$ and $P_p(\Theta-\pi, p)$ in a fixed image plane measured at a projection angle with a $180^\circ$ shift. In a static object, the signal difference between two opposite parallel-beam projections measured in the same plane equals zero. If the object is moving, the magnitude of the signal difference can be used to measure the displacement of a particular anatomical structure and the amount of movement that occurs between measurements of the two opposite projections, i.e., within the time of a half rotation of the system (Fig. 4.39). For example, the sum of the absolute values of the individual signal differences of all complementary rays in the considered opposite projections $\Sigma_p |P_p(\Theta, p)-P_p(\Theta-\pi, p)|$ can be used as an indicator for the displacement and motion of an anatomical structure at the point in time when the projection $P_p(\Theta, p)$ was measured. With this approach, it could be demonstrated that the derived signal difference correlates with both the displacement and the motion of the heart during the cardiac cycle (Fig. 4.39). However, the accuracy and consistency of the signal for reproducible detection of equivalent phases in consecutive cardiac cycle is limited, so that use of the algorithm is restricted to assessing the presence or absence of cardiac motion in patients with irregular heart beats during the scan, and therefore as input information for interactive editing of the recorded ECG.

A more recent approach, the so-called Kymogram algorithm (KACHELRIESS 2002), derives cardiac motion from projection data based on the calculation of the “center of mass” of the scanned fan-beam projections ($m_{COM}$) and the center of mass of the object in the considered image plane. The center of mass $m_{COM}$ of a fan-beam projection $P_f(\alpha, \beta_m)$ is determined by the ray at angle $\beta_m$ within the projection such that the sum of the signals on both sides of that ray is equivalent (Eq. 4.10). The center of mass of the object in the considered image plane can be determined by the point of intersection of the rays that determine the centers of mass in two closely adjacent projections (Fig. 4.40).
Fig. 4.38a, b. Cardiac image quality obtained by a 4-slice CT scanner with 0.5-s rotation time in end-diastole based on R-wave- and P-wave-gated image reconstruction. a During P-wave gating, the end of the image reconstruction window is positioned on the P-wave. b During R-wave gating, the start of the image reconstruction window is positioned at 50% of the RR-interval. P-wave-gated reconstruction yields superior image quality compared to R-wave-gated reconstruction due to more consistent phase selection.

\[
\sum_{m=\text{first}+1}^{m=\text{last}+1} P_f(\alpha_q, \beta_m) = \sum_{m=\text{first}+1}^{m=\text{last}+1} P_f(\alpha_q, \beta_m)
\]

The center of mass changes its position with changing static anatomy from image plane to image plane and with changing positions of an anatomical structure due to motion. After subtracting the function of the center of mass for different slice positions, the variation of the center of mass over time represents a measure of cardiac motion. The resulting function of the variation of the center of mass over time is called the Kymogram (Fig. 4.41). Kachelriess (2002) demonstrated that peaks of the Kymogram often correlate to the position of R-waves in ECGs recorded during the scans of patients with regular heart rates, and that Kymogram-gated cardiac image reconstruction could yield results comparable to those obtained with ECG-gated image reconstruction (Fig. 4.42a). However, further clinical studies have demonstrated that Kymogram-gated image reconstruction results in inferior image quality compared to ECG-gated image reconstruction under clinical conditions, in both patients with regular heart rates and in those with irregular heart rates (Fig. 4.42b), (Fischbach 2004).

It can therefore be concluded that ECG-correlated scanning and image reconstruction remains
Fig. 4.39a, b. The principle of detection of cardiac motion from scan data. a The difference signal of opposite parallel projections is calculated and used as a measure for the motion and displacement of the cardiac anatomy in between measurement of the two projections. A difference signal is determined for every projection as the sum of the absolute values of the individual signal differences of all complementary rays in the considered opposite projections. b The difference signal correlated to the ECG that has been recorded during the scan is shown. Images that are reconstructed during the detected phase of low motion demonstrate fewer motion artifacts.

Fig. 4.40. Center of mass detection a within a projection and b for the object from two projections as the basis of Kymogram-gated reconstruction.
the method of choice to synchronize CT scanning with cardiac motion. If accurate automated P- and T-wave detection algorithms become available, ECG gating related to P- and T-waves will offer a promising alternative to R-wave-related ECG-gating. Other approaches to detect cardiac motion automatically from the projection data of the CT scan have demonstrated the feasibility of the principle, but they have yet to provide sufficient robustness in clinical routine. To date, the use of such algorithms is restricted to a back-up role for the ECG signal, e.g., in case of poor signal detection, R-wave misregistration, or failure of the ECG.

