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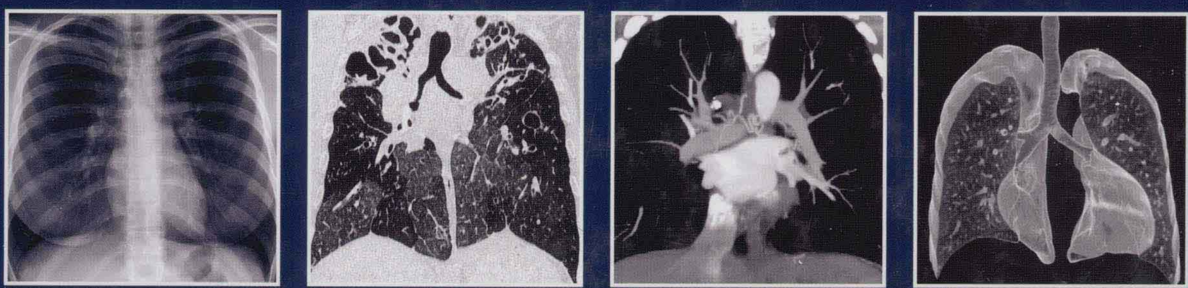
IMAGING OF DISEASES OF THE CHEST

FIFTH EDITION



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FIFTH EDITION

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PREFACE

This book has been written to provide radiologists, physicians, and thoracic surgeons with a one-volume account of chest imaging, primarily in the adult patient. An attempt has been made to present an integrated review of the appearances encountered in diseases of the lung, pleura and mediastinum using the various imaging techniques available in a modern imaging department.

From the preface to the first edition (1990)

We have again tried to meet the objectives set by our predecessors all of whom have retired since the first edition was published twenty years ago. Alex Bankier has kindly agreed to join us in our endeavors to bring this new edition up-to-date.

As with previous editions, the clinical and pathologic features of different diseases are provided in varying degrees of detail with more in-depth coverage given to rarer and less well understood conditions. There are new sections on emerging diseases, such as the surfactant deficiency disorders, but we are aware that current understanding of some of these conditions is incomplete and that inevitably their nomenclature and classification will change over the coming years.

References have been refreshed but the temptation to discard references simply because they are old has been resisted. Classic descriptions, notably elucidations of fundamental radiographic signs, have been retained. Apart from paying tribute to the legacy of earlier writers, many of these meticulous studies have not been bettered.

In line with current publishing aesthetics there are splashes of color that we hope will enliven the largely black and white contents. As an aide memoire, and to promote some degree of standardization in the descriptive language used in chest radiology, the latest version of the Fleischner Society's glossary of terms for thoracic imaging is reproduced in its entirety.

We hope that this new edition will be a useful resource and will provide more or less complete answers for anyone involved with thoracic imaging.

David M Hansell
David A Lynch
H Page McAdams
Alexander A Bankier
2010

ACKNOWLEDGEMENTS

Imaging of Diseases of the Chest is essentially the fruit of the labors of the current and previous authors but thanks are due to the contributors of subsections which have survived several editions.

Our admiration and gratitude is freely given to our dedicatee Peter Armstrong who was the inspiration and driving force behind the first and subsequent editions of *Imaging of Diseases of the Chest.* We have tried our best to emulate his lucid writing style.

As with previous editions our secretaries and helpers, Anne-Marie Henry, Tanya Mann, Nancy Williams, and Mary Anne Hansell have rendered superb assistance in many ways and in so doing have enabled us to meet what sometimes seemed to be impossible deadlines.

We would also like to thank innumerable colleagues and visitors to our departments who have, without knowing it, refined our thinking and improved the presentation of our material. We are grateful to Herb Kressel, Editor of *Radiology*, for allowing us to reproduce the Fleischner Society's glossary of terms for thoracic imaging.

Our publishers, notably Joanne Scott, Jess Thompson, and Michael Houston of Elsevier Mosby, have steered us through the production process with their usual calm efficiency and have given us encouragement when it was needed.

Our wives, Mary Anne, Anne, Emma and Francesca have once again shown great forbearance in allowing us to complete the task of updating *Imaging of Diseases of the Chest*.

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CONTENTS

CHAPTER

1 Technical considerations

The chest radiograph remains the prime imaging investigation in respiratory medicine and the basic technique has changed little over the past 100 years. Of all the cross-sectional imaging techniques, computed tomography (CT) has had the greatest impact on diagnosis of lung and mediastinal disease, while magnetic resonance imaging (MRI), ultrasonography, and positron emission tomography have complementary roles in specific clinical situations. Refinements to CT scanning protocols, notably since the widespread introduction of multidetector CT (MDCT), have led to a substantial increase in the total number of performed CT examinations. Subsequent increases in radiation burden delivered by diagnostic imaging have become a focus of public interest, and the ongoing refinement of means to reduce patient irradiation has become a priority.

CONVENTIONAL CHEST RADIOGRAPHY

Technique

The standard views of the chest are the erect posteroanterior (PA) and lateral projections. The PA chest radiograph is taken at near total lung capacity (inspiratory film) with the patient positioned such that the medial ends of the clavicle are equidistant from the spinous process of the thoracic vertebra at that level. The scapulae are held as far to the side of the chest as possible by rotating the patient's shoulders forward and placing the backs of the patient's wrists on the iliac crests. A chest radiograph obtained near residual volume (expiratory film) can substantially change the appearance of the mediastinal contour, as well as giving the misleading impression of diffuse lung disease (Fig. 1.1). Even on a correctly exposed film, just under half the area of the lungs is obscured by overlying structures.[1] Furthermore, many technical factors, notably the kilovoltage and film–screen combination used, determine how well lung detail is seen.

The steep S-shaped dose–response curve of conventional radiographic film–screen combinations makes it impossible to obtain perfect exposure of the most radiolucent and radiodense parts of the chest in a single radiograph. Methods of overcoming this shortcoming have included the use of high-kilovoltage (above 120 kV) techniques,[2] asymmetric screen–film combinations,[3] 'trough' or more complex filters,[4] and sophisticated scanning equalization radiographic units.[5]

High-kilovoltage radiographs have several advantages over low-kilovoltage films. Because the coefficients of X-ray absorption of bone and soft tissue approach each other at high kilovoltage, the bony structures no longer obscure the lungs to the same degree as on low-kilovoltage radiographs. Furthermore, the better penetration of the mediastinum with high-kilovoltage techniques allows greater detail of the large airways to be seen. At high kilovoltage, exposure times are shorter, so that structures within the lung tend to be sharper. Although scattered radiation is greater with high kilovoltage, the use of a grid causes a net reduction of image-degrading scattered radiation compared with a low-kilovoltage, nongrid technique. With a high-kilovoltage technique, an air gap of 15 cm in depth is often used, instead of a grid to disperse the scattered radiation; this is as effective as a grid, and the radiation dose to the patient is similar for the two techniques.[6] To counteract the unwanted magnification and penumbra effects of interposing an air gap, the focus–film (or anode-to-image) distance is increased to approximately 4 m. Although high-kilovoltage radiographs are preferable for routine examination of the lungs and mediastinum, low-kilovoltage radiographs provide excellent detail of unobscured

