Introduction

Since its clinical introduction in 1991, volumetric CT scanning using spiral or helical scanners has resulted in a revolution for diagnostic imaging. In addition to new applications for CT, such as CT angiography and the assessment of patients with renal colic, many routine applications such as the detection of lung and liver lesions have substantially improved. Helical CT has improved over the past eight years with faster gantry rotation, more powerful x-ray tubes, and improved interpolation algorithms [1, 2]. However, in practice the spiral data sets from monoslice systems suffered from a considerable mismatch between the transverse (in plane) and the longitudinal (axial) spatial resolution. In other words the isotropic 3-dimensional voxel could not be realized apart from some very specialized cases [3]. Similarly, in routine practice a number of limitations still remained which prevented the scanning protocol from being fully adapted to the diagnostic needs [4].

The introduction of subsecond spiral scanning with the SOMATOM Plus 4 at the RSNA 1994 was a first step to facilitate routine clinical work with respect to scannable volumes, total scan time and axial resolution [5]. Compared the 1 sec scanners which were standard at that time, the 750 ms rotation time allowed one to scan 33% longer volumes or to correspondingly reduce the total scan time or to correspondingly improve axial resolution.

The latest advance has been the recent introduction of multislice CT (MSCT) scanners. At the RSNA 1998, this new technology was introduced by several manufacturers representing an obvious quantum leap in clinical performance [6-8]. Currently capable of acquiring four channels of helical data simultaneously, MSCT scanners have achieved the greatest incremental gain in scan speed since the development of helical CT and have profound implications for clinical CT scanning.

Fundamental advantages of MSCT include substantially shorter acquisition times, retrospective creation of thinner or thicker sections from the same raw data, and improved three-dimensional rendering with diminished helical artifacts [9].

For example, the SOMATOM® Volume Zoom with a 500 ms rotation time and the simultaneous acquisition of 4 slices offers an 8-fold increase of performance compared to previous 1 s, single-slice scanning.

Obviously, such a quantum leap opens up a new area in spiral CT affecting all existing applications and allowing the realization of new clinical applications. The key issue is correspondingly increased volume coverage per unit time at high axial resolution and a correspondingly improved temporal resolution [6].

In general terms, the capabilities of spiral CT can be expanded in various ways: to scan anatomical volumes with standard techniques at significantly reduced scan times; or to scan larger volumes previously not accessible in practical scan times; or to scan anatomical volumes with high axial resolution (narrow collimation) to closely approach the isotropic voxel of high-quality data sets for excellent 3-dimensional postprocessing and diagnosis.

Basic Principles

For discussion of spiral imaging the following definition of pitch is commonly used:

\[ P = \frac{\text{Table travel per rotation}}{\text{Collimation of single slice}} \]

With this definition the present standard range \( 1 \leq p \leq 2 \) for single slice systems is extended to \( 1 \leq p \leq 8 \). In Fig. 1 various sampling patterns of a 4-slice spiral are shown for representative values of the pitch. In this schematic view the projections are symbolized by single arrows, which for simplicity are drawn in parallel.

From those examples we can draw the following conclusions:

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**Multislice Computed Tomography: Basic Principles and Clinical Applications**

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1. For pitch values smaller than 4 the four slices overlap to a certain degree after one rotation. Pitch 2 is a transparent case with double sampling. Therefore in this regime $1 \leq p \leq 4$ the multiple sampling can be used to reduce the tube current for a desired image noise and for a desired patient dose, respectively.

2. However, collecting data from multiple rotations degrades the temporal resolution of the system. Therefore for imaging of moving organs with high image quality, a pitch smaller than 4 should be avoided.

3. The distance between neighboring samples varies with pitch periodically. Consequently, $180^\circ$ LI and $360^\circ$ LI spiral interpolators would yield a corresponding non-monotonic dependence of slice width on pitch and are therefore not useful for practical purposes.

4. The distance between neighboring samples never exceeds the slice collimation up to a pitch of 8. This opens up the possibility to realize a slice width independent of pitch up to a pitch of 8 and to completely eliminate broadening of the slice width.

**Design Considerations for MSCT**

It is easy to design a multi-slice scanner for a fixed slice collimation, the challenge is to design the detector in such a way as to meet the clinical requirement of different slice collimations adjustable to the diagnostic needs. There are basically two different approaches, the matrix detector with elements of a fixed size or the adaptive array principle. Both principles will be briefly described and compared.