4.6 Radiation Exposure Considerations

Radiation exposure of patients during computed tomography and the resulting potential radiation hazards have recently gained increasing attention, both in the public and in the scientific literature (Brenner 2001, Nickoloff 2001). In particular, the radiation dose for ECG-synchronized cardiac scanning has been a topic of considerable controversy. This section will introduce the basic principles of radiation-dose measurement in CT, exposure estimations for ECG-gated cardiac examinations with 16-slice and 64-slice CT systems, and the most recent approaches to reduce radiation exposure during cardiac CT examinations.

4.6.1 Principles of Radiation Dose Measurement in CT

In CT, the average dose in the scan plane is best described by the weighted computerized tomographic dose index (CTDIw) (Morin 2003, McCollough 2003), which is determined from CTDI100 measurements both in the center and at the periphery of a 16-cm Lucite phantom for the head and a 32-cm Lucite phantom for the body. For the CTDI100 measurements, a 100-mm-long ionization chamber...
Fig. 4.42a, b. Comparison of cardiac image quality of a 16-slice CT scanner with 0.42-s rotation time in a patient with regular sinus rhythm. Reconstruction with a ECG-gating (at 50% of the RR-interval) and b Kymogram-gating (at 50% of the interval between the peaks). The peaks of the Kymogram correlate well with the detected R-Waves. The Kymogram-gated reconstruction demonstrates image quality comparable to the ECG-gated reconstruction except for moderate registration artifacts at the beginning and end of the scan. (Case courtesy of Münster University, Germany)

is used. Figure 4.43 shows the typical equipment for dose measurements. CTDI_w is a good estimate for the average patient dose as long as the patient’s size is similar to that of the respective phantoms. CTDI_w is defined according to (Eq. 4.11) (Morin 2003).

\[
\text{CTDI}_w = \frac{1}{3} \text{CTDI}_{100} \text{(center)} + \frac{2}{3} \text{CTDI}_{100} \text{(periphery)}
\]  

(4.11)

CTDI_w, given in mGy, is always measured in an axial scan mode. It depends on scanner geometry, slice collimation, beam pre-filtration, X-ray tube voltage (in kV), tube current (in mA), and gantry rotation time T_rot. The product of mA and T_rot is the mAs value of the scan. To obtain a parameter characteristic for the scanner used, it is helpful to eliminate mAs dependence and to introduce a normalized (CTDI_w)_n, given in mGy/mAs:

\[
\text{CTDI}_w = \text{mA} \cdot T_{rot} \cdot (\text{CTDI}_w)_n = \text{mAs} \cdot (\text{CTDI}_w)_n
\]  

(4.12)

CTDI_w is a measure of the dose in a single axial scan and depends on X-ray tube voltage and slice collimation. The latter parameters are needed to specify CTDI_w. For multi-slice CT systems, CTDI_w tends to increase with decreasing collimated slice width, as a consequence of the increasing relative contribution of the penumbra zones of the dose profiles. As a representative example, Figure 4.44 shows CTDI_w at 120 kV for the 32-cm body phantom as a function of the total collimated width of the detector for a 4-slice CT system and a 16-slice CT system with similar system geometry. Scan protocols for different CT scanners should always be compared on the basis of CTDI_w and never on the basis of mAs, since different system geometries can result in significant differences in the radiation dose applied at identical mAs.

(CTDI_w)_n can be used to calculate the radiation dose for axial scans, both for standard applications and for ECG-triggered sequential scanning. For most CT scanners, the mAs value for axial scans, which is
Fig. 4.43a, b. Comparison of cardiac image quality of a 16-slice CT scanner with 0.42-s rotation time in a patient with regular sinus rhythm. Reconstruction with a ECG-gating (at 50% of the RR-interval) and b Kymogram-gating (at 50% of the interval between the peaks). The peaks of the Kymogram correlate well with some R-waves during the scan but do not correlate well with the detected R-waves throughout the scan. The Kymogram-gated reconstruction demonstrates severe misregistration artifacts while the ECG-gated reconstruction yields artifact-free results. (Case courtesy of Münster University, Germany)

