ELSEVIER

European Journal of Radiology 45 (2003) S4–S10

EJR
EUROPEAN JOURNAL OF RADIOLOGY

www.elsevier.com/locate/ejrad

General principles of MDCT☆

Mathias Prokop *

University Medical Center Utrecht, Department of Radiology, Heidelberglaan 100, Utrecht NL-3508 GA, The Netherlands

Received 7 November 2002; received in revised form 30 November 2002; accepted 1 December 2002

Abstract

Multidetector CT (MDCT, multislice CT, multidetector-row CT, multisection CT) represents a breakthrough in CT technology. It has transformed CT from an transaxial cross-sectional technique into a true 3D imaging modality that allows for arbitrary cut planes as well as excellent 3D displays of the data volume. Multislice CT scanners provide a huge gain in performance that can be used to reduce scan time, to reduce section collimation, or to increase scan length substantially. The following article will provide an overview of the principles of multislice CT scanning. It describes the various detector systems and gives an introduction to the most important acquisition and reconstruction parameters. The article describes how reconstruction of thick multiplanar reformations can be used to take advantage of the 3D capabilities of multislice CT while keep radiation exposure to a minimum.

Keywords: Computed tomography; Radiation dose; Imaging parameters; Clinical protocols

1. Introduction

Multidetector CT (MDCT, multislice CT, multidetector-row CT, multisection CT) represents the next breakthrough in computed tomography (CT) technology. It has transformed CT from an transaxial cross-sectional technique into a true 3D imaging modality that allows for arbitrary cut planes as well as excellent 3D displays of the data volume [1,2] Multislice CT scanners provide a huge gain in performance that can be used to reduce scan time, reduce section collimation (SC), or to increase scan length substantially.

Multislice CT systems are equipped with two or more parallel detector arrays and always utilize a third-generation technology with synchronously rotating tube and detector array. Dual or split detector systems have been available since the early 1990s, systems with four active detector arrays were introduced in 1998, and systems with eight, 10, 16, or more active detector arrays became available in 2001 and 2002. Multislice CT has gained a rapid acceptance by the radiological community. In the first years, there was an almost exponential growths of the number of scanners: in 1998 there were ten scanners installed, in 1999 it was 100 by the middle of the year, and by the end of 2000, over 1000 scanners were in use worldwide.

The performance of many of these systems is further improved by a faster rotation time. As a result, a four-detector-row scanner with 0.5 s rotation has an about eight times higher performance than a conventional 1 s single-detector-row scanner.

2. Advantages

The performance of multislice CT with four-detector-rows is four to eight times higher than that of a conventional spiral CT scanner. With 16 detector rows, performance can even be more than 25 times greater (Table 2). In addition, data handling and image reconstruction have been improved with multislice CT, which directly translates into increased scanner productivity [3]. In consequence, multislice CT has overcome one of the most severe limitations of spiral CT, namely the inverse relation between scanning range and SC. Table 1 gives an overview of the advantages of multislice CT scanning.

☆ Third Bracco Symposium on MDCT, Paris, 14–15 June 2002.

* Tel.: +31-30-250-6687; fax: +31-30-258-1098.
E-mail address: m.prokop@azu.nl (M. Prokop).

0720-048X/03/$ - see front matter © 2002 Elsevier Science Ireland Ltd. All rights reserved.
doi:10.1016/S0720-048X(02)00358-3

Table 1
Advantages of multislice CT

Shorter scan duration
Reduced motion artifacts:
Children
Trauma patients
Acutely ill patients
Improved scanning of parenchymal organs
Well-defined phase of contrast enhancement
Perfusion imaging?
Longer scan ranges
CT angiography:
Aorta and peripheral run-off
Thoraco-abdominal aorta
Carotids from arch to intracerebral circulation
Thinner sections
Near-isotropic imaging:
Arbitrary imaging planes
Multiplanar reformats
3D rendering

3. Disadvantages

The downside of multislice CT is a markedly increased *data load*, especially if near-isotropic imaging is chosen [4,5]. For example a scan of the chest and abdomen (60 cm) can be performed with 16×1 mm collimation in less than 15 s and produces -600 images, depending on the degree of overlap. A chest scan with an identical collimation will produce as many images if reconstructed in a smoothing kernel for the mediastinum and an edge-enhancing kernel for the lungs. A CTA of the aorta and peripheral arteries may produce 1000 images and more.

