Multislice CT Technology

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Introduction

'Multislice CT' (MSCT) denotes the ability of a CT scanner to acquire more than one slice simultaneously. To be capable of doing so, the detector system must necessarily be composed of more than a single row of detector elements. Other terms often used, such as 'Multidetector CT', are somewhat misleading, as the number of detector rows is generally larger than the number N of slices. The latter, however, is the decisive feature of such a scanner.

The multislice CT era started already in 1992 with the introduction of the Elscint CT Twin, a dual slice scanner. The advantages of a MSCT scanner can shortly be characterized by the acronyms R, S, V, P, which stand for

- Resolution: improved spatial resolution along the z-axis
- Speed: reduced time for scanning a given body region
- Volume: increased length that can be scanned for a given set of scan parameters
- Power: improved usage of x-ray tube power

In 1998, the first four-slice scanners were presented, followed by the introduction of 16-slice scanners in 2001. The rapid development in this field is expressed by the presentation of 32- and 40-slice scanners and the announcement of 64-slice scanners at the 2003 RSNA meeting. Not only the number of slices has increased, but also the rotational speed, from formerly 1 s to presently 0.375 s per rotation.

Whereas dual-slice scanners allowed improving one specific aspect only (R or S or V or P), scanners with 16 and more slices are virtually unlimited (R and S and V and P). This has opened the field for new or improved applications, such as cardiac CT, CT angiography, CT perfusion, polytrauma CT, and orthopedics, to only name the most important.

Detector Layout and Slice Definition

The essential precondition for multislice CT is a multirow detector array. Both gas detectors and 4th generation scanners with 360° detector rings are no longer compatible with MSCT requirements. Consequently, all MSCT scanners are of the 3rd generation rotate-rotate type and employ solid-state detectors.

With four-slice scanners, various designs were used which differed in the number of detector rows, the dimensions of the detector elements and the total width of the array (Fig. 1 on the left). The universal matrix design used by GE allowed using the same detector for an eight-slice scanner also, which was introduced in 2001, at the expense of a large number of septa which are not contributing to detection. The progressive design commonly employed by Philips and Siemens aimed to reduce the number of septa between the rows, thereby improving the geometric efficiency of the array. The hybrid design introduced by Toshiba was the only one that offered four slices in sub-millimeter acquisition mode, however at the expense of an even greater number of septa. In addition, the Toshiba layout provided a total width of 32 mm for scanning four slices each 8 mm thick simultaneously.

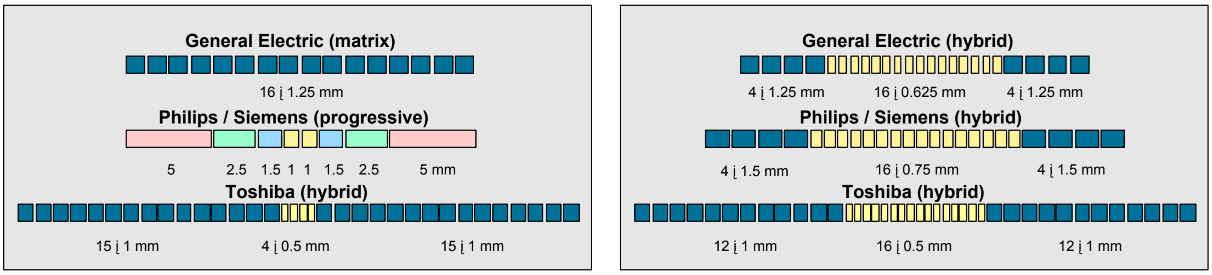

Figure 1: Detector array design of MSCT scanners. The layout of four-slice scanners (left) differs significantly from manufacturer to manufacturer, offering specific advantages and drawbacks. With 16-slice scanners, all manufacturers have made use of a hybrid arrangement.

With 16-slice scanners, all manufacturers employed a hybrid layout, allowing for submillimeter acquisition in 16-slice mode (Fig. 1 on the right). Only the size of the smallest detector elements and the total width of the array differ, with each manufacturer claiming to offer the most optimal design. However, the question what is optimal depends on all aspects involved (z-resolution, volume coverage, dose), not only one (e.g. z-resolution). As in daily life, the optimum is the result of the best compromise. This becomes evident in cardiac CT, which is the most demanding new MSCT application.