An example of a matrix detector is sketched in Fig. 2. In axial direction the detector is divided into 16 small elements each providing a 1 mm thick slice at the center of rotation [10]. The outer detector rows cannot be used individually. In the example of Fig. 2 the 1 mm slice is smeared over about 6 mm. The only way out is to sum the signals of the outer rows to generate thick slices. However, the unnecessary mechanical cuts and
optical separations between the small elements, correspondingly reduce the geometrical efficiency and therefore the dose efficiency of the system. In summary, the matrix detector is well suited to scan at 4 x 1 mm collimation but not more than 4. Wider collimations 4 x 2 mm, etc. can be realized by signal combination during read out but at the expense of dead zones and a corresponding waste of dose. These arguments lead to the development of the Adaptive Array Detector [9].

The design of the Adaptive Array Detector (AAD) takes into account the cone beam constraints for optimal image quality, optimizes the dose efficiency and in conjunction with an Adapted Axial Interpolator (AAI) provides a flexible selection of slice widths [6, 11, 12]. The design principle is depicted in Fig. 3. Narrow detector elements are close to the center; the width of the detector rows increases with distance from the center. Unnecessary dead spaces are avoided and with the corresponding prepatient collimator and the proper read-out schemes the following combinations of collimation can be achieved: 2 x 0.5 mm, 4 x 1 mm, 4 x 2.5 mm, 4 x 5 mm, 2 x 8 mm and 2 x 10 mm (Fig. 4). These combinations represent the collimation of the X-ray beam at center: e.g. a 4 x 5 mm collimation means a X-ray beam width at center of 20 mm. Consequently with sequential imaging four 5 mm slices would be generated during one rotation.

In spiral imaging the variety of axial sampling patterns as a function of pitch allows both: obtaining slice widths independent of pitch and reconstructing a multiplicity of slice widths from a scan with collimation narrower than s. To achieve this the AAI-scheme provides a set of linear and nonlinear interpolators which are adapted to the desired pitch, collimation and slice width. To give some examples: from a spiral scan with 4 x 1 mm collimation slice widths of 1, 1.25, 1.5, 2.0, 2.5, 3.0, 4.0, 5.0 up to 10 mm can be obtained by adjusting the width and the functional form of the interpolator. On the other hand, a selected slice width s, like s = 5 mm, may be obtained from different collimator settings, like 4 x 1 mm or 4 x 2.5 mm. This is important to remember for clinical applications, as the narrower collimation is preferable for image quality reasons, i.e. the reduction of partial volume effects [9].

Yet another design criterion should be emphasized: spiral scanning at large pitch, e.g. large table velocity, implies a more severe inconsistency between direct projections and the complementary projections, taken half a rotation later with a quarter detector offset. The reason is changing anatomy in axial direction, which is particularly pronounced for large pitch applications, i.e. rapid table movement. Consequently, image quality at high spatial resolution and large pitch is becoming more dependent on the flying spot technology which provides a quasi-instant doubling of the in plane sampling rate.
In conclusion, image quality in multi-slice spiral scanning must be optimized with respect to several factors [13]:

1. The narrowest collimation, consistent with the coverage of a certain volume and with a certain scan time, to minimize partial volume effects and to optimize image quality.

2. Fastest rotation time to maximize z-coverage and to minimize motion blurring.

3. Pitch greater than 4 to preserve temporal resolution and to minimize motion blurring.

4. The exploitation of flying focal spot technology to avoid artifacts at high spatial resolution.

To measure section profiles of multi-spiral scanning a thin gold plate (thickness 50 µm) in air has been aligned orthogonal to the scanner axis and has been scanned in spiral mode. Some of the resulting slice sensitivity profiles (SSP) are shown in Fig. 5 for a 4 x 1 mm collimation, a slice width of 2 mm and different pitch values. Obviously the AAI indeed results in slice widths independent of pitch; but even more important, the functional form of the SSPs is also identical and practically independent of pitch. Slice broadening and long-range tails of the SSP, which prohibited the use of fast table speeds in single slice spiral scanning, can be completely avoided. From this perspective the whole range up to a pitch of 8 can be used for practical purposes in multi-slice scanning [6].

The second new and attractive feature of the multi-spiral AAI is illustrated in Fig. 6. From a scan with 4 x 1 mm collimation SSPs with different width can be obtained: 1.25 mm, 2.0 mm and 4.0 mm respectively. Excellent agreement between theory and measurement is observed. This feature is the basis of Combi Scans. This provides considerable flexibility in image reconstruction [14] especially for imaging of the base of the skull and lung (Fig. 7).