Image noise grows as SC is reduced. For this reason, it is important to reconstruct thicker sections (MPR or axial sections) in order to keep image noise low. With very thin collimation the geometric efficiency of the scanners deteriorates. This effect can be seen at 1.25 mm-collimation or less and should not occur for larger collimation. It strongly varies between manufacturers and depends on the implementation of beam collimation and image interpolation algorithms. With 8- and 16-row scanners geometric efficiency improves again [6].

An increase in *patient dose* is only necessary if thin-section images of high quality are required. In all other cases, multislice CT requires less dose than conventional CT or a similar dose as spiral CT with a pitch of 2.

4. Detector types

Multislice scanners at present acquire two, four or eight simultaneous sections but with the exception of dual detector systems all multislice CT scanners have more than four-detector-rows in order to accomplish more than one collimation setting. This is done by proper collimation and adding together the signals of neighboring detector rows. Two types of detector arrays are available.

Matrix detectors [7] consist of parallel rows of equal thickness, while *adaptive array detectors* [1] consist of detector arrays with rows of varying thickness (Figs. 1–3). Both have advantages and disadvantages and no system is at present intrinsically superior to the other. Hybrid detectors use smaller detector rows in the center and larger ones towards the periphery of the detector array. They are used in all 16-slice scanners [6].

5. System performance

System performance is proportional to the number of detector rows N and improves with a shorter rotation time RT of the X-ray tube. This concept can be applied to standard spiral CT, as well as to dual detector scanners or multislice CT. Table 2 gives an overview over the relative performance of various scanner types and scanning protocols.

With a higher system performance, faster scans and thinner collimation become feasible. Multislice scanners with four-detector-rows allow for near-isotropic imaging of the chest or abdomen (section thickness <1.5 mm) but scan duration will be in the same range as with ‘conventional’ spiral CT. Only with eight- and 16-slice scanners can the scan duration be reduced substantially even in the case of near-isotropic scanning.

6. Scanning parameters

Like in spiral CT scanning, section collimation (SC), table feed per rotation (TF), and pitch (P) are the most important *acquisition parameters* in multislice CT. In addition to the reconstruction increment (RI), however, there is the effective section thickness or section width (SW) of the reconstructed images that contributes to the most important *reconstruction parameters* [8]. All the other parameters are varied only in exceptional cases. Together with the number of active detector rows N, the acquisition parameters can be given as ($N \times$ SC/TF), and the reconstruction parameters can be given as (SW/RI).

Two definitions of *pitch* factor are used with multislice CT scanners, depending on whether a section or the total collimation of the detector array ($N \times$ SC) is chosen as a reference. In order to distinguish between them, an asterisk is used to indicate the definition preferred by most manufactures (P*), while P denotes the ‘official’ $\in$C definition preferred by most physicists:

