CHAPTER 2

REVIEW OF RELATED LITERATURES

2.1 Theory

2.1.1 Multislice Computed Tomography (MSCT)

Computed Tomography (CT) was introduced in the early 1970s and has revolutionized not only diagnostic radiology but also the whole field of medicine. The introduction of spiral CT in the early 1990s constituted a fundamental evolutionary step in the development and ongoing refinement of CT imaging techniques .Until then, the examination volume had to be covered by subsequent axial scans in a "step-and-shoot" mode. Axial scanning required long examination times because of the interscan delays necessary to move the table incrementally from one scan position to the next, and it was prone to misregistration of anatomic details (e.g., pulmonary nodules) because of the potential movement of relevant anatomic structures between two scans (e.g., by patient motion, breathing, or swallowing).

Spiral CT, the patient table is continuously translated while scan data are acquired. A prerequisite for spiral scanning was the introduction of slip-ring gantries, which eliminated the need to rewind the gantry after each rotation and enabled continuous data acquisition during multiple rotations. For the first time, volume data could be acquired without the danger of misregistration or double registration of anatomic details. Images could be reconstructed at any position along the patient axis (longitudinal axis), and overlapping image reconstruction could be used to improve longitudinal resolution. Volume data became the very basis for applications, such as CT angiography (CTA), which has revolutionized noninvasive assessment of vascular disease. The ability to acquire volume data also paved the way for the development of three-dimensional image processing techniques, such as multi-planar reformations (MPR), maximum intensity projections(MIP), surface shaded displays(SSD), or volume-rendering techniques(VRT), which have become a vital component of medical imaging today.

Ideally, volume data are of high spatial resolution and isotropic in nature (i.e., each image data element [voxel] is of equal dimensions in all three spatial axes), as a basis for image display in arbitrarily oriented imaging planes. For most clinical scenarios, however, single-slice spiral CT with 1 s gantry rotation time is unable to fulfill these prerequisites. To avoid motion artifacts and to use the contrast bolus optimally, spiral CT body examinations need to be completed within a certain time frame, ordinarily one patient breathhold (25–30 s). If a large scan range, such as the entire thorax or abdomen (30 cm), has to be covered within a single breathhold, a thick collimation of 5 to 8 mm must be used. Although the in-plane resolution of a CT image depends on the system geometry and on the reconstruction kernel selected by the user, the longitudinal (z-axis) resolution is determined by the collimated slice width and the spiral interpolation algorithm. Using a thick collimation of 5 to 8 mm results in a considerable mismatch between the longitudinal resolution and the in-plane resolution (ordinarily 0.5–0.7 mm depending on the reconstruction kernel). With single-slice spiral CT, the ideal of isotropic resolution can only be achieved for very limited scan ranges.

Strategies to achieve more substantial volume coverage with improved longitudinal resolution include the simultaneous acquisition of more than one slice at a time and a reduction of the gantry rotation time. The ability at that time to decrease rotation times substantially was limited by mechanical forces on the rotating part of the gantry and also by the need to increase X-ray flux accordingly. Because most of the flux the X-ray tube is producing was blocked by the collimation of the X-ray tube window, this flux could be used by multiple detector rows with no additional energy cost. Interestingly, the very first medical CT scanners were two-slice systems, such as the EMI head scanner (EMI, London, UK) introduced in 1972 or the Siemens SIRETOM introduced in 1974 (Siemens Medical Solutions, Erlangen, Germany). With the advent of whole-body fan beam CT systems for general radiology, two-slice acquisition was no longer used. Apart from a dedicated two-slice system for cardiac applications, the IMATRON C-100 (General Electric Healthcare, Waukesha, WI), introduced in 1984, the first step toward multislice acquisition in general radiology was a two-slice CT scanner introduced in 1993 (Elscint TWIN; Philips Medical Systems, Best, The Netherlands). In 1998, all major CT manufacturers introduced multislice computed tomography(MSCT) systems, which typically offered simultaneous acquisition of four slices at a rotation time of 0.5s, providing considerable improvement of scan speed and longitudinal resolution and better use of the available X-ray power. These developments were quickly recognized as revolutionary improvements that would eventually enable users to do real isotropic three-dimensional imaging. Consequently, all user pushed toward more and more slices, effectively rendering the number of slices into the most important performance characteristic of a CT scanner.