Anatomical structures that generate spiral artifacts are well known from single slice spirals. The most difficult situations arise from bony structures (high

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**Figure 5**
Slice Sensitivity Profiles, Collimation 4 x 1 mm, slice width 2 mm, different pitch values.

**Figure 6**
Slice Sensitivity Profiles, Collimation 4 x 1 mm, pitch 3, different slice widths.
Figure 7

Combi Scan: lung study with 4 x 1 mm collimation, pitch 6. Reconstruction of 5 mm images for soft tissue (a) and standard lung window (b); 1.25 mm high resolution images for detection of interstitial disease (c). HR-MPR (d) clearly depicts segmental anatomy.
contrast) which are strongly inhomogeneous in axial direction. A demanding example is the base of the skull with bony structures abruptly ending in an image plane. Phantom studies have shown that a larger pitch and narrower collimation is much more favorable for suppressing artifacts than a lower pitch and wider collimation for equal z-coverage. The underlying reason is the better elimination of partial volume artifacts. From image quality reasons, only, the strength of multislice spiral scanning is the ability to cover anatomical volumes with narrowest collimation and thereby to minimize partial volume effects. This is only a matter of scanning technique (collimation) and is independent of the selected slice width. For optimization of image quality we derive the rule: the narrowest collimation should be selected which is consistent with volume coverage and scan time. This generally results in large pitch values (e.g. from 4 to 6) which are also helpful to avoid motion blurring [13].

Clinical Applications

The advantages of MSCT are important to many applications of CT scanning, including survey exams in oncologic or trauma patients and the characterization of focal lung and liver lesions through the creation of thin sections retrospectively. However, the greatest impact has been on CT angiography, cardiac imaging, virtual endoscopy, and high resolution imaging [15].

CT Angiography

A fundamental advantage of a multislice scanner over monoslice systems is its ability to obtain a first circulation study of a rapidly injected contrast bolus with thinner images. Determining circulation time either by a preliminary minibolus or by online bolus tracking software is important in matching the acquisition interval to the first-circulation time of the injected bolus. As a result of the shorter acquisition time, the contrast dose can be significantly reduced.

CT angiography of the intracranial vessels benefits from the quick and detailed scanning technology of MSCT. At 1 mm-collimation, the circle of Willis can be scanned within 10 seconds. This permits the entire scan to be completed during the first pass of iodinated contrast material through the arterial system. The use of MSCT with CT angiography demonstrates the spatial relationship of an aneurysm with the feeding vessel, as well as the shape of the aneurysm itself, because the same bolus can be followed continuously throughout its course. For CT angiography of the thoracoabdominal aorta a collimation of 2.5 mm, a pitch of 5-6 at a rotation speed of 0.5 s are used. Volume coverage is from the thoracic inlet to the inguinal region, a 50 to 55 cm area that can be covered in an acquisition interval of approx. 20 s (Fig. 8). Reconstructions with 50% overlap are used, generating 400-450 images for the three-dimensional data set. A complete lower extremity study

Figure 8
MSCTA of the thoracoabdominal aorta. Stanford Type B Dissection. MPRs from images with 3 mm slice-width and 1 mm increment. The spiral scan was acquired with 4 x 2.5 mm collimation, pitch 5, and 0.5 s rotation time. The total scan time for a spiral length of 550 mm was 22 s. Courtesy of Dr. Baum, Erlangen.
can be performed from the level of the renal vascular pedicles to the ankles. A 2.5 mm collimation, a pitch of 6 and 50% overlap reconstructions are used (Fig. 9). A first-pass circulation study can be obtained without venous overlay. The total number of images generated is 800 to 1000. With multislice CT operating at pitch 6 and collimation of 4 x 1 mm, a CTA of the pulmonary arteries can be acquired in less than 25 s. Image thickness of 1.25 mm significantly increases detectability of subsegmental emboli in comparison to monoslice spiral CT using 2 to 3 mm image thickness [16]. CTA of the visceral branch vessels can be performed significantly faster and/or with increased spatial resolution. Using a

Figure 9
Peripheral Runoffs. Spiral scan with 4 x 2.5 mm collimation, pitch 6, 0.5 s rotation time.
The total scan time for a spiral length of 1000 mm was 34 s.