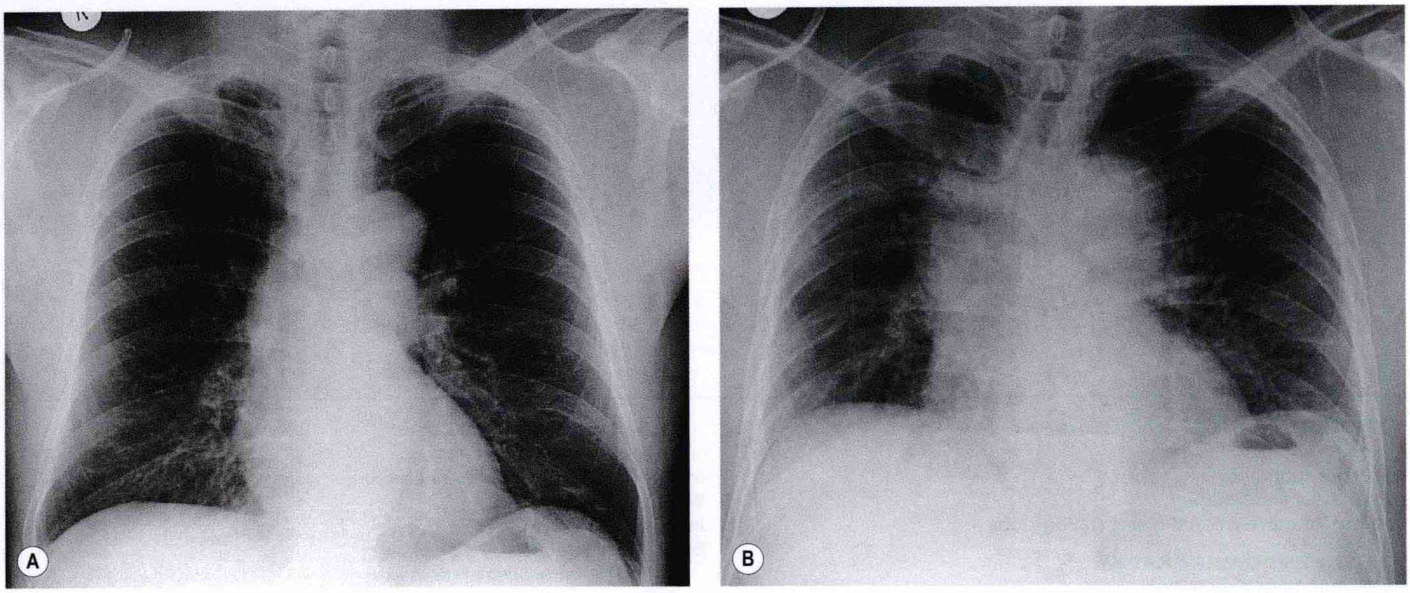


Fig. 1.1 **A**, Normal chest radiograph of an individual breathholding at full inspiration. **B**, By comparison, at full expiration the mediastinum appears abnormally widened and there is the appearance of a diffuse increase in lung density.

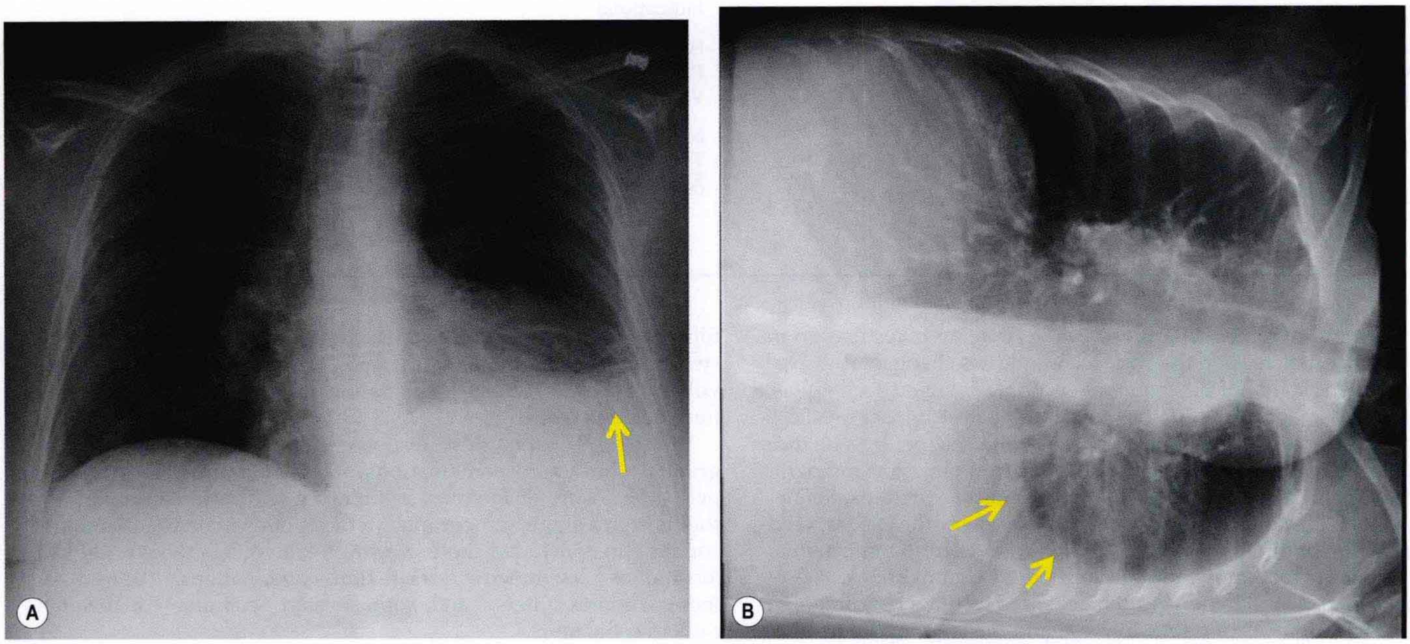


Fig. 1.2 Lateral decubitus view. **A**, Erect frontal chest radiograph shows blunting of the left costophrenic angle (arrow), potentially caused by pleural effusion. **B**, Lateral decubitus view confirms pleural effusion by showing that some of the pleural fluid gravitates along the left lateral chest wall (arrows).

lung because of the better contrast between lung vessels and surrounding aerated lung. Moreover, calcified lesions, such as pleural plaques, and small pulmonary nodules,[7] are particularly well demonstrated on low-kilovoltage films.

Extraradiographic views

The frontal and lateral projections suffice for most clinical indications. Other radiographic views are becoming much less frequently requested because of the ready availability of CT. Nevertheless, an additional view may occasionally solve a particular clinical problem quickly and in a cost-effective manner.