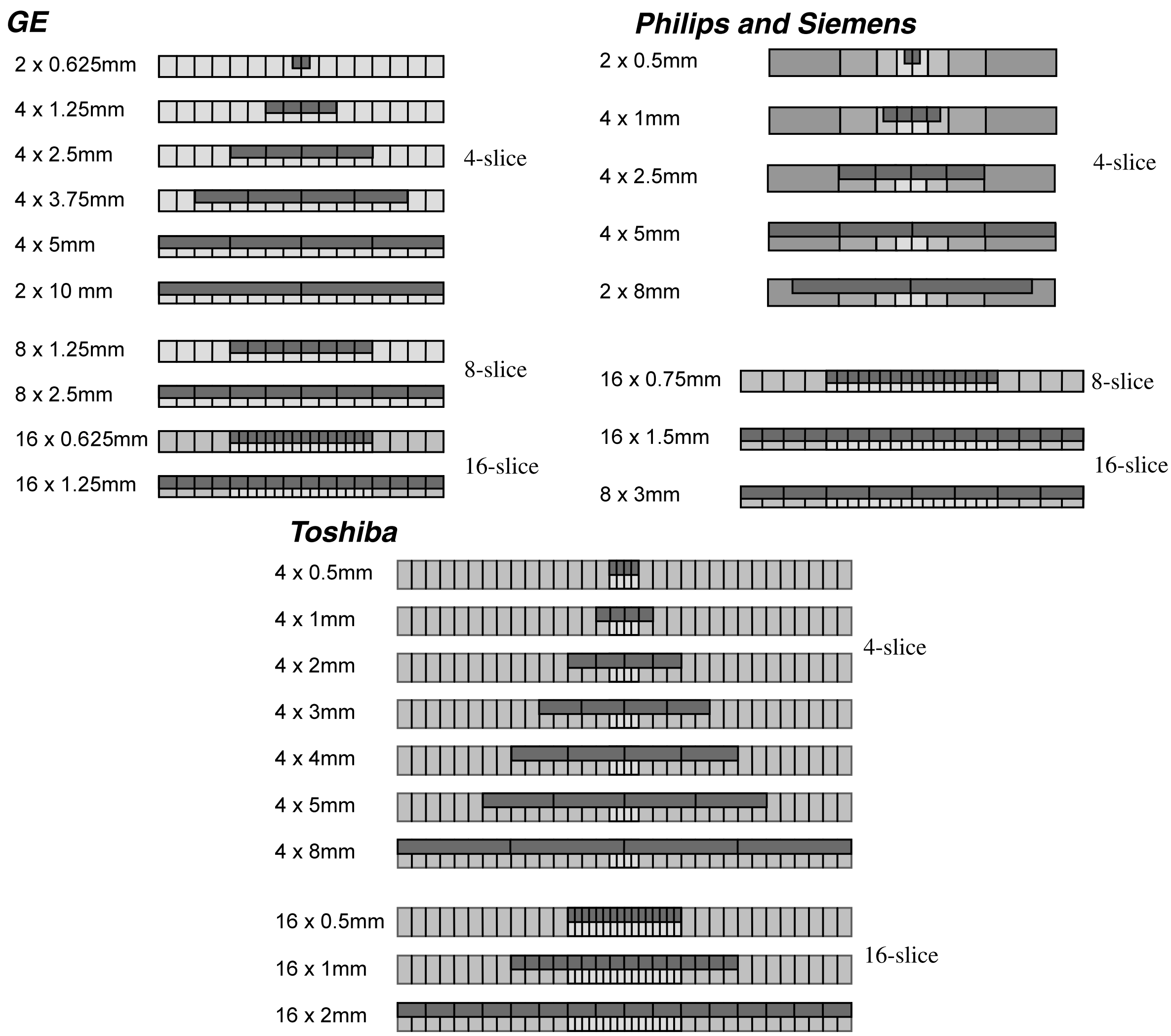

Fig. 1. Detector configurations. Note that the four- and eight-row GE detector is the only true matrix detector. The Toshiba detector contains four to 16-rows of 0.5 mm in the center of its detector array. The Philips and Siemens detectors consist of adaptive arrays whose width increases from 1 to 1.5, 2.5, and 5 mm for the four-slice system. The 16-slice scanner has a hybrid configuration like all other 16-row systems. In either case, the detectors are partially collimated from the sides and data from multiple detector rows are combined if necessary (e.g. the innermost 1 and 1.5 mm detectors for 4 × 2.5 mm collimation). * Philips only.