Simultaneous acquisition of M slices results in an M-fold increase in speed if all other parameters, such as slice thickness, are unchanged. This increased performance of MSCT compared with single-slice CT allowed for the optimization of a variety of clinical protocols. The examination time for standard protocols could be significantly reduced, which proved to be of immediate clinical benefit for the quick and comprehensive assessment of trauma victims and uncooperative patients, and for breathhold imaging. Alternatively, the scan range that could be covered within a certain scan time (e.g., breathhold time) was extended by a factor of M, which is relevant for oncologic staging or for CT angiography with extended coverage, for example of the lower extremities or the coverage of the complete lung. The most important clinical benefit, however, proved to be the ability to scan a given anatomic volume within a given scan time with substantially reduced slice width, at M times increased longitudinal resolution. This way, for many clinical applications the goal of isotropic resolution was within reach with four-slice CT systems. Examinations of the entire thorax or abdomen could now routinely be performed with 1mm or 1.25mm collimated slice width. MSCT also dramatically expanded into areas previously considered beyond the scope of third-generation CT scanners based on the mechanical rotation of X-ray tube and detector, such as cardiac imaging with the addition of ECG gating capability. With a gantry rotation time of 0.5s and dedicated image reconstruction approaches, the temporal resolution for the acquisition of an image was improved to 250 ms and less, which proved to be sufficient for motion-free imaging of the heart in the mid-to end-diastolic phase at slow to moderate heart rates. With four simultaneously acquired slices, coverage of the entire heart volume with thin slices and ECG-gating within a single breathhold became feasible, enabling noninvasive visualization of the coronary arteries .

Despite all these promising advances, clinical challenges and limitations remained for four-slice CT systems. True isotropic resolution for routine applications had not yet been achieved, because the longitudinal resolution of about 1 mm does not fully match the in-plane resolution of about 0.5 to 0.7 mm in a routine scan of the chest or abdomen. For large volumes, such as CTA of the lower extremity run-off, thicker (e.g., 2.5 mm) collimated slices had to be chosen to complete the scan within a reasonable time frame. Scan times were often too long to allow image acquisition during pure arterial phase. For a CTA of the circle of Willis, for instance, a scan range of about 100 mm must be covered. With four-slice CT, at 1 mm collimated slice width, pitch of 1.5, and 0.5s gantry rotation time, this volume can be covered in about 9s scan time, not fast enough to avoid venous overlay assuming a cerebral circulation time of less than 5s. For ECG gated coronary CTA, stents or severely calcified arteries constituted a diagnostic dilemma, mainly because of partial volume artifacts as a consequence of insufficient longitudinal resolution.

As a next step, the introduction of an eight-slice CT system in 2000 enabled shorter scan times, but did not yet provide improved longitudinal resolution (thinnest collimation 8 x 1.25 mm). The latter was achieved with the introduction of 16-slice CT, which made it possible routinely to acquire substantial anatomic volumes with isotropic submillimeter spatial resolution. Improved longitudinal resolution goes hand in hand with considerably reduced scan times that enable high-quality examinations in severely debilitated and severely dyspneic patients. For patients with suspicion of ischemic stroke, both the status of the vessels supplying the brain and the location of the intracranial occlusion can be assessed in the same examination. Additional brain perfusion CT permits differentiation of irreversibly damaged brain tissue from reversibly impaired tissue at risk. Examining the entire thorax (350mm) with submillimeter collimation requires a scan time of approximately 11 s. Because of the short breathhold time, central and peripheral pulmonary embolism can be reliably and accurately diagnosed. A 16-multislice CT enables whole body angiographic studies with submillimeter resolution in a single breathhold. Compared with invasive angiography, the same morphologic information is revealed. ECG-gated cardiac scanning with 16-multislice CT systems benefits from both improved temporal resolution achieved by gantry rotation times down to 0.375 s and improved spatial resolution.

The race for more slices is ongoing. In 2004, all major CT manufacturers introduced the next generation of MSCT systems with 32, 40, or even 64 simultaneously acquired slices, which brought about a further leap in volume coverage speed. Some of these scanners use refined z-sampling techniques enabled by a periodic motion of the focal spot in the z-direction (z-flying focal spot) to further enhance longitudinal resolution and image quality in clinical routine. With gantry rotation times down to 0.33 s, temporal resolution for ECG-gated examinations is again markedly improved.