The *lateral decubitus view* is not, as its name implies, a lateral view. It is a frontal view taken with a horizontal beam, with the patient lying on his or her side. Its main purpose is to demonstrate the mobility of fluid in the pleural space. If a pleural effusion is not loculated, it gravitates to the dependent part of the pleural cavity (Fig. 1.2). If the patient lies on his or her side, the fluid layers between the chest wall and the lung edge. Because the ribs, unlike the diaphragm, are always identifiable, comparison of a standard frontal view with a lateral decubitus view is a reliable way of recognizing unloculated pleural fluid. A lateral shoot through radiograph may be used to advantage to show a small anterior pneumothorax in recumbent patients in intensive care.[8]

The *lordotic view* is now rarely used, but is included here for completeness. It is performed by angling the X-ray beam 15° cranial either by positioning the patient upright and angling the beam up or by leaving the beam horizontal and leaning the patient backward. The lung apices are thereby better penetrated, and are

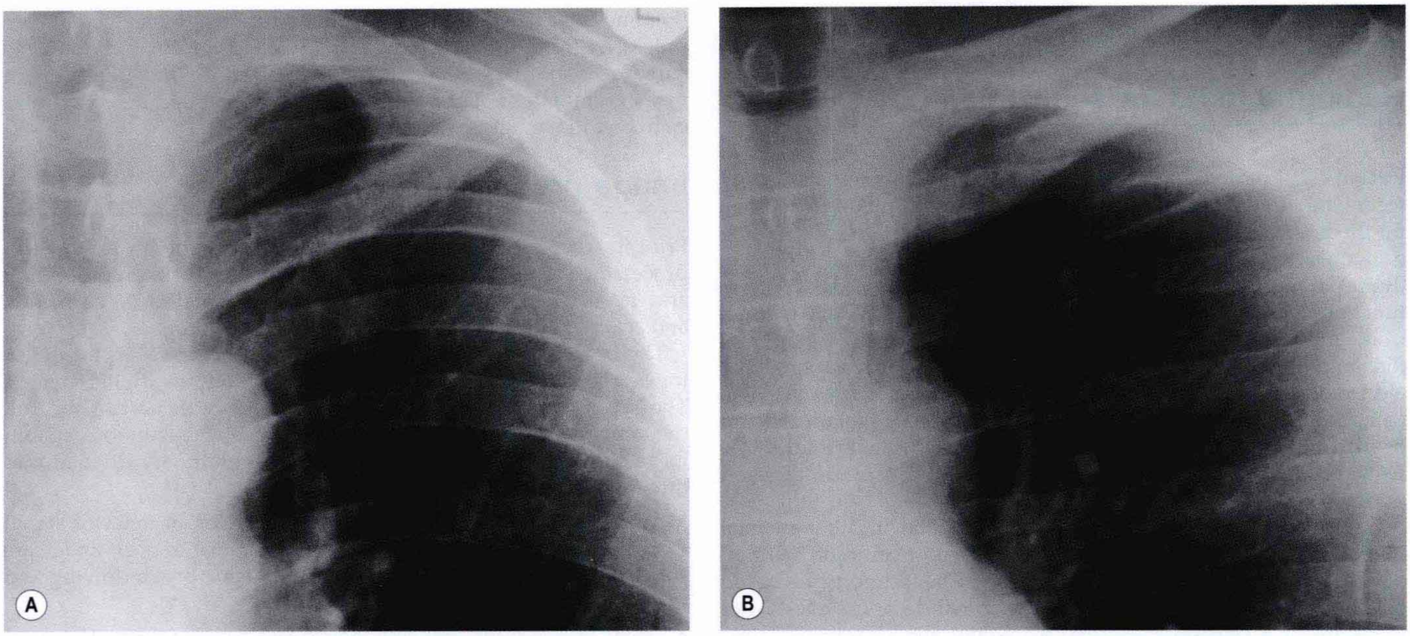


Fig. 1.3 Use of the lordotic view. **A**, Selective view of the apex shows a small opacity projected over the anterior end of the left first rib. **B**, Lordotic view confirms that the opacity is intrapulmonary, rather than part of the calcified costochondral cartilage.

free from the superimposed clavicle and first rib. The lordotic view may be useful for distinguishing a focal pulmonary opacity from incidental calcification of the costochondral junctions (Fig. 1.3). With the exception of identifying rib fractures and confirming the presence of a rib lesion, *oblique views* of the thorax are rarely required.

Portable chest radiography

Portable or mobile chest radiography has the obvious advantage that the examination can be performed without moving the patient to the radiology suite. In many centers, the proportion of portable to departmental chest radiographs has increased over time. However, portable radiography has a number of disadvantages.

The shorter focus–film distance results in undesirable magnification. High-kilovoltage techniques cannot be used because portable machines are unable to deliver the high kilovoltage and because accurately aligning the X-ray beam with a grid is difficult. Furthermore, the maximum milliamperage is severely limited, necessitating long exposure times with the risk of significant blurring of the image. Portable lateral radiographs with conventional film radiography are even less likely to be successful because of the long exposure times. Radiation exposure of nearby patients and staff is a further caveat.

Positioning of bed-bound patients is difficult, and the resulting radiographs often show half-upright or rotated subjects. Even in the so-called erect position with the patient sitting up, the chest is rarely as vertical as it is in a standing patient. More important, the patient is unable to take a deep breath when propped up in bed. Many patients cannot be moved to the radiology department and the improved quality of digital portable radiographs, notably flat panel detectors, represents a substantial improvement.

Limitations of conventional chest radiography

The chest constitutes a large part of the body and an image of the chest needs to encompass at least 40 cm. This large field-of-view imposes constraints on the image receptor, because the receptor must provide consistent and uniform response over the entire field. This field-of-view also increases the contribution of scattered radiation that can decrease image quality.[9]

The wide latitude of X-ray transmission through the thorax imposes a limit on the visualization of subtle abnormalities. For a typical X-ray beam used in chest radiography, regional variations in transmission through the thorax can range over two orders of magnitude.[9] Ideally, an imaging system should have enough latitude to capture and effectively display the diagnostically meaningful part of the X-ray transmission. Coverage of such wide latitude, however, can limit depiction of subtle low-contrast lesions. Maintaining wide latitude while preserving the visibility of low-contrast features is thus a particular challenge.[9,10]

The combination of high X-ray photon energy, a thick body part, and a large field-of-view results in a substantial amount of scattered radiation. This can account for 95% of the detected X-ray flux in the mediastinum and up to 70% in the lung in radiographs acquired without a grid.[11] Scattered radiation degrades contrast and increases image noise. The contribution of scattered radiation to image noise is not correctable.[9]

Conventional chest radiography involves the projection of a three-dimensional structure onto a two-dimensional image. Anatomic structures can therefore overlie each other, sometimes referred to as anatomic noise.[9] Anatomic noise can reduce the detectability of lesions. The projection of ribs is of particular concern for detection of lung nodules, because the ribs overlie about 75% of the area of the lungs. Moreover, a substantial portion of the lungs is projected over the heart and parts of the diaphragm.[9] The influence of anatomic noise on the detectability of lung nodules has been extensively studied several decades ago.[12,13] More recently, Samei et al.[14] demonstrated that anatomic noise is far more important than quantum noise in limiting the detectability of lung nodules.