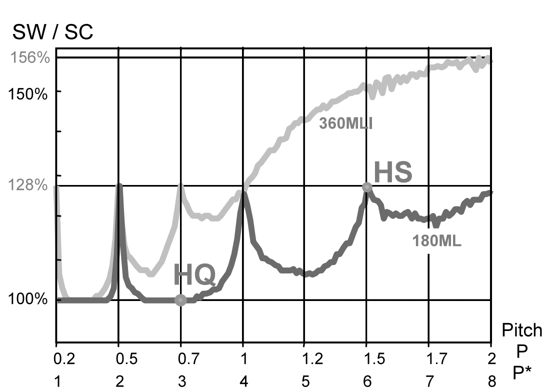

Fig. 2. Effective slice thickness (SW) varies strongly as a function of pitch for four-row multislice CT scanners if 180 MLI and 360 MLI raw data interpolation is used. The graph indicates SW as a percentage of SC. HQ = high quality mode. HS = high speed mode (GE scanners). Data courtesy of S. Schalles.

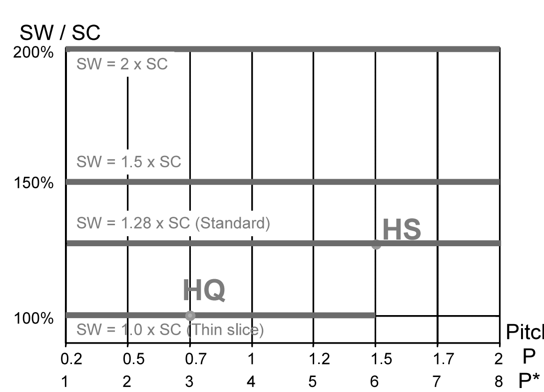

Fig. 3. Effective slice thickness (SW) remains independent of pitch if adaptive z-filtering is employed (Siemens) [1].

Table 2
System performance relative to a 1 s spiral CT scanner with $P = 2$ (example)

System	Rotation time (s)	Detector rows (N)	Pitch (P)	Pitch ($P*$)	Rel. performance
1 s-scanner	1	1	2	2	1
0.8 s-subsecond scanner	0.8	1	2	2	1.25
0.5 s-subsecond scanner	0.5	1	2	2	2
1 s-dual detector scanner	1	2	2	4	2
0.8 s-multislice scanner	0.8	4	0.75	3	1.875
0.8 s-multislice scanner	0.8	4	1.5	6	3.75
0.5 s-multislice scanner	0.5	4	1.5	6	6
0.5 s-multislice scanner	0.5	4	2	8	8
0.5 s-multislice scanner	0.5	8	1.35	10.8	10.8
0.5 s multislice scanner	0.42	16	1.5	24	28.5

$$P = \frac{\mathrm{TF}}{N \times \mathrm{SC}} \quad \text{pitch, 'detector pitch'}$$

$$P* = \frac{\mathrm{TF}}{\mathrm{SC}} \quad \text{'volume pitch', 'beam pitch'}$$

As with single slice spiral CT, the pitch P can be theoretically increased up to 2, independent of the number of detector rows N. In multislice scanners with four active detector rows this corresponds to $P* = 8$, in 16-row scanners this corresponds to $P* =$ 32. In practice, the maximum pitch varies between 1.5 and 2, depending on manufacturer and the number of detector rows of the scanner.

The SW (effective section thickness) can be chosen independent from SC as long as SW is larger or equal SC. The available choices for SW depend on the manufacturer and the type of multislice raw data interpolation and reconstruction. In 4-slice systems, raw data interpolation (z-filtering) can be based on algorithms that are analogous to 180° LI and 360° LI from spiral CT, but performance of these algorithms in terms of effective section thickness and noise increases and decreases a number of times as the pitch $P*$ is varied from 1 to 8 for four-detector-rows (Fig. 2) [7,9–12]. This behavior is due to the fact that the spiral trajectory of the first detector row may overlap with that of the second, third or fourth row ($P* = 1$, 2, and 3) and thus yield redundant data. For this reason, one vendor (GE) only made two distinct pitch factors and corresponding SWs available. Scanning with $P* = 3$ is called *HQ* (*'high quality'*) *mode*, and with $P* = 6$ it is called *HS* (*'high speed'*) *mode*. All other vendors allow for arbitrary pitches between 1 and 8. Toshiba, in addition, gives the choice of various *z-filters* that allow adapting SW and noise behavior to the individual need of the user. Siemens has decided to use an *adaptive z-filtering* that guarantees constant SW independent of the chosen pitch factor (Fig. 3). At the same time, image noise at a constant patient dose is also independent of the pitch. This latter approach makes life much easier for users since they no longer have to worry about the relation between pitch, dose and image quality.