the product of tube current mA and slice exposure time $T_{exp}$ is indicated on the user interface. Usually, $T_{exp}$ equals the gantry rotation time $T_{rot}$. For cardiac applications, however, partial scans are generally used for ECG-triggered sequential scanning. In this case, the slice exposure time is about two-thirds of the gantry rotation time. The principle of radiation exposure calculation for ECG-triggered scan protocols can be demonstrated with a practical example for ECG-triggered scanning of coronary calcium quantification. The slice exposure time of a 16-slice CT scanner for a partial scan at 0.42-s gantry rotation time is about 0.3 s. The scan protocol recommended by the manufacturer for ECG-triggered sequential scanning for coronary calcium quantification uses a tube current of 100 mA at a voltage of 120 kV. With these parameters, $100 \times 0.3$ mAs = 30 mAs are applied. This value is shown on the user interface. $(CTD_{w})_n = 0.072$ mGy/mAs can be found in the data sheet of this scanner for the given protocol at 120 kV and for $12 \times 1.5$-mm collimation that provides $6 \times 3$-mm slices per scan. A $CTD_{w} = 0.072$ mGy/mAs $\times$ 30 mAs $= 2.16$ mGy can therefore be determined for this protocol (Flohr 2003).

For ECG-gated multi-slice spiral scans of the heart, the situation is more complicated. To represent the dose in a multi-slice spiral scan, it is essential to account for gaps or overlaps between the radiation-dose profiles from consecutive rotations of the X-ray source (Morin 2003). For this purpose, $CTD_{vol}$, the volume $CTD_{w}$, has been introduced

$$CTD_{vol} = \frac{1}{pitch} \cdot CTD_{w} = mA \cdot T_{rot} \cdot \frac{1}{pitch} \cdot (CTD_{w})_n$$

(4.13)

The factor 1/pitch accounts for the increasing dose accumulation with decreasing spiral pitch due to the increasing spiral overlap. In principle, Eq. 4.13 is valid for both single-slice and multi-slice spiral CT. For ECG-gated cardiac scanning, low pitch values (typically pitch $= 0.2\text{--}0.35$, depending on the number of detector rows, the gantry rotation time,
and the number of segments used for image reconstruction) have to be used to ensure gap-less coverage of the heart volume in all phases of the cardiac cycle. The highly overlapping data acquisition has to be taken into account when calculating the dose for an ECG-gated spiral scan. Again, the principle of radiation exposure calculation for ECG-gated spiral scan protocols can be demonstrated with a practical example of an ECG-gated 16-slice spiral CT scan protocol used for coronary calcium quantification. The example protocol uses a 0.42-s rotation time, 16×1.5-mm collimation, pitch 0.28, 120 kV, and 100-mA tube current. The mAs value for this scan is 100 mA × 0.42 s = 42 mAs, which, in terms of the applied radiation dose, cannot directly be compared to mAs values for ECG-triggered sequential scanning, since in the latter the pitch dependency would be neglected. In fact, using CTDI$_w = 0.070$ mGy/mAs, the radiation dose in this case is 0.070 mGy/mAs × 42 mAs × 1/0.28 = 10.5 mGy. For spiral scanning, some manufacturers, such as Siemens and Philips, have introduced the concept of an “effective” mAs, which includes the factor 1/pitch into the mAs definition:

$$ (\text{mAs})_{\text{eff}} = \text{mA} \cdot T_{\text{rot}} \cdot 1/pitch = \text{mAs} \cdot 1/pitch $$

(4.14)

For spiral scans, (mAs)$_{\text{eff}}$ is indicated on the user interface. Inserting Eq. 4.14 into Eq. 4.13, the dose of a multi-slice spiral CT scan is given by

$$ \text{CTDI}_{\text{vol}} = (\text{mAs})_{\text{eff}} \times (\text{CTDI}_w)_n $$

(4.15)

Other manufacturers (i.e., Toshiba and GE) have retained the conventional mAs definition, so that the user has to perform the 1/pitch correction. When comparing the scan parameters for CT systems of different manufacturers, the underlying mAs definition has to be taken into account. In the above example of an ECG-gated multi-slice spiral scan for coronary calcium quantification, the user applies 42 mAs, but 42 mAs × 1/0.28 = 150 effective mAs.