Perceptual and cognitive processes are of particular importance in chest radiography because of the complexity of the tasks and the confounding effect of technical and anatomic parameters.[9] Perceptual errors can occur at both the visual and the cognitive level. Incompleteness of the search task may contribute to about 55% of missed lesions. These errors occur when the observer fails to look at the location of the lesion[12,15] or when he or she does not fix their eyes on this territory for a dwell time of at least 0.3 second.[16] Cognitive errors account for 45% of missed lesions and can occur when the fixation time on a potential abnormality exceeds the above limit but the reader fails to call the nodule pathologic.[12]

DIGITAL CHEST RADIOGRAPHY

Radiographic data acquisition

Digital chest radiography is expected to completely replace analog technology in the near future. There are several compelling reasons for the transition from analog to digital techniques. The decoupling of acquisition and display functions of the acquisition device in digital radiography makes it possible to optimize either of those functions independently. The availability of image data in electronic form makes it possible to post-process the image for optimal display and to display the images on viewing workstations. The electronic format also makes it possible to safely archive the data using less space and fewer resources for storage. Digital images can be distributed widely and copies can be made available to multiple viewers. Finally, acquisition and processing units can be integrated into one system.[9]

Computed radiography (CR) was one of the first commercial digital imaging techniques[17] and is still the most common technology for acquiring digital chest radiographs. The technology is based on photostimulable properties of barium halide phosphors. After exposure of a phosphor cassette to X-rays, the cassette is transported to a computed radiography reader device that scans the cassette with a laser beam. The laser releases the energy locally deposited by X-rays on the screen and causes the screen to fluoresce. The released light is used to form the image after it is collected by a light guide, digitized, and associated with the geometric location of the laser beam at the time of stimulation.[9] While CR has the largest number of installations in digital radiography, its disadvantages in terms of image quality per unit dose and suboptimal workflow have encouraged the development of flat-panel detector technology.

Flat-panel detectors are made of thin layers of amorphous silicon thin-film transistors (TFTs) deposited on a piece of glass. The TFT layer is coupled with an X-ray absorptive layer. Indirect flat-panel detectors use a phosphor screen to convert the X-rays to light photons, which are detected by the photodiode array associated with the TFT layer and converted to a charge deposited in the capacitors associated with each TFT.[18,19] Direct flat-panel detectors use a photoconductor layer that converts the X-ray energy directly to charge, which is subsequently directed to the collecting TFT-capacitor array through the application of a strong electric field.[20] After exposure, the charge on the capacitors is collected line by line and pixel by pixel by using the associated grid and data lines, thereby forming the raw digital image data for processing and display.[9]

Charge-coupled device (CCD) and complementary metal-oxide semiconductor (CMOS) cameras use an alternative technology for the acquisition of digital chest radiographs. With these detectors, the X-ray energy is first converted to light within a phosphor layer. The light is then directed to a single or a multitude of CCD or CMOS cameras that detect the light image and form the radiograph.[21] An important component of these detectors is the coupling of the phosphor layer and the camera. Because most CCD and CMOS sensors are limited in size, it is necessary to minify the original light image generated on the phosphor screen so that it can be captured by the camera. This is accomplished by using either a fiberoptic coupler or a lens system. In either case there is a loss of efficiency, since only a small fraction of the light photons generated by the phosphor are detected by the camera(s). Consequently, the inherent efficiency of these detectors is limited.[9]

A recent development takes advantage of slot-scan technology to reduce the amount of scattered radiation on digital chest radiographs.[22] The detector consists of a cesium iodide scintillation layer fiberoptically coupled to a series of linear CCDs. With no antiscatter grid in place, a narrow-fan X-ray beam synchronized with the movement of the detector assembly scans the chest. Image data are continuously read from the CCDs as the patient is scanned by using the time-integration method.[23,24] After scanning, the image data are processed for optimal display. The advantage of this technology is superior scatter rejection with little effect on the detection of primary radiation. This can enhance the effective detection efficiency of the imaging system.[9]

Image processing

Prior to display, digital images commonly undergo a series of processing steps. These processes can be divided into preprocessing and post-processing. Image preprocessing consists of correction and scaling.

The first type of processing includes image corrections for detector defects or nonuniformities often present on raw digital images. The second type of preprocessing includes reduction of the full dynamic range of the raw image to the range of perception capability of the human eye.[9] Post-processing is commonly divided into three types:

- *Gray-scale processing* involves the conversion of detector signal values to display values. The display intensities of an image are changed by means of either a look-up table or windowing and leveling.
- *Edge enhancement* aims to enhance fine details within the image by manipulating the high-frequency content of the radiograph, using a variant of the unsharp masking technique in which a blurred version of the image is formed, and a fraction of the resultant image is subtracted from the original image.
- *Multifrequency* processing involves an even more flexible manipulation of multiple portions of the frequency spectrum. The image is initially decomposed into multiple frequency components, and the component images are then weighted and added back together. If the processing parameters are set optimally, the resultant image can compress the overall dynamic range of the image while at the same time enhance local contrast.[9]

Image display

Soft-copy display is the optimal way of viewing digital chest radiography. The conventional method of displaying digital radiographs has been on cathode-ray tubes, which still dominate the market.[25] Active-matrix liquid-crystal displays are rapidly replacing cathode-ray tubes.[26–28] The advantages of liquid-crystal displays include improved resolution, reduced weight, smaller form factor, reduced reflection, improved bit depth, and improved luminance range, although disadvantages in terms of limited viewing angle and structured noise may be practically relevant.[29,30]

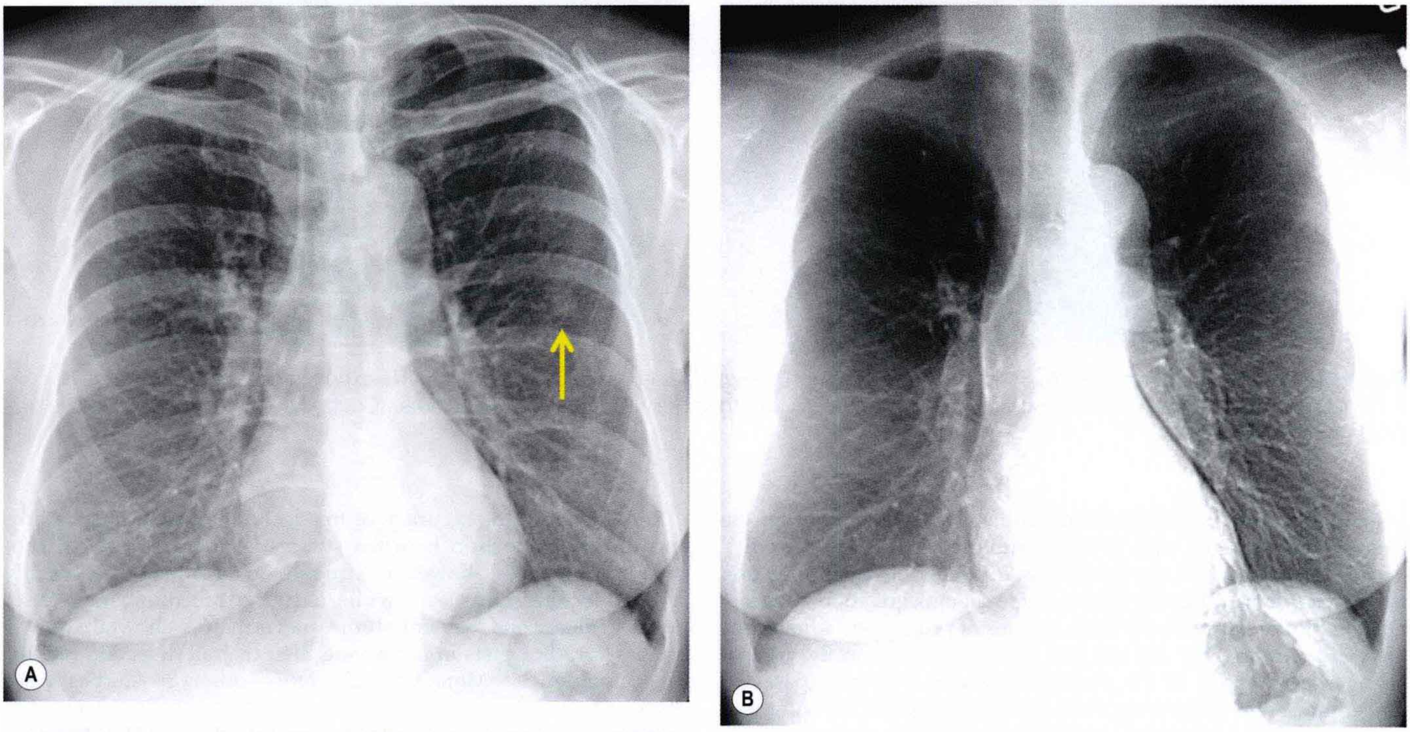
Another recent trend has been the increased acceptance of color monitors, some of which have shown acceptable technical performance for radiographic applications.[31] The use of color monitors offers the advantage of being able to accommodate applications other than image viewing on the same device, with workflow and multitasking advantages.[9] Color monitors also make it possible to take advantage of color for viewing multidimensional chest images on the same display. It is thus expected that color liquid-crystal displays will gradually replace monochrome monitors in clinical practice.[9]