With eight- and 16-row scanners the cone beam effect becomes more prominent and requires new algorithms for raw data interpolation and reconstruction. A number of techniques (*cone beam interpolation*) are currently being used that range for true 3D back-projection to reconstruction of oblique planes for every z-position that are then interpolated to form a true 3D data volume [13–17].

With GE scanners the user chooses the section width SW first depending on the clinical requirements, much like in single slice spiral CT, but in discrete steps that are multitudes of the minimum detector collimation (i.e. 1.25, 2.5, 3.75, 5, 7.5 and 10 mm) [7] for 4-slice scanners. The user then has to decide whether to scan with $4 \times$ 1.25 mm or with thicker collimation. For multiplanar reformations (MPR), the data set has to be reconstructed a second time later with thinner sections.

With all other scanners, the acquisition parameters are chosen first, and then one or more sets of reconstruction parameters can be determined that suit the clinical question. The reconstructed SW can be varied arbitrarily (usually in 1 mm steps) with the only precondition that it is larger than the chosen collimation SC.

In most *clinical settings*, the reconstructed SW will be similar to the numbers used in single-slice spiral CT. For the chest and abdomen, 5–7.5 mm will suffice for most routine settings [8]. For HRCT of the lungs or skeletal, a SW of 1.5–2 mm will yield good results.

As soon as imaging planes other than the primary axial sections are required, MPR will have to be performed from a thin-section overlapping axial data set. For this '*secondary raw data set*', thin section width SW and a reconstruction inverval RI that is roughly 50% of SW should be chosen. For small fields of view (FOV), RI need not be smaller than the pixel size (= FOV/512). For optimum signal-to-noise ratios, a SW that is some 25–30% wider than the collimation SC should be chosen (i.e. SW = 1.25 mm for 4×1 mm collimation) because image noise grows dysproportionally if SW is forced to be identical to SC. The secondary raw data set is then used to create P.R of arbitrary orientation (even axial) and a (through-plane) thickness that depends on the noise level and clinical imaging task. Only the thick images will be used for clinical reporting in most instances.

7. Radiation exposure

When identical milliamperes settings are used as compared with spiral CT scanners of the same vendors, patient exposure may be markedly increased [18]. This can be due to changes in scanner geometry that result in a higher CT dose index (CTDI) per milliamperes (e.g. GE) or due to the fact that no longer the milliampere settings are given but that the effective milliampere settings (mA_{eff} = m/s/P) are provided (e.g. Siemens). In the latter situation, a spiral CT scan that used 160 mAs with 5 mm collimation and 8 mm table feed ($P = 1.6 \Rightarrow$ 100 mAs_{eff}) would require the multislice scanner to be set to 100 mAs (mAs_{eff} to be more precise) for identical patient exposure. Some users, however, still use 160 mAs_{eff} and, therefore, increase dose to the patient.

Using thinner collimation will substantially increase image noise. This has to be at least partially compensated by increasing mAs_{eff} as compared with spiral CT. Because of less partial volume effects, a higher image noise can be tolerated with thin sections. However, reconstructing thicker sections, both axially as well as on MPR may avoid this increase in dose requirements.

While, axial images can be directly reconstructed from the raw data, optimum quality MPR requires reconstruction of a 'secondary raw data set' first. Such a data set can be used to reconstruct coronal, sagittal sections or reformations of arbitrary orientation through the data volume. Again, by increasing the thickness of these reformations, image quality can be increased (Fig. 4). With this technique, dose requirements with multislice CT can remain in the same order of magnitude as in spiral CT with the same section thickness.