2.1.2 Principles of a state-of-the-art MSCT system

The fundamental demands on a modern volume scanner can be summed up in the following two requirements: continuous data acquisition (the possibility to reconstruct images at any z-position); and ability to scan a long distance in short time without compromising longitudinal (z) resolution. The first requirement calls for a spiral acquisition. The breakthrough of spiral MSCT since 1998 is caused by the fact that it is able to fulfill the second requirement: the maximum z-resolution is defined by the detector size alone rather than by a combination of spiral pitch and detector collimation as in single-slice spiral. Moreover, multislice spiral scanners provide the new feature of allowing specification of the z-resolution in the reconstruction phase, after the scan has been done. The technical challenges of MSCT are manifold. A detector capable of measuring several thousand channels at a time has to be built; the data have to be transferred to the image reconstruction system; and a suitable reconstruction algorithm has to be devised.

The basic system components of a modern third-generation CT system are shown in (Figure2.1). Third-generation CT scanners use the so-called rotate–rotate geometry, in which both the X-ray tube and the detector rotate about the patient (Figure 2.2) In an MSCT system, the detector comprises several rows of 700 and more detector elements that cover a scan field of view of usually 50 cm. The X-ray attenuation of the object is measured by the individual detector elements. All measurement values acquired at the same angular position of the measurement system are called a projection of view. Typically, 1,000 projections are measured during each 360° rotation.

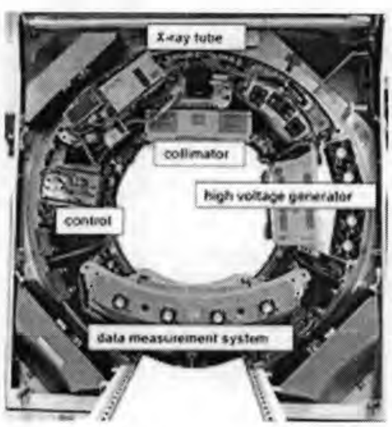

Figure 2.1 Basic system components of a modern third-generation CT system.

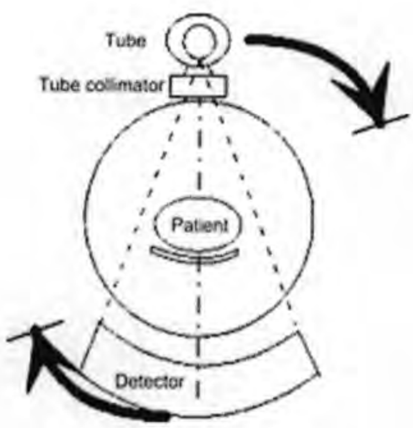

Figure 2.2 Measurement principle of a modern third-generation CT system.

The overall performance of a CT system depends on several key components. These components include the X-ray source, a high-powered generator, detector and detector electronics, data transmission systems (slip-rings), and the computer system for image reconstruction and manipulation.

State-of-the-art X-ray tube and generator combinations provide a peak power of 60 to 90 kW, usually at various, user-selectable voltages (e.g., 80, 100, 120, and 140 kV). Different clinical applications require different X-ray spectra and hence different kilovolt settings for optimum image quality or best possible signal-to-noise ratio at lowest dose. As an example, CT angiographic examinations generally benefit from lower tube voltage, resulting in a higher contrast resolution for CTA using iodine-based contrast material.

In a conventional tube design, an anode plate of typically 160-220mm diameter rotates in a vacuum housing. The heat storage capacity of anode plate and tube housing (measured in megaheat units) determines the performance level: the bigger the anode plate, the larger the heat storage capacity, and the more scan seconds can be delivered until the anode plate reaches its temperature limit. Typically, a conventional state-of-the-art X-ray tube has a heat storage capacity of 5 to 9 megaheat units, realized by thick graphite layers attached to the backside of the anode plate. The heat produced in the anode during X-ray emission is mainly dissipated by thermal radiation, and only a small percentage by thermal conduction. Constructive efforts aim at increasing both heat storage capacity and heat dissipation rate (e.g., by increasing the anode diameter, by using circular grooves in the anode support to increase the contact area for improved cooling, or by using special liquid-metal vacuum bearings that allow for a faster anode rotation). An alternative design is the so-called rotating envelope tube. The anode plate constitutes an outer wall of the rotating tube housing; it is in direct contact with the cooling oil and can be effectively cooled by thermal conduction. This way, a very high heat dissipation rate of 5 megaheat units/min is achieved, eliminating the need for heat storage in the anode, which consequently has a heat storage capacity close to zero. Because of the fast anode cooling, rotating envelope tubes can perform high-power scans in rapid succession. With the Straton tube, five full power 10s spirals are possible within 100s. Ultimately, the performance of both conventional and rotating envelope tubes is limited by the maximum heat dissipation of the CT system itself. Because there are no moving parts and no bearings in the vacuum, the tube design can be small and compact (anode diameter, 12 cm) and has the potential better to withstand the high gravitational forces associated with gantry rotation times of less than 0.4s. Because of the central rotating cathode, permanent electromagnetic deflection of the electron beam is needed to position and shape the focal spot on the anode. The versatile electromagnetic deflection is a prerequisite for the double z-sampling technology used in a recently introduced 64-multislice CT system.