CTDI$_w$ is a measure of physical dose; it does not provide full information on the radiation risk associated with a CT examination. For this purpose, the concept of “effective dose” has been introduced by the ICRP (International Commission on Radiation Protection). The effective dose is given in milli-Sieverts (mSv). It is a weighted sum of the dose applied to all organs in a CT examination and includes both direct and scattered radiation. The weighting factors depend on the biological radiation sensitivities of the respective organs. Effective dose can be measured using whole-body phantoms, such as the Alderson phantom, or it is obtained by computer simulations using Monte Carlo techniques to determine scattered radiation. The effective patient dose depends on the scanned range. For a comparison of effective dose values for different protocols, scan ranges should be similar. A suitable program for calculation of effective dose values is, e.g., WinDose (Kalender 1999). This is a PC-based program that calculates organ dose and effective dose values for arbitrary scan parameters and anatomical ranges. Values for primary radiation have to be measured in terms of CTDI$_w$ and are used as input; values for scattered radiation are derived from Monte Carlo calculations.

The user has to keep in mind that both CTDI$_w$ and effective patient dose, which is derived from CTDI$_w$, give a correct estimation of the radiation dose for the patient only when the cross-section of the patient’s anatomy is comparable in size to the cross-section of the phantom used for the evaluation of CTDI$_w$ (32-cm Lucite for the body). For smaller patients, the dose may be considerably higher (Fig. 4.45).
Radiation Exposure for Selected Cardiac Examination Protocols

Typical values for the effective patient dose of selected multi-slice CT protocols are 1–2 mSv for a head examination, 5–7 mSv for a chest CT, and 8–11 mSv for CT of the abdomen and pelvis (McCollough 2003, Morin 2003). This radiation exposure must be appreciated in the context of the average annual background radiation, which is 2–5 mSv (about 3.6 mSv in the US).

This section provides estimates of the effective patient dose for selected cardiac CT protocols with 16-slice and 64-slice CT systems, using the Siemens SOMATOM Sensation 16 and Sensation 64 scanners as representative examples. The effective patient dose for the standard protocols recommended by the manufacturer was calculated with WinDose (Kalender 1999). The scan protocols included are ECG-triggered coronary calcium quantification, ECG-gated coronary calcium quantification, and ECG-gated coronary CT angiography. The scan range used is 12 cm for coronary calcium quantification and 10 cm for coronary CT angiography. The scan parameters recommended by the manufacturer are listed in Table 4.2. For sequential scans, the mAs values are based on the slice exposure times for an axial scan that is the partial-scan time. For spiral

### Table 4.2. Scan protocols used for calculation of the effective patient dose

<table>
<thead>
<tr>
<th></th>
<th>ECG-triggered calcium scoring</th>
<th>ECG-gated calcium scoring</th>
<th>ECG-gated cardiac/coronary CT angiography</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Sensation 16</td>
<td>Sensation 64</td>
<td>Sensation 16</td>
</tr>
<tr>
<td>Collimation (mm)</td>
<td>12 × 1.5</td>
<td>30 × 0.6</td>
<td>16 × 1.5</td>
</tr>
<tr>
<td>Reconstructed slice (mm)</td>
<td>3.0</td>
<td>3.0</td>
<td>3.0</td>
</tr>
<tr>
<td>Rotation time (s)</td>
<td>0.42</td>
<td>0.375</td>
<td>0.375</td>
</tr>
<tr>
<td></td>
<td>0.42</td>
<td>0.375</td>
<td>0.375</td>
</tr>
<tr>
<td>kV</td>
<td>120</td>
<td>120</td>
<td>120</td>
</tr>
<tr>
<td>mA</td>
<td>100</td>
<td>111</td>
<td>111</td>
</tr>
<tr>
<td>mAs</td>
<td>30</td>
<td>30</td>
<td>30</td>
</tr>
<tr>
<td>Pitch</td>
<td>n.a.</td>
<td>n.a.</td>
<td>0.28</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Table feed</td>
<td>18 mm per axial scan</td>
<td>18 mm per axial scan</td>
<td>16 mm/s</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Effective mAs</td>
<td>n.a.</td>
<td>n.a.</td>
<td>n.a.</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>(CTDI&lt;sub&gt;w&lt;/sub&gt;)&lt;sub&gt;n&lt;/sub&gt; (mGy/mAs)</td>
<td>0.072</td>
<td>0.072</td>
<td>0.072</td>
</tr>
<tr>
<td>CTDI&lt;sub&gt;w&lt;/sub&gt; (mGy)</td>
<td>2.2</td>
<td>2.2</td>
<td>2.2</td>
</tr>
</tbody>
</table>

Fig. 4.45. CTDI<sub>w</sub> at 120 kV for the 32-cm body phantom as a function of the total collimated width of the detector for 4-slice (black line) and 16-slice (blue line) CT systems with similar geometries.
scans, effective mAs values are used; these have been already corrected for spiral pitch, i.e., they take into account the dose accumulation with decreasing spiral pitch (see above). Table 4.3 summarizes the CTDIw values for the 32-cm body-phantom and values of the effective patient dose for the recommended standard protocols, for the 16- and the 64-slice CT systems. In Table 4.3, the scan parameters listed in Table 4.2 were used.