Low dose and ultra-low dose protocols also are available, especially for high-contrast structures as the lungs or the skeleton. In situations in which radiation attenuation of the examined body cross-section is moderate (children, slim individuals, chest, neck, extremities), a low-kVp technique increases the CT attenuation of bone and contrast material and thus may allow for dose reduction without impairing signal-to-noise. Good results for CTA of the pulmonary vasculature at 80–90 kVp, for example, requires as little as 1 mSv effective dose (Fig. 5), and ultra-low-dose applications can go down to 0.4 mSv, a dose that is

Fig. 5. Low-dose scanning with multislice CT is possible. This CTA of chronic thrombembolic pulmonary hypertension was obtained with approximately 1 mSv effective dose. Acquisition parameters: 4 × 1/6; reconstruction parameters: 1.25/0.7 for both examinations.

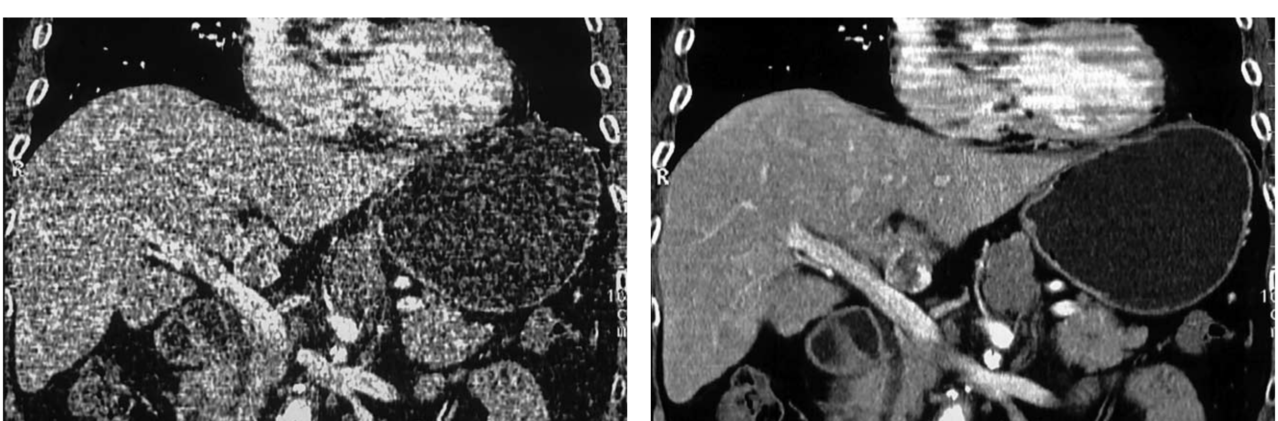

Fig. 4. Reconstructing thick MPR substantially increases image quality without requiring an increase in patient exposure. As compared with a 1-pixel thick MPR (a), the 5 mm-thick MPR (b) suffers much less from image noise. Acquisition parameters: 4 × 1/5; reconstruction parameters: 1.25/0.7.

Fig. 6. Volume rendered image of a patient with liver cirrhosis and two small foci of HCC.

comparable to an exam consisting of a conventional p.a. and lateral chest radiograph using a 100 speed screen-film system.

8. Image display and evaluation

Like in spiral CT, image review is mainly based on interactive viewing of (thick) axial sections in a cine display. In addition or even as a substitute, MPR (e.g. in a coronal plane) may be reviewed. Reconstruction of these MPR may be performed by the technologist so that images are readily available for interpretation, or it may be performed interactively by the radiologists as a problem-solving tool in cases that remain unclear on axial sections.

Various 3D rendering techniques provide excellent image quality, especially for CTA and bone work (Fig. 5). Recently volume rendering has become increasingly interesting also for standard applications (Fig. 6) and may be used as a primary mode for image review of near-isotropic multislice CT data sets [4].