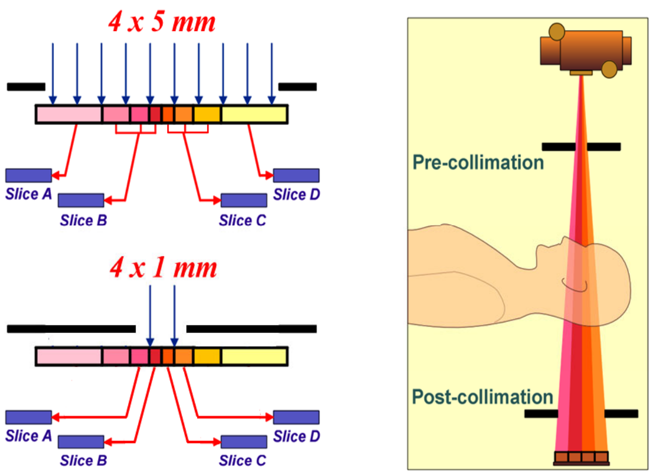

Figure 2: Slice definition is achieved by electronically combining adjacent detector rows and employing appropriate pre- and post-patient collimation.

Slice definition is achieved by combining adjacent detector rows and employing appropriate pre- and post-patient-collimation (Fig. 2). Thus a variety of slice collimations can be offered, such as 4 · 5 mm, 4 · 2.5 mm, 4 · 1 mm and 2 · 0.5 mm for the progressive design shown here. Similar considerations apply to the other designs. It is important to note that the slice thickness used for image presentation can differ from that during data acquisition (slice collimation). Thicker slices can be generated from thin slice data, either during reconstruction or by post-processing. However, once a certain slice collimation has been selected, it is not possible to reconstruct thinner slices later.

Hybrid designs are also used for the most recently presented scanners with 32 and more slices. However, when going to detector arrays with more than approximately 45 rows, a technological barrier appears (Fig. 3). This is caused by the minimal spacing of the wirebonds linked to the data readout lines, which cannot be made smaller than 60 µm. In traditional, front-illuminated photodiode design (Fig. 3, bottom left), these lines must be arranged in a horizontal fashion (Fig. 3, top), thus limiting the number of detector rows which can electrically be connected to the ASIC of the data acquisition system. This bottleneck has recently been overcome with the advent of back-illuminated photodiodes (Fig. 3, bottom right). As these can vertically be guided to the ASIC via a conductive epoxy, the number of detector rows is no longer limited by spatial restrictions.

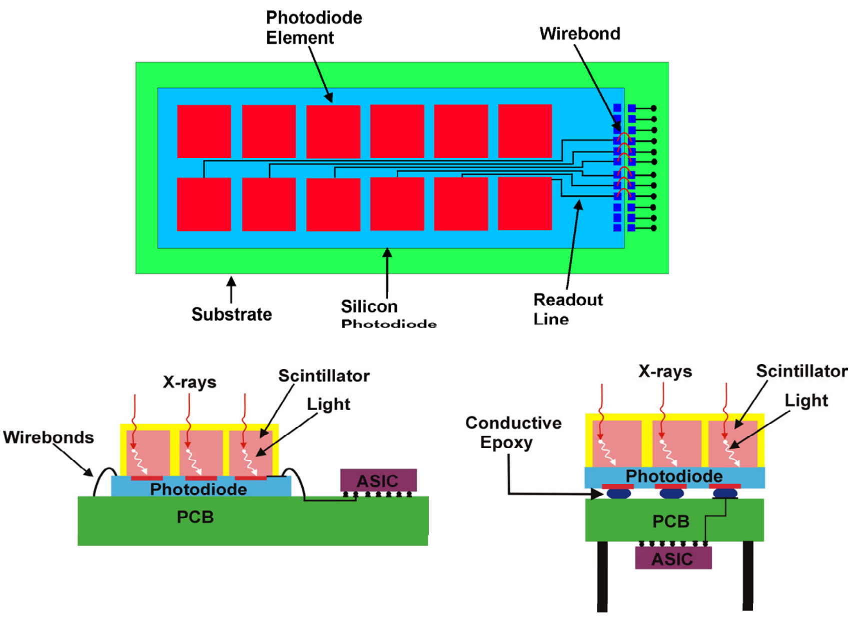

Figure3: The number of rows in a MSCT detector array is restricted in traditional technology with front-illuminated photodiodes, as wirebond spacing (at least 60 µm) is the limiting factor for horizontal read-out. With back-illuminated photodiodes, which can vertically be guided to the ASIC chip, this barrier has been overcome.