With increasing number of detector rows and decreasing gantry rotation times, the data transmission systems of MSCT scanners must be capable of handling significant data rates: a 4-multislice CT system with 0.5s rotation time roughly generates 1,000 x 700 x 4 x 2 bytes = 5.6 megabytes (MB) of data per rotation, corresponding to 11.2 MB/s; a 16-multislice CT scanner with the same rotation time generates 45 MB/s; and a 64-multislice CT system can produce up to 180 - 200 MB/s. The acquired signal of every detector element is during this process digitized by analog-digital (A/D) converters in the submillisecond range with typically 16-bit resolution.

This stream of data is a challenge for data transmission off the gantry and for real-time data processing in the subsequent image reconstruction systems. In modern CT systems, contactless transmission technology is generally used for data transfer, which is either laser transmission or electromagnetic transmission with a coupling between a rotating transmission ring antenna and a stationary receiving antenna. In the image reconstruction computer images are reconstructed at a rate of up to 40 images/s using special array processors.

Modern CT systems generally use solid-state detectors. Each detector element consists of a radiation-sensitive solid-state material (e.g., cadmium tungstate or rare earth-based material such as gadolinium-oxide, gadolinium oxi-sulfide, or yttrium-gadolinium- oxide with suitable dopings), which converts the absorbed X-rays into visible light. The light is then detected by a Si-photodiode. The resulting electrical current is amplified and converted into a digital signal. Key requirements for a suitable detector material are good detection efficiency and very short afterglow time to enable the fast gantry rotation speeds that are essential for ECG-gated cardiac imaging. Gas detectors, such as the Xe-detectors used in previous generations of single-slice CT systems, have become obsolete because of their inherent limitations. To achieve acceptable detection efficiency, the Xe-chambers had to be designed deep enough to absorb and convert most of the X-ray quanta. This resulted in great challenges to manufacture such detectors, especially in combination with a multirow design.

A CT detector must provide different slice widths to adjust the optimum scan speed, longitudinal resolution, and image noise for each application. With a single-slice CT detector, different collimated slice widths are obtained by prepatient collimation of the X-ray beam. For $M > 2$, this simple design principle must be replaced by more flexible concepts requiring more than M detector rows simultaneously to acquire M slices. To be able to select different slice widths, several detector rows are electronically combined to a smaller number of slices according to the selected beam collimation and the desired slice width. For the four-slice CT systems introduced in 1998, two detector types have been commonly used. The fixed-array detector consists of detector elements with equal sizes in the longitudinal direction. A representative example for this scanner type, the GE Lightspeed scanner (General Electric Healthcare), has 16 detector rows, each of them defining 1.25mm collimated slice width in the center of rotation. The total coverage in the longitudinal direction is 20 mm at isocenter; because of geometric magnification the actual detector is about twice as wide. By prepatient collimation and combination of the signals of the individual detector rows, the following slice widths (measured at isocenter) are realized: 4 x 1.25 mm, 4 x 2.5 mm, 4 x 3.75 mm, 4 x 5 mm. The same detector design is used for the eight-slice version of this system, providing 8 x 1.25 and 8 x 2.5mm collimated slice width. A different approach uses an adaptive array detector design, which comprises detector rows with different sizes in the longitudinal direction. Scanners of this type, the Philips Mx8000 4-multislice scanner and the Siemens SOMATOM Sensation4 scanner (Siemens Medical Solutions), have eight detector rows. Their widths in the longitudinal direction range from 1-5 mm (at isocenter) and allow for the following collimated slice widths: 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.