For ECG-triggered coronary calcium quantification with sequential scanning, patient dose is about 0.5 mSv for males and 0.68 mSv for females. Coronary calcium quantification with ECG-gated spiral scanning that results in reduced interscan variability (Ohnesorge 2002) increases the patient dose to 2.3–2.8 mSv for males and 3.3–3.9 mSv for females. ECG-gated high-resolution coronary CT angiography using thin-slice spiral data acquisition and yielding both adequate visualization of the small and complex cardiac and coronary anatomy and detection and classification of coronary plaques requires a dose of 7.9–10.8 mSv for males and 11.1–15.2 mSv for females. Improved longitudinal resolution by thinner collimation directly translates into increased patient dose if the signal-to-noise ratio of the images is maintained.

A comparison of radiation doses for different CT scanner types requires that imaging parameters, such as slice width, tube voltage, tube current, and scanned volume, be taken into account. If the high spatial resolution of modern multi-slice CT systems achieved with sub-millimeter slice collimation is not required, decreased longitudinal spatial resolution with slice-widths between 1.5 and 3.0 mm can always be traded off with a correspondingly reduced patient dose.

The pitch values in the ECG-gated multi-slice spiral CT protocols shown in Table 4.2 are selected such that the vast majority of patients can be scanned without significant protocol adjustments, including patients with low heart rates. Radiation exposure, however, can be reduced by increasing the pitch in patients with sufficiently high heart rates when respecting the constraint considerations given in Section 4.4.3 and in Figure 4.16. The radiation exposure is then reversed proportional to the pitch increase, as compared to the values presented in Tables 4.2 and 4.3.

Table 4.3. Values for the effective patient dose based on the scan protocols listed in Table 4.2

<table>
<thead>
<tr>
<th></th>
<th>ECG-triggered calcium scoring</th>
<th>ECG-gated calcium scoring</th>
<th>ECG-gated cardiac/coronary CT angiography</th>
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<td>Sensation64</td>
<td>Sensation16</td>
</tr>
<tr>
<td>Collimation (mm)</td>
<td>12 × 1.5</td>
<td>30 × 0.6</td>
<td>16 × 1.5</td>
</tr>
<tr>
<td>Rotation time (s)</td>
<td>0.42</td>
<td>0.375</td>
<td>0.375</td>
</tr>
<tr>
<td>CTDIw (mGy)</td>
<td>2.2</td>
<td>2.2</td>
<td>2.2</td>
</tr>
<tr>
<td>Scan range (mm)</td>
<td>120</td>
<td>120</td>
<td>120</td>
</tr>
<tr>
<td>Effective patient dose (mSv), males without ECG-pulsing</td>
<td>0.5</td>
<td>0.5</td>
<td>0.5</td>
</tr>
<tr>
<td>Effective patient dose (mSv), females without ECG-pulsing</td>
<td>0.68</td>
<td>0.68</td>
<td>0.68</td>
</tr>
<tr>
<td>Effective patient dose (mSv), males with ECG-pulsing</td>
<td>n.a.</td>
<td>n.a.</td>
<td>1.2–1.6</td>
</tr>
<tr>
<td>Effective patient dose (mSv), female with ECG-pulsing</td>
<td>n.a.</td>
<td>n.a.</td>
<td>1.6–2.2</td>
</tr>
</tbody>
</table>
4.6.3 Exposure Reduction with ECG-Gated Tube-Current Modulation

The relatively high radiation exposure for ECG-gated multi-slice spiral imaging of the heart is caused by continuous X-ray exposure and data acquisition at low and highly overlapping spiral pitch (pitch << number of slices). The low spiral pitch is a consequence of the phase-consistent coverage of the heart volume in specific phases of the cardiac cycle. However, if the spiral pitch is limited such that one phase can be covered consistently, all other phases of the cardiac cycle can be covered as well. If reconstruction in different cardiac phases is not needed and instead only a very limited interval (i.e., diastolic phase) in the cardiac cycle is targeted during reconstruction, a significant portion of the acquired data and radiation exposure is redundant.