Left: MIP from images with 3 mm slice width, 2 mm increment.
Right: VRT from images with 6 mm slice width, 4 mm increment.

Figure 10
CTA (a, b) of visceral branch vessels (collimation 4 x 1 mm, pitch 5, 120 cc contrast).
On the VR images a stenosis in the SMA (arrows) can be readily appreciated which was confirmed at conventional angiography (c).
collimation of 1 mm allows for detection of even minor branches (Fig. 10). The ability to cover the entire chest in 10 s allows the scanning of children without sedation. This can be used to easily perform CTA of the large thoracic vessels before or after surgery of complex malformations [8].

**Cardiac Imaging**

Electron Beam CT scanning (EBCT) has been established as a non-invasive imaging modality for imaging coronary calcification. Major clinical applications are the detection and quantification of coronary calcium and non-invasive CT angiography (CTA) of the coronary arteries [17]. Current limitations of EBCT imaging include the limited reproducibility of coronary calcium quantification [18], the inability to detect non-calcified atherosclerotic plaques and the limited spatial resolution of 3D visualizations of the coronary arteries. Because of the restriction to axial, non-spiral scanning in ECG-synchronized cardiac investigations, acquisition of 3D volume images by using EBCT can only provide limited z-resolution within a single breath-hold scan. Retrospectively ECG-gated single-slice spiral scanning does not allow for sufficient continuous volume coverage within reasonable scan times. Retrospectively ECG-gated multi-slice spiral scanning, however, has the potential to completely cover the heart volume without gaps within one breath-hold [19]. Mechanical multi-slice CT systems with simultaneous acquisition of four slices, half-second scanner rotation and 125 ms maximum temporal resolution allows for considerably faster coverage of the heart volume, compared to single-slice scanning. This increased scan speed allows using thinner collimated slice widths and thus to increasing the z-resolution of high-resolution examinations such as CTA of the coronary arteries [20].

ECG-synchronized multi-slice spiral scans are acquired with heart rate dependent table feed (“pitch”) adaptation. Dedicated spiral algorithms provide 125 ms (60 ms as theoretical limit) temporal resolution and are optimized with regard to volume coverage. This allows reconstruction of overlapping images (increment < slice-width) at arbitrary z-positions and during any given heart phase [21]. This reconstruction technique combines partial scan reconstruction and multi-slice spiral weighting in order to compensate for table movement and to provide a well-defined slice sensitivity profile [19].

For retrospectively ECG-gated reconstruction each image is reconstructed using a multi-slice partial scan data segment with an arbitrary temporal relation to the R-wave of the ECG-trace. Image reconstruction during different heart phases is feasible by shifting the start point of image reconstruction relative to the R-wave. For a given start position, a stack of images at different z-positions covering a small sub-volume of the heart can be reconstructed owing to mutlislice data acquisition.

Fig. 11 shows an example of how the cardiac volume is successively covered with stacks of axial images (shaded stacks) reconstructed in consecutive heart cycles. All image stacks are reconstructed at identical time-points during the cardiac cycle. At the same time, the 4 detector slices travel along the z-axis relative to the patient table. In each stack, single-slice partial scan data segments are generated with equidistant spacing in the z-direction depending on the selected image reconstruction increment [21]. Continuous volume coverage can only be achieved, when the spiral pitch is adapted to the heart rate in order to avoid gaps between image stacks that are reconstructed using data from different heart cycles. In order to achieve full volume coverage, the image stacks reconstructed in subsequent heart cycles must cover all z-positions. Thus, the pitch, which can be used for image acquisition is limited by the patient’s RR-interval time [19].

The cardiac multi-slice data acquisition on the SOMATOM Volume Zoom are performed using 500 ms full rotation time and 4 x 1 mm or 4 x 2.5 mm collimated slice width. Non-contrast enhanced spiral scans for coronary calcium scoring are performed with 3 mm slice-width (SW = 3 mm, SWcoll = 2.5 mm) and 1 mm image reconstruction. For CTA of the coronary arteries and for functional heart imaging 2 different scan protocols with 3 mm slice-width and with 1.25 mm slice-width can be used (Fig. 12). For both protocols, non-ionic contrast material is intravenously injected at a flow rate of 3 ml/s. The delay times between start of contrast injection and scan start for optimal contrast enhancement is determined individually for each patient by injection of a 20 ml test bolus [20].
Figure 12
49-year-old male with CHD of the RCA. S/P posterolateral myocardial infarction 4 months ago.
Patient underwent balloon angioplasty of subtotal stenosis at the beginning of the descending part of the RCA.
Patient came in for follow-up performed with both coronary CT angiography (a) and conventional angiography (b).
Both conventional angiography and CT angiography (3D Virtuoso®, Siemens) clearly show the patency of the RCA at the level of the balloon angioplasty.
There is only minor residual stenosis of approx. 10-20% (arrows).
Scan: Siemens SOMATOM® Volume Zoom; 140 kV, 300 mAs, collimation 4 x 1 mm; slice width 1.25 mm, increment 0.6 mm.