The selection of the collimated slice width determines the intrinsic longitudinal resolution of a scan. In a step-and-shoot axial mode, any multiple of the collimated width of one detector slice can be obtained by adding the detector signals during image reconstruction. In a spiral mode, the effective slice width, which is usually defined as the full width at half maximum of the spiral slice sensitivity profile, is adjusted independently in the spiral interpolation process during image reconstruction. Hence, from the same dataset, both narrow slices for high resolution detail or for three dimensional post processing and wide slices for better contrast resolution or quick review and filming may be derived.

The established 16-multislice CT systems have adaptive array detectors in general. A representative example for this scanner type, the Siemens SOMATOM Sensation 16 scanner, uses 24 detector rows. The 16 central rows define 0.75mm collimated slice width at isocenter, the four outer rows on both sides define 1.5-mm collimated slice width. The total coverage in the longitudinal direction is 24 mm at the isocenter. By appropriate combination of the signals of the individual detector rows, either 12 or 16 slices with 0.75 or 1.5mm collimated slice width can be acquired simultaneously. The GE Lightspeed 16 scanner (General Electric Healthcare) uses a similar design, which provides 16 slices with either 0.625 or 1.25mm collimated slice width. The total coverage in the longitudinal direction is 20 mm at the isocenter. Yet another design, which is implemented in the Toshiba Aquilion scanner (Toshiba Medical Systems Corp., Tokyo, Japan), allows the use of 16 slices with 0.5, 1, or 2mm collimated slice width, with a total coverage of 32 mm at isocenter.

In 2004, the latest generation of MSCT systems providing more than 16 slices was introduced. The Siemens SOMATOM Sensation 64 scanner has an adaptive array detector with 40 detector rows. The 32 central rows define 0.6mm collimated slice width at the isocenter, the four outer rows on both sides define 1-mm collimated slice width. The total coverage in the longitudinal direction is 28.8 mm. Using a periodic motion of the focal spot in the z-direction (z-flying focal spot), two subsequent 32 slice readings with 0.6mm collimated slice width are slightly shifted in the z-direction and combined to one 64-multislice projection with a sampling distance of 0.3 mm at the isocenter. With this technique, 64 overlapping 0.6mm slices/rotation are acquired.

Alternatively, 24 slices with 1.2mm slice width can be obtained. The Philips Brilliance 40 scanner (Philips Medical Systems) provides 40 slices with 0.625mm collimation or 32 slices with 1.25mm collimation, with coverage of 40 mm at the isocenter. Both Toshiba and GE use fixed array detectors for their systems. The Toshiba Aquilion scanner has 64 detector rows with a collimated slice width of 0.5 mm. By appropriate combination of the signals of the individual detectors, 32 slices with either 0.5 or 1mm collimated slice width can be acquired simultaneously. The total z-coverage at the isocenter is 32 mm. The GE VCT scanner has 64 detector rows with a collimated slice width of 0.625 mm, enabling the simultaneous read out of 64 slices, with a total coverage of 40 mm in the longitudinal direction.

2.1.3 Dose and dose reduction

Factors affecting radiation dose

1. CT scanner design factors

The majority of scanner design features that affect dose and dose efficiency need not differ between single and multi-slice systems. Indeed, some manufacturers have a range of systems from single to 16-slice which are identical in terms of most of the features. The exception is that the single bank of detectors of a single slice scanner is replaced by multiple detector banks along the z-axis. It is this factor which primarily causes differences in dose efficiency between single and multi-slice scanners and will be discussed in a further section.

2. Scanner use

User selected scanning parameters, such as kV, mAs and scan length, will largely determine the radiation dose to the patient. Variation in the scan parameters used from site to site results in dose differences much greater than those due to scanner design factors. The increased capabilities of multi-slice scanners, which allow higher mAs values, longer scan lengths and multi-phase contrast studies, have the potential of directly increasing patient doses. Another indirect but significant effect on dose can result from the imaged slice width. Scanning is usually performed with narrower slices than on single slice scanners, so for the same noise, higher mAs values would need to be used. Furthermore, multi-slice scanners introduce new applications previously not possible in CT. The most notable of these is in the field of cardiology where high doses can result from the use of low pitch values in some applications.