9. Summary

With multislice CT, the performance of CT scanning has improved substantially (up to 25 times with current 16-slice units). The collimation of a single active detector row (section collimation SC) and the table feed per rotation (TF) are the most important acquisition parameters. The pitch may be used as the second acquisition parameter instead of TF but two definitions of pitch are used that may contribute to misunderstandings. Multiple data sets with varying section width SW can be reconstructed from the same raw data set. With 16-slice units there is a trend towards first reconstructing an overlapping 'secondary raw data set' and then performing multiplanar reformations of axial, coronal or arbitrarily angulated sections with an predefined section width SW. This thick-MPR technique is very effective for reducing image noise so that radiation exposure can be kept low. Low-kVp techniques improve signal-to-noise for contrast-enhanced studies and can be employed to further reduce radiation exposure. Thick MPR will become the primary mode for image review but 3D volume rendering can be expected to become a potent alternative, particularity for visualization of vascular structures. In conclusion, multislice scanning has transformed CT into a fully three-dimensional imaging technique.

References

[1] Klingenbeck-Regn K, Schaller S, Flohr T, Ohnesorge B, Kop AF, Baum U. Subsecond multi-slice computed tomography: basics and applications. Eur J Radiol 1999;31:110–24.

[2] Mahesh M. The AAPM/RSNA physics tutorial for residents: search for isotropic resolution on CT from conventional through multiple-row detector. Radiographics 2002;22:949–62.

[3] Jhaveri KS, Saini S, Levine LA, Piazzo DJ, Doncaster RJ, Halpern EF, Jordan PF. Thrall JH. Effect of multislice CT technology on scanner productivity. AJR 2001;177:769–72.

[4] Rubin GD. Data explosion: the challenge of multidetector row CT. Eur J Radiol 2000;36:74–80.

[5] Flohr T, Stierstorfer K, Bruder H, Simon J, Schaller S. New technical developments in multislice CT, part 1: approaching isotropic resolution with sub-millimeter 16-slice scanning. RöFo Fortschr Röntgenstr 2002;174:839–45.

[6] Prokop M. Radiation exposure and image quality. In; Prokop M, Galanski M (eds.) Spiral and multislice computed tomography of the body. Thieme; Stuttgart, New York 2003:131–160.

[7] Hu H, He HD, Foley WD, Fox SH. Four multidetector-row helical CT: image quality and volume coverage speed. Radiology 2000;215:55–62.

[8] Prokop M. Optimization of scanning technique. In; Prokop M, Galanski M (eds.) Spiral and multislice computed tomography of the body. Thieme; Stuttgart, New York 2003:109–130

[9] Hsieh J. Investigation of the slice-sensitivity profile for step-and-shoot mode multi-slice computed tomography. Med Phys 2001;28:491–7.

[10] Fuchs T, Krause J, Schaller S, Flohr T, Kalender WA. Spiral interpolation algorithms for multislice spiral CT—part II: measurement and evaluation of slice sensitivity profiles and noise at a clinical multislice system. IEEE Trans Med Imaging 2000;19:835–47.

[11] Wang G, Vannier MW. The effect of pitch in multislice spiral/helical CT. Med Phys 1999;26:2648–53.

[12] Taguchi K, Aradate H. Algorithm for image reconstruction in multi-slice helical CT. Med Phys 1998;25:550–3.

[13] Kachelreiss M, Schaller S, Kalender WA. Advanced single-slice rebinning in cone-beam spiral CT. Med Phys 2000;27:572–4.

[14] Kachelriess M, Watzke O, Kalender WA. Generalized multi-dimensional adaptive filtering for conventional and spiral single-slice, multi-slice, and cone-beam CT. Med Phys 2001;28:475–90.

[15] Proksa R, Kohler T, Grass M, Timmer J. The n-PI-method for helical cone-beam CT. IEEE Trans Med Imaging 2000;19:848–63.

[16] Schaller S, Stierstorfer K, Bruder H, et al. Novel approximate approach for high quality image reconstruction in helical cone-beam CT at arbitrary pitch. Proc SPIE 2001;4322:113–27.
[17] Taguchi K, Zeng GL, Gullberg GT. Cone-beam image reconstruction using spherical harmonics. Phys Med Biol 2001;46:N127–38.
[18] McCollough CH, Zink FE. Performance evaluation of a multi-slice CT system. Med Phys 1999;26:2223–30.