Prospective ECG triggering combined with “step-and-shoot” acquisition of axial slices has the benefit of smaller patient dose compared to ECG-gated spiral scanning, since scan data are acquired in the previously selected heart phases only. It does, however, not provide continuous volume coverage with overlapping slices and misregistration of anatomical details cannot be avoided. Furthermore, reconstruction of images in different phases of the cardiac cycle for functional evaluation is not possible. Since ECG-triggered axial scanning depends on a reliable prediction of the patient’s next RR-interval based on the mean of the preceding RR-intervals, the method encounters its limitations for patients with severe arrhythmia. To maintain the benefits of ECG-gated spiral CT but reduce patient dose ECG-controlled dose modulation has been developed. On-line reduction of the tube output in each cardiac cycle during phases that are of less importance for ECG-gated reconstruction has a high potential for exposure reduction. The nominal tube output is only required during those phases of the cardiac cycle that will be reconstructed. During ECG-gated tube-current modulation, the tube output is modulated on-line with prospective ECG control (Ohnesorge 2000b, Jacobs 2002, Poll 2002), and commonly used ECG-gated reconstruction algorithms can be further used unchanged. Within every cardiac cycle, tube output is raised to the nominal level during a limited interval in a pre-selected phase (usually the diastolic phase) in which the data are most likely to be reconstructed with thin slices and a high signal-to-noise ratio needs to be maintained. During the remaining part of the cardiac cycle, the tube output can be reduced by about 80% by a corresponding decrease of the tube current $mA_{\min} = 0.2 \ mA_{\max}$ (Fig. 4.46). Thus, continuous volume reconstruction is still possible in all phases of the cardiac cycle. In particular, functional imaging is still feasible, as it does not require thin slice reconstruction. For imaging during phases of reduced tube output, an appropriate signal-to-noise ratio can be maintained by primary or secondary reconstruction of thicker slices. The position of the windows of nominal tube output within the heart cycles needs to be defined prior to the scan in relation to the phase targeted for reconstruction.

The width of the time interval $\Delta T_N$ with nominal tube output during diastole has to be selected such that patient-individual shifting of the ECG gating interval is still possible to obtain the best possible image quality. Additionally, a well-defined overlap of $\Delta T_N$ with the window of temporal resolution (i.e., 165–250 ms) can compensate for inconsistent prospectively ECG-controlled timing of nominal tube output due to changes in heart rate during the scan. In addition, an extended length of the temporal window with nominal tube output allows for high-resolution reconstruction within an extended duration during the cardiac cycle; this may be useful for optimization of image quality. An appropriate trade-off of exposure reduction and ECG gating flexibility can be achieved with the selection of $\Delta T_N = 400$ ms. Shorter $\Delta T_N = 300$ ms for further exposure reduction may be possible for scanners with very fast rotation times ($\geq 0.37$ s). However, the possibility to reconstruct high-resolution images within an extended range of the cardiac cycle is then limited. The relative exposure reduction with ECG-controlled tube-current modulation as a function of heart rate is shown in Figure 4.47. For normal heart rates between 50 and 90 bpm, the exposure is reduced by 35–50% for $\Delta T_N = 400$ ms and by 45–60% for $\Delta T_N = 300$ ms. Radiation exposure savings are maximized for low heart rates, as the total time with low tube output during the scan is high. For increasing heart rates, the relative reduction decreases, as the time intervals of low tube output
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Fig. 4.46. Retrospectively ECG-gated 4-slice spiral scanning with prospectively ECG-controlled tube-current modulation for reduced radiation exposure. Tube current is at a nominal value (time interval $\Delta T_N$) during diastole and is reduced by 80% during phases of high cardiac motion.