3. Geometric efficiency

One of the main dosimetric differences between single and multi-slice scanners is in geometric efficiency. The geometric efficiency of an X-ray beam is the proportion of the total beam that is utilized in the imaging process. If the geometric efficiency decreases from 100% to 50% whilst all other factors are kept equal, then, to maintain the same image quality, the dose would need to be doubled. The overall geometric efficiency is sub-divided into two aspects. The first is the z-axis geometric efficiency, where the proportion of the overall X-ray beam width (dose profile) utilized along the long axis of the patient is considered. The second aspect, often overlooked, is the detector array geometric efficiency. This defines the proportion of the overall detector area that contains active detector material.

4. Z-axis geometric efficiency

Due to the finite size of the focal spot, an X-ray beam always has a reduced intensity at the periphery of the field and this region is referred to as the penumbra. On single slice systems, in most situations, the entire X-ray beam, including the penumbra, is utilized in image formation. If the penumbra were utilized in image production on multi-slice scanners, the outer detectors would receive a less intense X-ray beam than the inner ones. This would lead to the images from these detectors being narrower and noisier. To avoid this, the collimation of the X-ray beam on multi-slice systems is increased such that the penumbra lies beyond the active detectors and they are all irradiated uniformly .

In the latest IEC standard on CT safety, z-axis geometric efficiency is defined as the percentage of the X-ray beam width in the z-direction that is 'seen' by the detectors. On multi-slice scanners, the width seen by the detectors will be equal to the total active detector width. Where post-patient collimation is used to partially irradiate active detectors, the width 'seen' will be the nominal post-patient collimation width. The definition is given below as an equation :

$$\text{Z axis geometric efficiency} = \frac{\text{area under z-axis dose profile falling within active detector}}{\text{area under total z-axis dose profile}}$$

Scanners that acquire a greater number of simultaneous slices have an advantage in terms of z-axis geometric efficiency. This is because for narrow slice widths a wider total collimation can be used. An example for a 1.25 mm acquired slice width. On multi-slice scanners z-axis geometric efficiencies are generally in the range 80-98% for collimations of 10 mm and above, and about 55-75% for collimations of around 5mm. For collimations around 1-2mm z-axis geometric efficiencies are as low as 25% on some systems, although in dual slice mode they can be much higher. Therefore, the reduced z-axis geometric efficiency of multi-slice scanners, for the wider collimations most commonly used, leads to dose increases of around 10% when compared to single slice systems. However, very narrow collimations can result in a tripling, or more, in dose.

In helical scanning, additional information is required at each end of the planned image volume in order to provide interpolation data for the first and last images. On single slice scanners a half, or one, extra rotation is generally required at each end of the imaged volume. On multi-slice scanners the number of additional rotations required can potentially depend on a number of factors such as the interpolation method, the pitch and the reconstructed image width. Each additional rotation on a multi-slice scanner generally contributes a greater percentage to the dose than that on a single slice scanner because the total collimated beam width is usually greater.

When planning scan lengths to avoid sensitive organs such as eye lenses or gonads, the additional rotations associated with helical scanning should be borne in mind. The additional dose will be particularly significant for short scan lengths and in these situations it may be preferable to perform the scan in conventional slice by slice mode which does not require the extra rotations. As an example, for a collimated X-ray beam of 20 mm with one extra rotation at each end of a scan run, the increase in effective dose, relative to axial scanning, will be 10% for a scan length of 400 mm, and 40% for a scan length of 100 mm. This compares to increased dose figures of 2.5% and 10% respectively, for the same scan lengths on a single slice scanner, assuming half an extra rotation at each end of the scan .

With MSCT, a certain dose increase compared with single-slice CT is unavoidable because of the underlying physical principles. The collimated dose profile is a trapezoid in the transverse direction. This is a consequence of the finite length of the focal spot and the pre-patient collimation. In the plateau region of the trapezoid, X-rays emitted from the entire area of the focal spot illuminate the detector. In the penumbra regions, only a part of the focal spot illuminates the detector, whereas other parts are blocked off by the pre-patient collimator. With single-slice CT, the entire trapezoidal dose profile can contribute to the detector signal and the collimated slice width is determined as the full width at half maximum of this trapezoid. With MSCT, only the plateau region of the dose profile may be used to ensure equal signal level for all detector slices. The penumbra region has to be discarded, either by a post-patient collimator or by the intrinsic self-collimation of the multislice detector, and represents wasted dose. The relative contribution of the penumbra region increases with decreasing slice width, and it decreases with increasing number of simultaneously acquired slices. Which shows the minimum width dose profiles for a 4-multislice CT system and a corresponding 16-multislice CT system with equal collimated width of one detector slice. Correspondingly, the relative dose utilization of a representative 4-multislice CT scanner (SOMATOM Sensation 4) is 70% for 4 x 1mm collimation and 85% for 4 x 2.5mm collimation. A comparable 16-multislice CT system (SOMATOM Sensation 16) has an improved dose use of 76%, respectively, 82% for 16 x 0.75mm collimation, and 85%, respectively, 89% for 16 x 1.5mm collimation, depending on the size of the focal spot (large or small).