Novel applications

Digital chest radiography lends itself the development of new techniques to improve the detection of subtle lesions. These techniques include algorithms typically coupled with methodological innovations that use imaging physics to improve lesion conspicuity. Three notable novel applications are dual-energy imaging, temporal subtraction imaging, and digital tomosynthesis. All of these techniques are implemented by using a conventional chest radiography system coupled with a digital imaging receptor.[9] Whether they will be widely adopted remains to be seen.

Dual-energy subtraction imaging is used to generate images of two independent tissue types, most commonly bone and soft tissue. The dual-energy technique distinguishes bone from soft tissue by using the known energy dependence of X-ray attenuation in soft tissue and in bone. By means of photoelectric absorption, calcified structures attenuate more heavily than soft tissue structures. Therefore, the contrast of calcium diminishes with increases in beam energy more than does the contrast between soft tissues. Thus, the image obtained at the lower energy will show a larger fraction of contrast from bone than from soft tissue. These two images may be combined such that the soft tissue and calcium components can be isolated. Typically, an image containing only calcified structures and an image containing only soft tissue structures are generated. Dual-energy subtraction radiography can improve lung nodule conspicuity by eliminating overlying anatomic noise from the bones (Figs 1.4 and 1.5). The technique can also be used to better demonstrate calcium in lesions.[32–36]

Temporal subtraction techniques aim to selectively enhance areas of interval change by subtracting the patient's previous radiograph from the current radiograph.[37] The quality of the difference image strongly depends on the success of two-dimensional registration and warping of the two radiographs, so that the variations in patient positioning can be minimized.[38,39] The difference image is uniformly gray in areas of no change, whereas areas that stand out on the gray background indicate interval change (Fig. 1.6). Several studies[40–42] have shown that temporal subtraction improves the



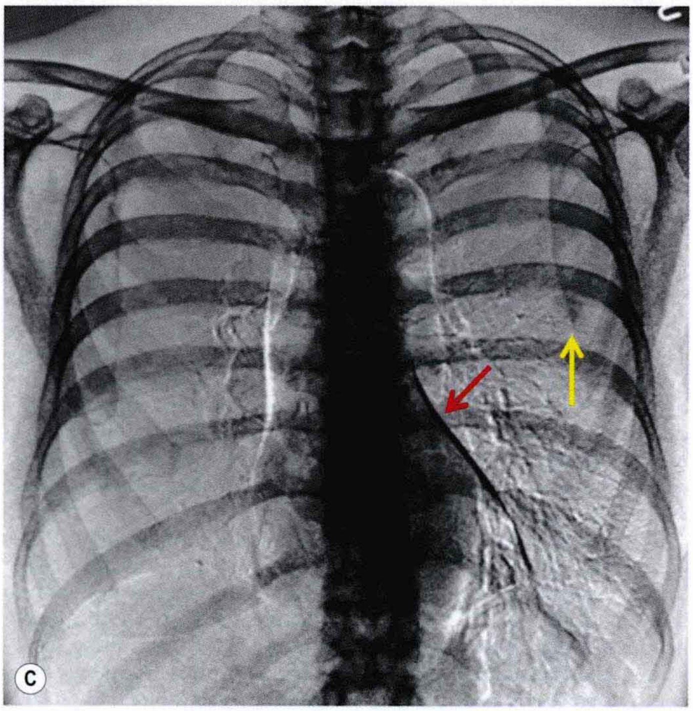


Fig. 1.4 Dual-energy subtraction. **A**, Frontal chest radiograph shows possible left lung nodule (arrow). **B**, Bone subtracted image shows no evidence of nodule. **C**, Soft tissue subtracted image shows nodular opacity consistent with callus from healing rib fracture (yellow arrow). Note motion-induced artifact along the left heart border (red arrow). (Courtesy of H Page McAdams, Durham, NC, USA.)

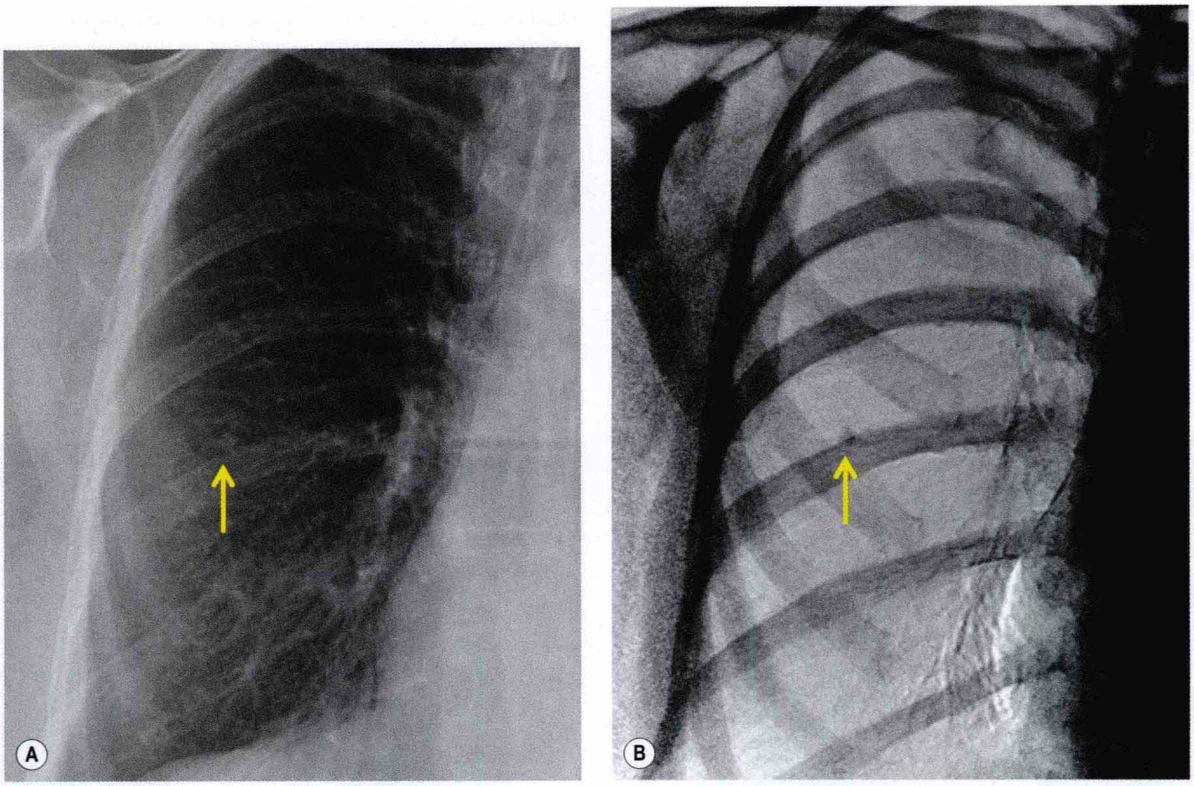


Fig. 1.5 Dual-energy subtraction. **A**, Frontal chest radiograph shows right lung nodule (arrow) in a TB-exposed patient. **B**, Soft tissue subtracted image confirms that the nodule is calcified (arrow) and thus likely to represent granuloma. (Courtesy of H Page McAdams, Durham, NC, USA.)