Fig. 4.47. Exposure reduction with ECG-controlled tube-current modulation dependent on heart rate for different time intervals $\Delta T_N$ of nominal tube output. A 35–55% reduction of exposure can be achieved for regular heart rates between 50 and 90 bpm using $\Delta T_N = 400$ ms.

are shorter. However, for very high heart rates, the selection of a higher spiral pitch can decrease exposure via shorter scan time as compared to low heart rates. Figure 4.48 directly compares two consecutive scans for quantification of coronary calcification in the same patient. The scans were performed without and with ECG-controlled tube output modulation, and equivalent reconstructions were carried out in the diastolic phase. For the patient with a stable heart rate between 58 and 61 bpm, an exposure reduction of 51% could be achieved with the same image and diagnostic qualities of the clinical result. An example for a CT angiography scan of the heart and coronary arteries using ECG-controlled tube-current modulation is shown in Figure 4.49. High signal-to-noise ratio is available for reconstruction during diastole with thin sections and high resolution. Reconstruction during systole for functional information shows lower signal-to-noise ratio; however, diagnostically sufficient image quality can be achieved with secondary multi-planar reformation (MPR) reconstruction. For the patient with a stable heart rate
Fig. 4.48a–c. Case example of two repeated ECG-gated spiral scans of the same patient using a 4-slice CT scanner at 0.5-s rotation time and 4 × 2.5-mm collimation for imaging calcified plaques in the coronary arteries by applying ECG-controlled tube-current modulation. The patient showed stable sinus rhythm at heart rate 59 bpm during both scans. Calcifications in the left anterior descending and right coronary arteries that were identified with the first scan (a) can be accurately reproduced with the second scan (b). By ECG-controlled modulation of the tube current between 100 mA during diastole and 20 mA during systole (c), the amount of radiation can be reduced by 51% without compromising the signal-to-noise ratio for 59 bpm average heart rate. (Images courtesy of Klinikum Grosshadern, Munich, Germany)

between 74 and 77 bpm, the exposure reduction with ECG-gated tube-current modulation was about 37%. Clinical studies based on patients separated in two groups with similar patient parameters (i.e., patient size, heart rate variations, median heart rates) have demonstrated dose reductions of 40–50% in male and female patients without remarkable influence on image noise and image quality by using ECG-controlled tube-current modulation (Jakobs 2002).

Where available, this technique should be used for all patients with reasonably steady heart rates (Trabold 2003). However, if substantial arrhythmia is expected during the scan, prospective control of the tube-current modulation and retrospective positioning of the reconstruction intervals may not match and ECG-controlled tube-current modulation may not be adequately applied. To allow for more reliable use of ECG-controlled tube-current modulation in patients with irregular heart rates, new developments are under way that employ intelligent ECG analysis algorithms that detect extrasystolic beats in real time and disenable or adjust the temporal window for tube-current reduction in the affected heart beats (Fig. 4.50).

4.6.4 Optimization for Different Patient Sizes

In both ECG-triggered sequential scanning and ECG-gated spiral scanning, adaptation of the dose to patient size and weight is an important potential for
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Fig. 4.49a–c. Example of a coronary CT angiography examination using a 64-slice CT scanner with $64 \times 0.6$-mm slices, 0.33-s rotation time, and ECG-controlled dose modulation. **a** End-diastolic image with full dose, reconstructed at 70% of the cardiac cycle; **b** end-systolic image with reduced dose, reconstructed at 20% of the cardiac cycle; **c** The ECG signal of the patient shows a stable heart rate between 74 and 77 bpm. Although the signal-to-noise ratio of the end-systolic image is reduced, it is considered adequate for functional evaluation. (Images courtesy of University of Erlangen, Germany)