The most important potential for dose reduction is an adaptation of the dose to the patient size. As a rule of thumb, the dose necessary to maintain constant image noise has to be increased by a factor of 4 if the patient diameter is increased by 8 cm. Correspondingly, for patients 8 cm smaller than the average, a quarter of the standard dose is sufficient for adequate image quality, which is of tremendous importance in pediatric scanning. Most CT manufacturers offer dedicated pediatric protocols (e.g., with dose recommendations according to the weights of the children). In contrast-enhanced studies, such as CT angiographic examinations, the contrast-to-noise ratio for fixed patient dose increases with decreasing X-ray tube voltage. As a consequence, to obtain a given contrast-to-noise ratio, patient dose can be reduced by choosing lower kilovolt settings. Compared with a standard scan with 120 kV, the same contrast-to-noise ratio in a 32cm phantom, corresponding to an average adult, is obtained with 0.49 times the dose for 80 kV (1.3 times the milliamperes) and 0.69 times the dose (1.1 times the milliamperes) for 100 kV. Ideally, 80 kV should be used for CTAs at lowest patient dose. In practice, the maximum tube current available at 80 kV, which is generally not sufficient to scan obese patients, limits the application spectrum. In these patients, 100 kV is a good compromise and the preferable choice for CTA examinations, such as thoracic CTA or cardiac CTA. In anatomic dose modulation approaches, the tube output is adapted to the patient geometry during each rotation of the scanner to compensate for the varying X-ray attenuation in asymmetric body regions, such as shoulder and pelvis. Depending on the body region, dose can be reduced by 15 to 35% without degradation of image quality. In more elaborate approaches, the tube output is modified according to the patient geometry in the transverse direction to maintain adequate dose when moving to different body regions, (e.g., from thorax to abdomen), so-called automatic exposure control. In the special case of ECG gated spiral scanning for cardiac applications, the output of the X-ray tube can be modulated according to the patient's ECG. It is kept at its nominal value during a user-defined phase of the cardiac cycle, in general the mid- to end-diastolic phase. During the rest of the cardiac cycle, the tube output is reduced to 20% of its nominal value. Depending on the patient's heart rate, the dose can be reduced by 30 to 50% using ECG controlled dose modulation.

2.1.4 Future perspective of MSCT

Because of its ease of use and its common availability, general purpose CT will evolve into the most widely used diagnostic modality for routine examinations, especially in emergency situations or for oncologic staging. CT gives morphologic information only; in combination with other modalities, however, functional and metabolic information can be additionally obtained. This is an important step on the way toward a "one stop shop" examination; combined systems will gain increasing importance in the near future. The combination of state-of-the art MSCT with positron emission tomography scanners, for instance, opens a wide spectrum of applications, ranging from oncologic staging to comprehensive cardiac examinations.

For general purpose CT, there will be a moderate increase of the number of simultaneously acquired slices in the near future; the resulting clinical benefits, however, may not be substantial and have to be carefully considered in the light of the necessary technical efforts. Potential further improvement of the spatial resolution will have to be reserved to special applications because of the inevitable increase of dose that has to be applied for adequate signal-to-noise ratio. It will have to go along with the development of more powerful X-ray tubes and generators. Instead of a mere quantitative enhancement of scan parameters with doubtful clinical relevance, the introduction of area detectors large enough to cover entire organs, such as the heart, the kidneys, the brain, or a substantial part of a lung, in one axial scan (120mm scan range) could bring a new quality to medical CT. With these systems, dynamic volume scanning would become possible, and a whole spectrum of new applications, such as functional or volume perfusion studies, could arise. Area detector technology is currently under development, yet no commercially available solution so far fulfills the high requirements of medical CT concerning dynamic range of the acquisition system and fast data read-out. The combination of area detectors with sufficient quality with fast gantry rotation speeds is a promising technical concept for medical CT systems.