visual perception of subtle abnormalities. A 20% reduction in the average reading time with temporal subtraction was also noted.[9,43]

Digital tomosynthesis can produce an unlimited number of section images at arbitrary depths from a single set of acquisition images.[44] During motion of the X-ray tube, a series of projection radiographs are acquired, and the anatomy at different depths in the patient changes orientation in the projection images owing to parallax. These projection images are then shifted and added to bring into focus objects in a predefined plane. By varying the amount of shift, different plane depths can be reconstructed (Fig. 1.7). Objects outside of the focus plane are blurred. Currently, chest imaging with tomosynthesis is one of the areas receiving the most clinical and research interest.[9,44]

Computer-aided diagnosis

Computer-aided detection (CAD) and computer-aided diagnosis (CADx) systems rely on combinations of image-processing, pattern-recognition, and artificial intelligence techniques. The application of CAD and CADx analysis in chest radiography has followed a traditional model of first detecting and then characterizing potential abnormalities.[45,46] Image-processing algorithms are applied to identify regions of interest that appear suspicious according to predefined clinical expectations. Image feature analysis then seeks to determine the morphologic and textural characteristics of candidate regions. Finally, feature-based decision analysis provides a definitive assessment of candidate regions.[9]

The majority of CAD applications involve the detection of pulmonary nodules.[47] Typically, morphology-based image processing is applied to detect nodular-appearing structures, while more detailed morphologic and texture analyses eliminate false-positive nodule-like structures (Figs 1.8–1.10). The final decision is made by applying a linear classifier, a neural network, or a rule-base algorithm that merges the image findings into a final binary decision.[9]

Reports about the accuracy of this technique vary substantially, and direct comparison between studies is not possible. All proposed approaches, however, struggle to maintain a clinically acceptable sensitivity level while reducing the number of false-positive detections. Several studies nevertheless show that CAD can assist radiologists in improving their overall detection rate for lung nodules.[48–51] Moreover, laboratory observer studies have shown promising results for applications designed to determine the malignant potential of pulmonary nodules.[40,52] CAD techniques have also been applied to the detection and differentiation of interstitial lung disorders, with varying success.[53–56] Finally, less fully explored CAD applications include the detection of cardiomegaly,[57] pneumothorax,[58,59] interval changes,[60] and tuberculosis.[61]

COMPUTED TOMOGRAPHY

CT relies on the same physical principles as conventional radiography: the absorption of X-rays by tissues with constituents of differing atomic number. With multiple projections and computed calculations of radiographic density, differences in X-ray absorption can be displayed in a cross-sectional format. The basic components of a CT machine are an X-ray tube and an array of X-ray detectors opposite the tube. The signal from the X-ray detectors is reconstructed by a computer. The speed with which a CT scanner acquires a single sectional image depends on the time the anode takes to rotate around the patient.

Volumetric (formerly referred to as spiral or helical) CT has altered the clinical CT imaging protocols developed in the 1990s.[62] The basic principle of volumetric CT entails moving the patient into the CT gantry at a constant rate while data are continuously acquired, often within a single breathhold.[63,64] The resulting 'corkscrew' of information is then reconstructed, most frequently as a contiguous set of axial images, similar to conventional single-slice

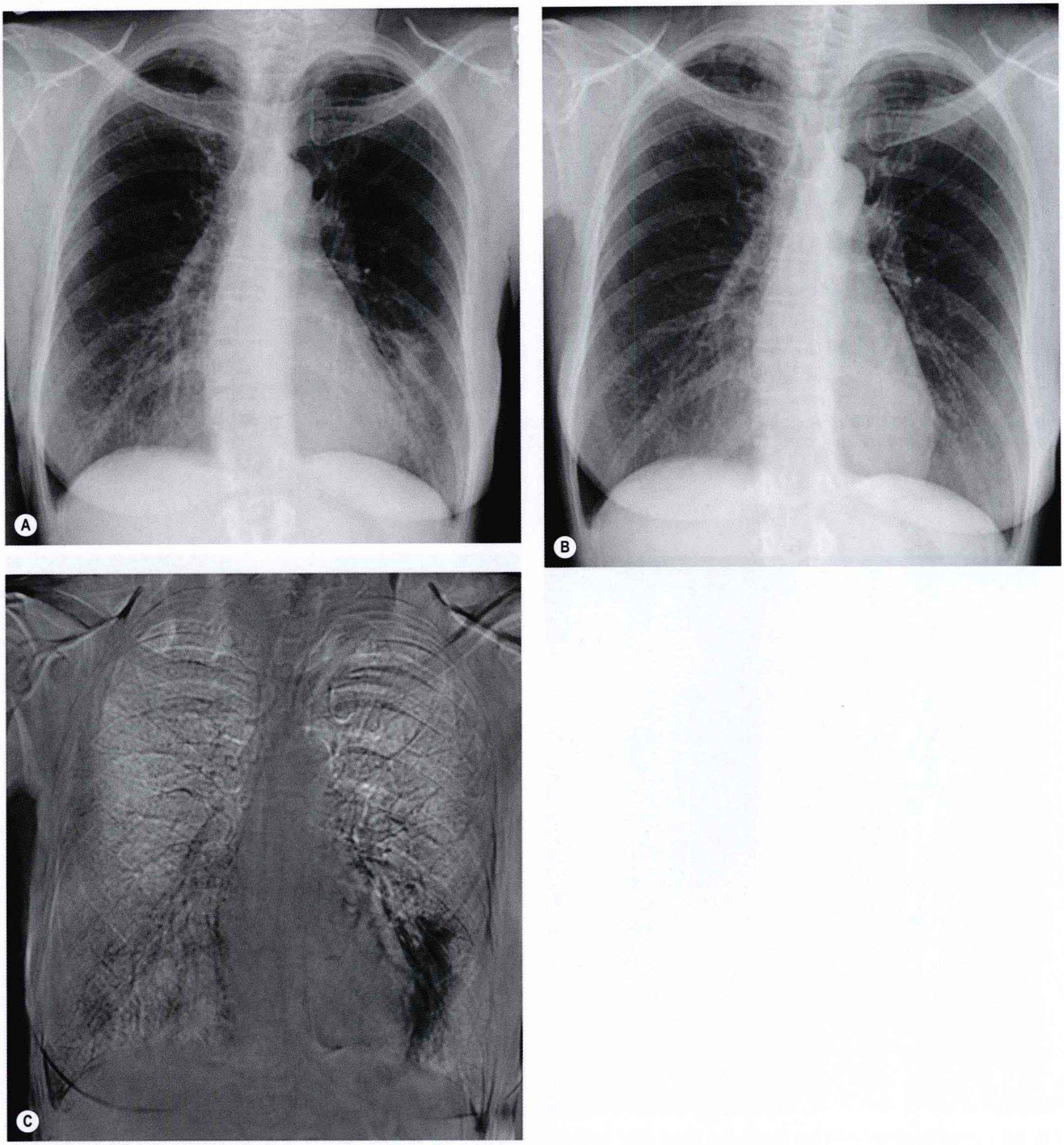


Fig. 1.6 Temporal subtraction. **A**, Current radiograph shows subtle lingular opacity. **B**, The opacity was not present on the previous radiograph. **C**, Temporal subtraction image emphasizes the new opacity. (Courtesy of Heber MacMahon, Chicago, IL, USA.)