z-Interpolation, Pitch and mAs per Slice Conception

MSCT can be used both in sequential and spiral scanning modes. As with singleslice CT (SSCT), data acquired in spiral mode have to be interpolated in order to achieve axial slices. However, a different interpolation scheme is used in most MSCT implementations: While a two-point interpolation between a pair of data point closest to the reconstructed slice position is employed in SSCT, MSCT scanners from Philips, Siemens and Toshiba make use of a multi-point interpolation (z-filtering, Fig. 4). All data points falling inside a pre-selected filter width FW (which defines the reconstructed slice thickness) are taken into account, either equally or in a weighted fashion.

This new interpolation scheme offers significant advantages: other than for singleslice scanners, slice profile width (effective slice thickness) can be kept constant independent of the pitch factor selected (Fig. 5). However, also different from SSCT, image noise now changes with pitch, as the number of data points available for interpolation also changes (see Fig. 4). To avoid this, the electric tube current is automatically adapted proportional to the increase (or decrease) in pitch factor settings (‘effective mAs’ or ‘mAs per slice’ conception).

As a consequence, slice thickness, image noise and average patient dose are independent of the pitch factor settings for a pre-selected, constant value of mAs per slice (i.e. electric mAs divided by pitch). So pitch merely serves to control the scan speed. This holds for MSCT scanners of Philips and Siemens, whereas Toshiba users need to adapt the mAs settings manually if felt necessary. By using pitch factors < 1, a greater data density is achieved which can be used to virtually increase the available mAs per slice value despite the limited load-ability of the x-ray tube. In addition, spiral artifacts are reduced for pitch factor settings < 1, both at the expense of reduced volume coverage per unit time, however. Some manufacturers like GE and Toshiba also offer (or recommend) dedicated pitch factor settings only ('pre-ferred pitches') in order to optimize data sampling and to minimize artifacts, while Philips and Siemens users are not restricted in this context.

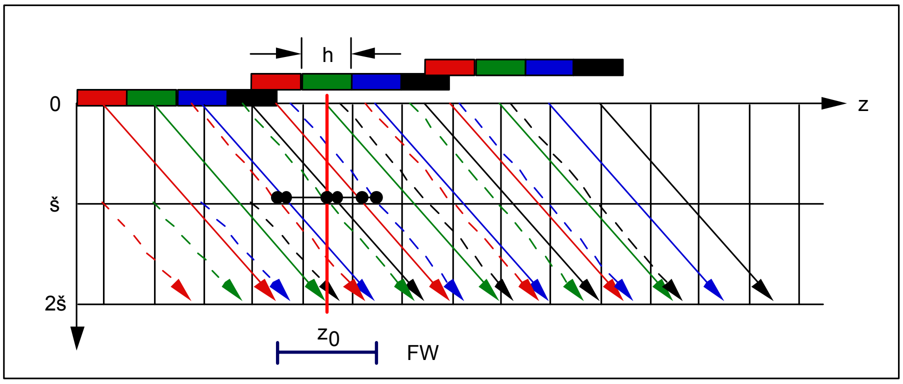

Figure 4: Multi-point interpolation scheme (z-filtering) for a four-slice scanner where all data points falling inside a pre-selected filter width FW are used. In this example, FW is twice the slice collimation h.