Fig. 4.50a, b. Automatic detection of extra-systolic heart beats during ECG-controlled tube-current modulation in a patient with a regular heart rate of 75 bpm. **a** An extra-systolic heart beat may lead to a mismatch in several consecutive heart beats of the temporal window with maximum tube output and desired cardiac phase for high-resolution image reconstruction. **b** If an extra-systolic heart beat is detected, the reduction of tube current can be disenabled during the affected heart beats and image quality can be maintained throughout the entire scan.
dose reduction. For quantification of coronary calcifications, a reduction of the mean effective patient dose by 11.8% for males and 24.8% for females in a group of 50 patients was demonstrated by individual body-weight-adapted tube-current settings (Mahnken 2003). For cardiac and coronary CT angiography, a similar potential for radiation dose reduction has been reported (Jung 2003). The International Consortium of Standardization in Cardiac CT, a group of scientists, physicians, and CT manufacturers, has published recommendations to generate more accurate and calibrated results for coronary artery calcification independent of the scanner and of patient size. The researchers used an anthropomorphic calibration phantom with fixed amounts of calcium to derive dose recommendations for small, medium-sized, and large patients. The anthropomorphic calibration phantom consisted of tissue-, water- and bone-equivalent materials (QRM, Quality Assurance in Radiology and Medicine GmbH, Möhrendorf, Germany) and is shown in Figure 4.51 (Ulzheimer 2003). To account for different patient sizes, the original phantom can be used with up to two additional attenuation rings. The three categories for patient size are defined as small: < 32.0-cm lateral thickness, medium: 32.0- to 38.0-cm lateral thickness, and large: > 38.0-cm lateral thickness. An A/P localizer image (topogram, scout view) is used to determine the lateral thickness, which is measured from skin-to-skin, at the level of the proximal ascending aorta. Target noise levels of 20 HU for small and medium-sized patients and 23 HU for large patients in the corresponding calibration phantom are proposed. For ECG-gated spiral scanning of coronary artery calcification using a 16-slice CT scanner with 16 × 1.5-mm collimation, 0.42-s rotation time, and 120 kV, the following effective mAs values are feasible: small = 80–85 effective mAs, medium = 200–240 effective mAs, and large = 450–500 effective mAs. Similar relations can be found for high-resolution and cardiac coronary CT angiography protocols in which target noise levels of 15–20 HU with thin-slice reconstruction of 1 mm and below need to be achieved. The above example demonstrates the high potential for dose reduction, especially in medium-size and small patients, for coronary calcium quantification as well as cardiac and coronary CTA imaging protocols. In some modern CT scanners, the mAs value applied during an ECG-gated spiral scan can be automatically adapted to the average size of the patient (e.g., CARE Dose4D, Siemens, Forchheim, Germany).

4.6.5 Optimization of Contrast-to-Noise Ratio

Another means to reduce the radiation dose is to adapt the X-ray tube voltage to the intended application. Typically, 120 kV are used for quantification of coronary calcifications and for coronary CT angiography. In these two applications, however, the contrast-to-noise ratio for fixed patient dose can be increased by decreasing the X-ray tube voltage. As a consequence, to obtain a desired contrast-to-noise ratio, the patient dose can potentially be reduced by choosing lower kV settings of 80 or 100 kV. The potential for dose saving is more pronounced for small and medium-sized patients. Phantom measurements using small tubes filled with diluted contrast agent embedded in Lucite phantoms with different diameters (Schaller 2001) were used to demonstrate the potential to increase contrast and reduce dose in general vascular and cardiac CT angiography examinations in small and medium-sized patients when using 100 kV. Initial clinical experiences have confirmed this finding in patients weighing up to 90 kg (Fig. 4.52). Other studies (Jacobs 2003) compared ECG-gated spiral scan protocols using 120 and 80 kV for the quantification of coronary calcium. An 80-kV scan protocol was used in combination with ECG-controlled tube-current modulation, resulting in a significantly lower patient radiation exposure (0.72 mSv for 80 kV compared to 2.04 mSv for 120 kV). CTDI phantom measurements revealed a 65% reduction of radiation dose with the 80-kV protocol. In this study, the 80-kV protocol led to a significant reduction in radiation exposure during multi-slice CT coronary artery calcium screening, but did not affect the detection and quantification of coronary artery calcification. However, it should be noted that the maximum X-ray tube current available at 80 kV is generally not sufficient to scan larger patients and an increase in the tube voltage is required. In such patients, use of the 100-kV tube voltage may be a good compromise.
Fig. 4.52. Case example of a CT angiographic examination of the coronary arteries in a patient weighing 82 kg. The image was obtained on a 64-slice CT scanner with $64 \times 0.6$-mm slice acquisition and a 0.33-s rotation time. To reduce radiation exposure, 100-kV tube voltage and ECG-controlled tube-current modulation were used. The use of 100 kV instead of 120 kV reduces radiation exposure by about 33%, from 6.8 to 5.0 mSv. The high contrast-to-noise ratio allows visualization of the entire 3D cardiac and coronary anatomy and of atherosclerotic plaques in the left anterior descending and circumflex coronary arteries using an overlay of VRT and MIP displays. (Images courtesy of Medical University of South Carolina, Charleston, USA)

Fig. 4.51. Anthropomorphic calibration phantom (QRM Quality Assurance in Radiology and Medicine GmbH, Möhrendorf, Germany) recommended by the International Consortium for Standardization in Cardiac CT. The original phantom can be used with up to two additional attenuation rings to simulate small, medium-sized, and large patients.

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