CT sections. To achieve this, interpolation is needed because direct reconstruction results in nonorthogonal images of nonuniform thickness. Continuous volume CT scanning has several advantages: (1) rapid scan acquisition in one or two breathholds; (2) reduced volume of contrast needed for optimal opacification of vessels, for example the pulmonary arteries; (3) no misregistration between sections obtained in one acquisition, thus improving detection of small structures; and (4) potential for multiplanar or three-dimensional reconstructions.[65–68]

The advent of MDCT technology has revolutionized the diagnostic potential of CT by definitively transforming CT from an axial cross-sectional technique into a true three-dimensional technique that allows for arbitrary selection of section planes and volumetric display of the acquired data (Figs 1.11–1.13). Most importantly, MDCT permits shorter acquisition times and greater anatomic coverage.[69] The potentially huge number of images routinely generated by clinical protocols, however, represent a challenge in terms of efficient interpretation and the logistics of image storage

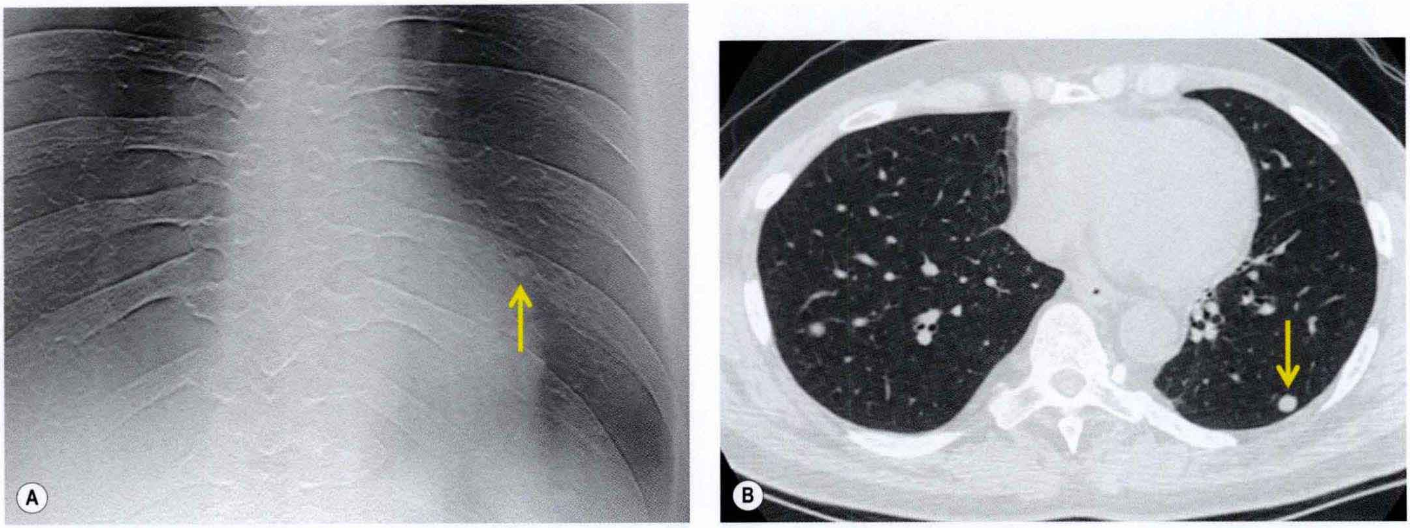


Fig. 1.7 Digital tomosynthesis. **A**, Tomosynthesis image shows left lung nodule overlying a rib (arrow). **B**, Presence of subpleural nodule is confirmed by CT (arrow). (Courtesy of H Page McAdams, Durham, NC, USA.)

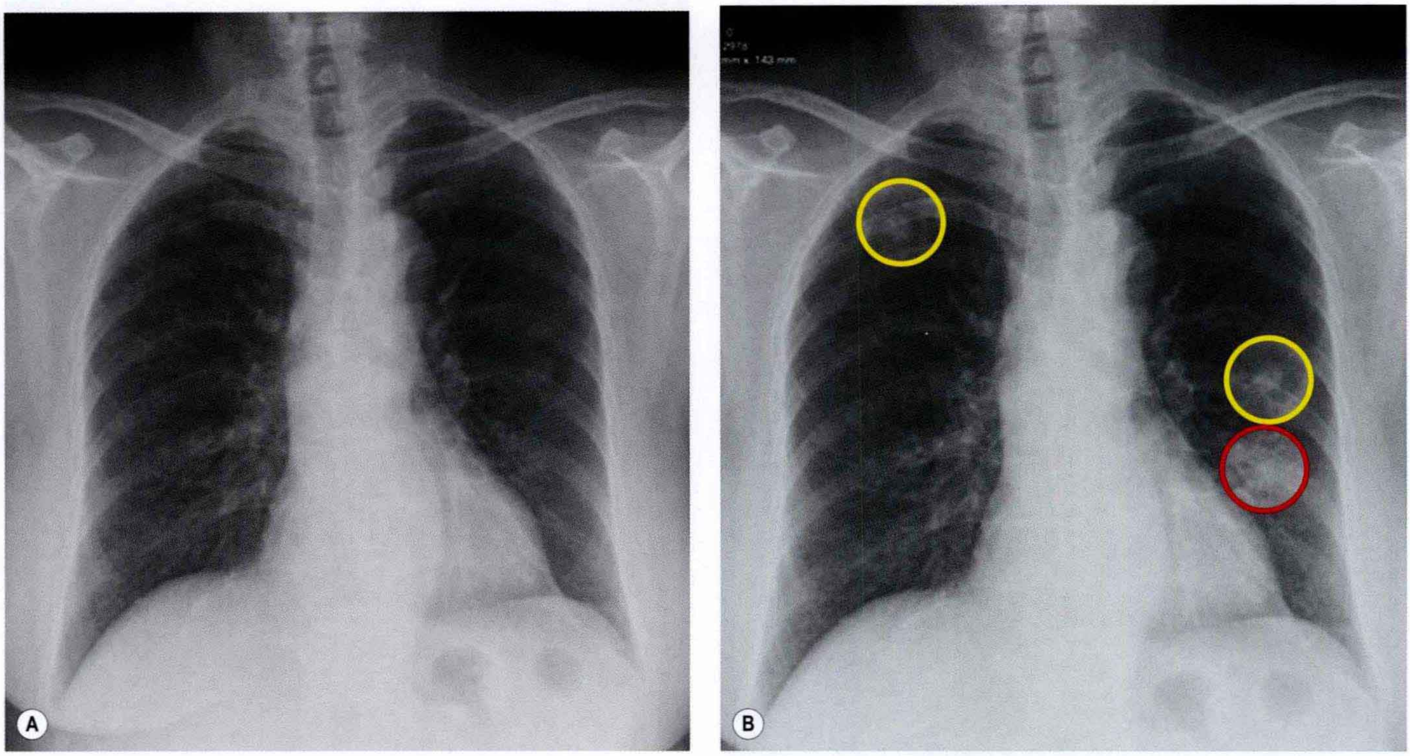


Fig. 1.8 Computer-aided detection (CAD). **A**, Frontal chest radiograph shows three lung nodules, all of which are successfully detected by CAD shown in **B**. While two of the nodules are easy to see (yellow circles), one nodule (red circle) is more difficult to detect due to an overlying rib.