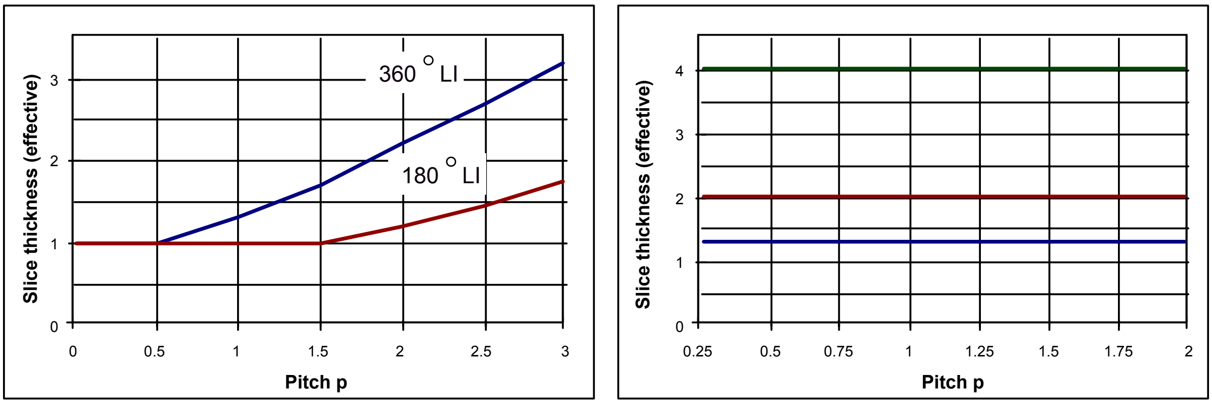

Figure 5: In singleslice CT, the effective slice thickness inevitably becomes wider for increased pitch settings, depending on the interpolation algorithm used (on the left). MSCT scanners which employ z-filtering are able to keep slice profile width constant independent of pitch factor settings (on the right).

The pitch factor used here follows the universal pitch definition given in IEC standards as the ratio of the table feed per rotation and the total collimation $N \cdot h$, where pitch 1 denotes a 'slice-by-slice' acquisition. Most manufacturers now comply with this definition instead of another one formerly used ('volume pitch' p^* = table feed divided by slice collimation h only), resulting in large values (1 to 8 for four-slice, 2 to 30 for 16-slice scanners). This was highly misleading, as a scanner can be used in different 'number of slices' configurations (e.g. 16-, 6-, 4-, and 2-slice modes with some 16-slice scanners).

Conebeam Issues

With an increasing number of slices, the cone angle, which characterizes the divergence of the radiation beam along the z-axis, becomes larger. This causes a misregistration of diagnostically relevant details, as is schematically demonstrated in figure 6. For up to six slices this effect can almost be neglected. Consequently, conventional 2-D fanbeam reconstruction schemes, which assume that each slice was irradiated by a separate radiation source opposite to it, can still be used. Beyond six slices, however, cone beam artifacts become increasingly visible.

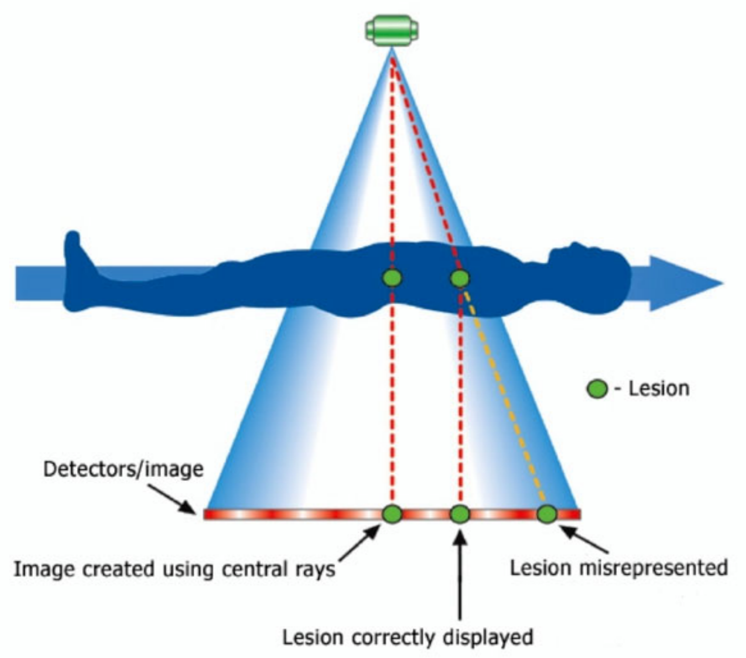

Figure 6: Misregistration of diagnostically relevant details apart from the central ray, which is caused by the enlarged cone angle.

These are most pronounced for details which are located off-axis and most distant from the central rows of the detector, as demonstrated in figure 7. To avoid this, advanced algorithms are mandatory. Philips and Toshiba employ true 3-D conebeam algorithms (Philips: Cobra, Toshiba: TCOT). These treat the data on a voxel-by-voxel base, thus ensuring that all details are correctly displayed on the final image. The two other manufacturers have further refined their 2-D fanbeam reconstruction algorithms (GE: MDMP with CrossBeam & Hyperplane, Siemens: AMPR). These are of the 'nutating slices' type, which reconstruct oblique planes individually adapted to the spiral trajectory in a first step. These are then interpolated to axial slices in a second step.