2.2 Related literatures

Many researchers performed the CT radiation dose for children. Chan et al [10] performed CT in children aged 1-12 years with several different milliampere second (mAs) values, by using observer-based subjective image assessment showed that a 40% reduction in milliampere per second could be used in pediatric cranial CT without affecting the diagnostic quality of the images, Cohnen et al [11] also studied CT dose in pediatric head CT and concluded that a 40% reduction was possible.

Boone JM et al [12] compared six size of phantom cylinders (10-32 cm in diameter), scanned by a clinical multi-detector row CT scanner, CT technique factor from 80 to 140 kVp and 10 to 300 mAs. As compared with that in a reference cylindrical adult abdomen of 28 cm in diameter, contrast-to-noise ratio (CNR) was maintained at a contrast level in pediatric patients of 15, 20 and 25 cm in diameter, when milliampere second values of 0.557, 0.196 and 0.054 of the adult mAs values were used. The corresponding dose were reduced to 0.642, 0.287 and 0.090 of the 28 cm diameter adult dose, respectively. CT techniques charts for examination of pediatric heads were measured of 15 and 13 cm diameter, respectively. They involve the use of mAs values of 0.527 and 0.366 of those used for examination of standard 17 cm diameter adult head. That means CT technique charts for pediatric abdomen and head examinations were produced on the basis of physically measure data.

Brody [13] has discussed on pediatric thoracic CT in general terms and has provided specific recommendations for milliampere second (mAs) reductions on the basis of patient mass. Ihab et al [14] determined whether a lower radiation dose technique can be used for Computed Tomography of the pediatric pelvis without significant loss of diagnostic image quality. CT is an extremely important imaging modality for infants and children. Despite this importance, and increasing use, the risks of radiation and a lack of attention to these risks are recently being addressed.

Moss and McLean [15] studied pediatric and adult computed tomography practice and patient dose in Australia by survey of 52 sites representing 53 CT scanners to reflect the current practice of CT scanning in pediatric patients. For any age group, there was a large spread of effective dose. There was a clear trend for centers that frequently carry out CT scans on pediatric patients to have the lowest radiation doses. Effective dose was closely related to mAs. Centers using lower mAs for younger patient, could reduce the patient dose. The results of the survey emphasize the need for continuing education and protocol review, particularly in pediatric CT examination, in a complex and fast changing environment.

Morgan [16] studied image quality improvement and dose reduction in CT pediatric imaging, for any given scan technique, patient dose depends upon the size of the patient and the attenuation. The greater the patient attenuation, the smaller the patient dose will be for given scan technique. Consequently, pediatric patients in particular will receive a higher dose than that displayed by $CTDI_{w}$.

Frush P. et al[17] performed CT in an extremely important imaging modality for infants and children. Despite this importance, and increasing use, the risks of radiation and a lack of attention to these risks are only recently being addressed. Pediatric health care providers are an essential element in the selection of CT evaluation of children. Because of this, it is necessary to understand the benefits and radiation risk of CT and work with radiologist to develop strategies that reduce exposure from CT in infant and children.

Starck et al [18] evaluated noise characteristic using phantoms and dosimetric measurements. The low radiation dose capabilities of CT system may be very different despite similar tube and detector designs. Dose reductions archieve in their study using patient-specific scan parameters illustrate the considerable differences in radiation dose required in CT imaging when the patient in size from slim to corpulent. The selection of scan parameters for each individual patient would allow a reduction in radiation dose in most CT applications. However, such a method of manual exposure control in CT imaging requires this facility to be incooperated into the scanning system.

Tsapaki V et al [19] measured radiation dose for computed tomography (CT) of head, chest, and abdomen and compared to the diagnostic reference levels, as part of the International Atomic Energy Agency (IAEA) Research coordination project. Mean $CTDI_{vol}$ and DLP for head were 39 mGy and 544 mGy.cm, for chest $CTDI_{vol}$ and DLP wre 9.3 and 384 mGy.cm and for abdomen $CTDI_{vol}$ and DLP were 10.4 and 549 mGy.cm. All the result were below the European diagnostic reference levels.(brin; CTDIvol = 50 mGy and DLP = 1050 mGy.cm). Estimated effective doses were 1.2, 5.9 and 8.2 mSv.