Figure 13
Noninvasive CT coronary angiography: Volume rendered image (3D Virtuoso®, Siemens) depicts three high grade stenoses (arrows) in the LAD and at the origin of the diagonal branches. These findings are confirmed at conventional angiography.
The high spatial resolution, the absence of motion artifacts and the good overall image quality of the clinical MSCT images of the entire heart volume let appear multislice spiral CT as a promising modality for the non-invasive evaluation of coronary calcification. The scan time needed to acquire continuous ECG-gated multislice spiral CT image data is significantly reduced compared to EBCT (≈ factor 2.5) and single slice CT (≈ factor 5). Data with 3 mm slice width can be used for volumetric coronary calcium scoring. This alternative scoring method has the potential to improve the reproducibility of repeat calcium scoring compared to the conventional Agatston-score. Phantom studies have shown that non-overlapping sequential scanning is an important contributor to the inter-scan variability of Agatston- and volumetric Ca-scores due to partial volume errors in plaque registration [22]. ECG-gated volume coverage with multi-slice spiral CT and overlapping image reconstruction, however, was found to improve the reliability of coronary calcium quantification especially for small plaques. ECG-gated multi-slice spiral CT can potentially be of high value for coronary calcium evaluation especially for patients undergoing lipid-lowering statin therapy and for follow-up evaluations of patients after heart transplantation [23].

In contrast to sequential CT scanning z-resolution of ECG-gated spiral images with 3 mm slice-width can be improved by using overlapping reconstruction with 1 mm slice increment. Moreover, the fast scan speed allows covering the entire heart with 1.25 mm slices within a single breath-hold (10 cm in 25-35 s). 3D reconstruction with 1.25 mm slice-width and sub-millimeter image increment allows generating high-resolution visualizations of the coronary arteries, which may be suitable for a highly accurate diagnosis of coronary artery disease (Fig. 13, 14) [24]. Even more important, the first results indicate that MSCTA not only allows non-invasive imaging of coronary plaques but also assessment of plaque composition (i.e., soft, fibrous, calcified) (Fig. 15). Thus, this new technology holds promise to allow for the non-invasive imaging of rupture-prone soft coronary lesions and may have the option to lead to early onset of therapy [25].

**High Resolution Imaging**

Imaging of the temporal bone is a major challenge for clinical CT and a good example of the need for high resolution CT. Imaging of the temporal bone is improved using MSCT because the in-plane resolution can be greatly increased. The temporal bone is a structure of high intrinsic contrast and is routinely scanned with thin collimation. Both the 2 x 0.5 mm and the 4 x 1 mm mode on the SOMATOM Volume Zoom are applicable (pitch 2 or 3.5) and the tube current can be reduced to below 200 mAs (Fig. 16). The reconstruction kernels enable maximum spatial resolution up to 24 line pairs/
cm. At this resolution, the delicate structures of the middle and inner ear are sharply delineated and even subtle changes can be assessed. Another advantage of using MSCT to image the temporal bone is that the slice thickness is reduced to 1 mm or below, thus minimizing the partial volume effects and further increasing image quality of subtle structures [26].

**Impact on Radiological Practice**

The large number of images generated by MSCT is a major issue for workstation performance, film display, and PACS archiving. The images should be reviewed in a cine paging format at a workstation for diagnostic purposes [15]. Film hard copy will only be used for reference display. Planning for PACS must take into account this exponential growth in image data generated by this new technology. Rapid generation of 3D data sets is important for inclusion in the diagnostic study interpretation and for demonstration to referring clinicians. The hundreds or thousands of thin sections acquired using MSCT are incompatible with traditional viewing practice; thus MSCT will force a rapid transition of radiology from two-dimensional to volumetric imaging.

**Literature**