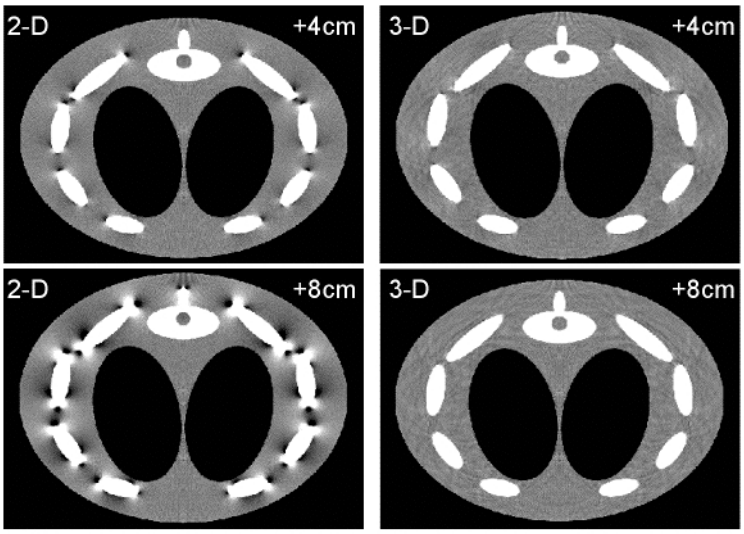

Figure 7: Cross-sectional images of a chest phantom, acquired with a 16-slice scanner (16 · 1.5 mm) and reconstructed with 2-D fanbeam (left) and 3-D conebeam algorithms. Relative slice position is located 4 cm (top) and 8 cm (bottom) from the centre of the detector array.

True 3-D conebeam algorithms, however, require considerably increased computational power compared to 2-D fanbeam algorithms (by more than one order of magnitude). Nevertheless, with dedicated reconstruction hardware, up to 40 images per second with full conebeam correction are state-of-the-art at present. While true 3-D conebeam algorithms are capable to cope with even more slices, 2-D fanbeam approximations will sooner or later hit onto their limits.

Other Technological Challenges

Finally, two other challenges need to be mentioned. Firstly, the dramatic increase in data due to the increased number of slices and reduced rotation speed. Data rates of more than 1000 Mbits/s must be handled in A/D conversion and data transfer from the gantry to the console. Dedicated ASICs (application-specific integrated circuits), optical slip-rings and high-speed data buses are required and have become available to solve this issue.

Secondly, the dramatic increase in the number of images (up to 1000 per study and sometimes even more). With the advent of 16-slice scanners at the latest, CT has finally become a volumetric data acquisition modality that no longer can be used successfully based on axial slices alone. Advanced viewing tools have become available, allowing to easily switch between different MPR orientations and 3-D representations and to modify slice thickness in real-time. While some manufacturers put emphasis on a 'secondary raw data set' of overlapping thin slices, which serve as a data stack for arbitrary secondary reformatting, others recommend to directly reconstruct MPR and 3-D representations from the raw data on each occasion. The future will show which conception ensures the most convenient workflow.

Summary and Outlook

From its beginning in 1992, multislice CT has experienced a rapid development. Multi-row detectors, ultra-fast data acquisition systems, rapid reconstruction hardware, dedicated conebeam reconstruction algorithms, and advanced viewing tools are the key factors to fully exploit the clinical benefits of this technology. The future challenge will be to make conebeam CT with large area detectors available for clinical routine, which would allow to cover entire organs (e.g. heart, brain) in one single axial scan. This technology is currently under development, and first prototypes have already shown excellent results for small high-contrast objects (joints, inner ear, contrast-filled vessel specimens). Yet none of the commercially available solutions, which are based on flat-panel CsJ-aSi technology, so far fulfills the high requirements of CT imaging in the medical environment. Although conebeam CT is a promising technical concept for medical imaging, it will not be available in the near future due to present technical limitations.

Suggested Reading

1. Kalender WA (2000) Computed tomography – fundamentals, system technology, image quality, applications. Viley-VCH Verlag, Weinheim
2. Prokop M, Galanski M (2003) Spiral and multislice computed tomography of the body. Thieme, Stuttgart New York, pp 1-82
3. Flohr T, Ohnesorge B, Schaller S (2004) Design, technique, and future perspective of multislice CT scanners. In: Reiser MF, Taskahashi M, Modic M, Becker CR (eds) Multislice CT 2nd revised edition. Springer, Berlin Heidelberg New York, pp 3-16