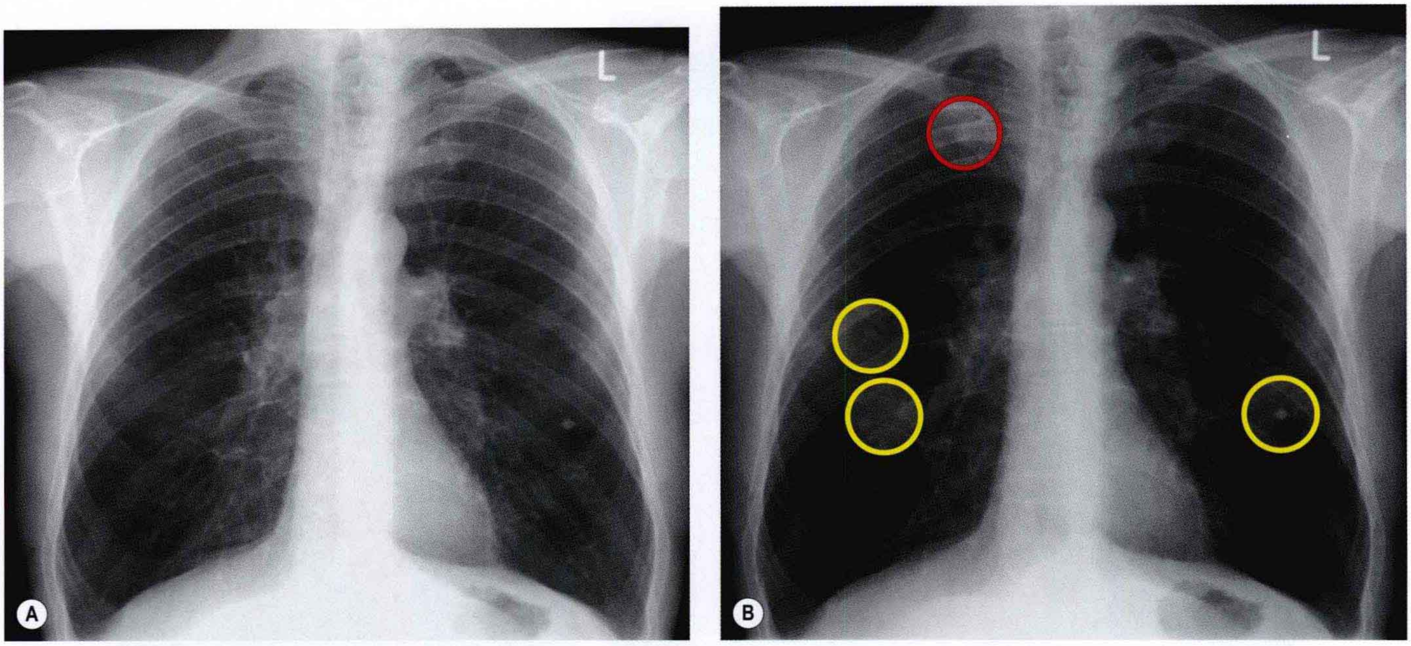


Fig. 1.9 Computer-aided detection (CAD). **A**, Frontal chest radiograph shows three lung nodules detected by CAD shown in **B**. The three nodules (yellow circles) are relatively easy to see. CAD also highlights one false-positive nodule (red circle), in fact corresponding to calcified costochondral cartilage of the first rib.

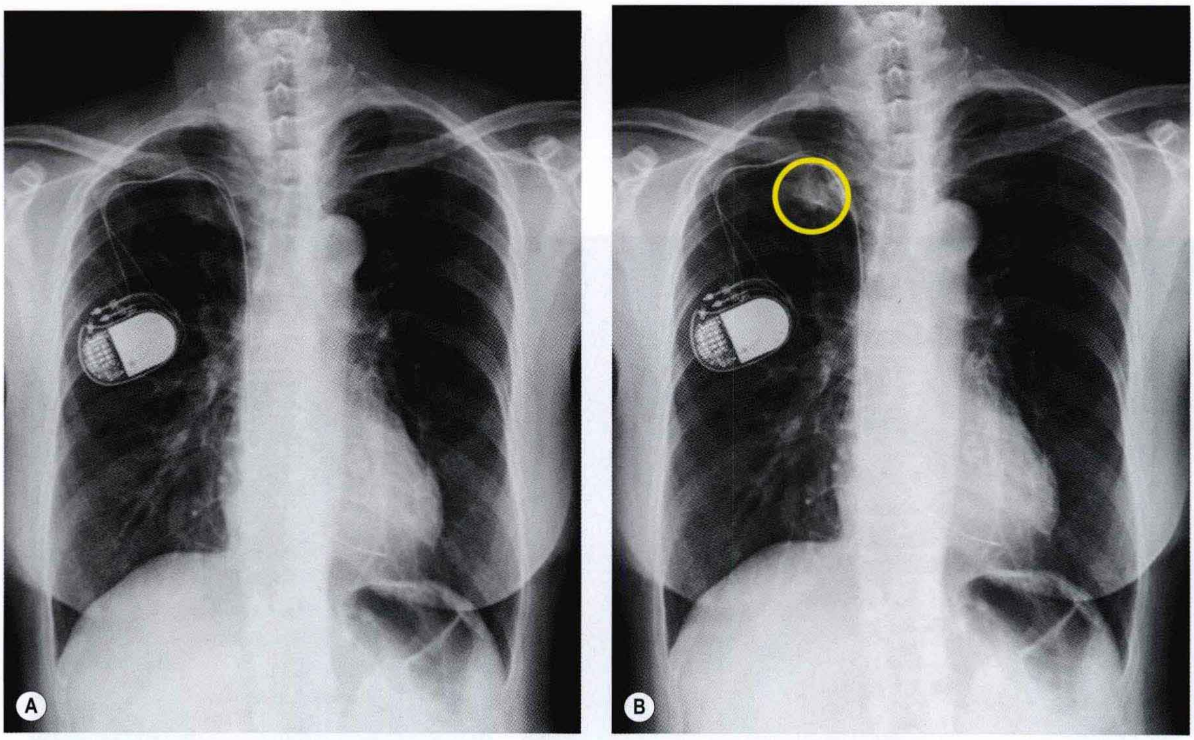


Fig. 1.10 Computer-aided detection (CAD). **A**, Chest radiograph shows a lung nodule correctly identified by CAD shown in **B** (yellow circle), despite being directly adjacent to the lower margin of the first rib.

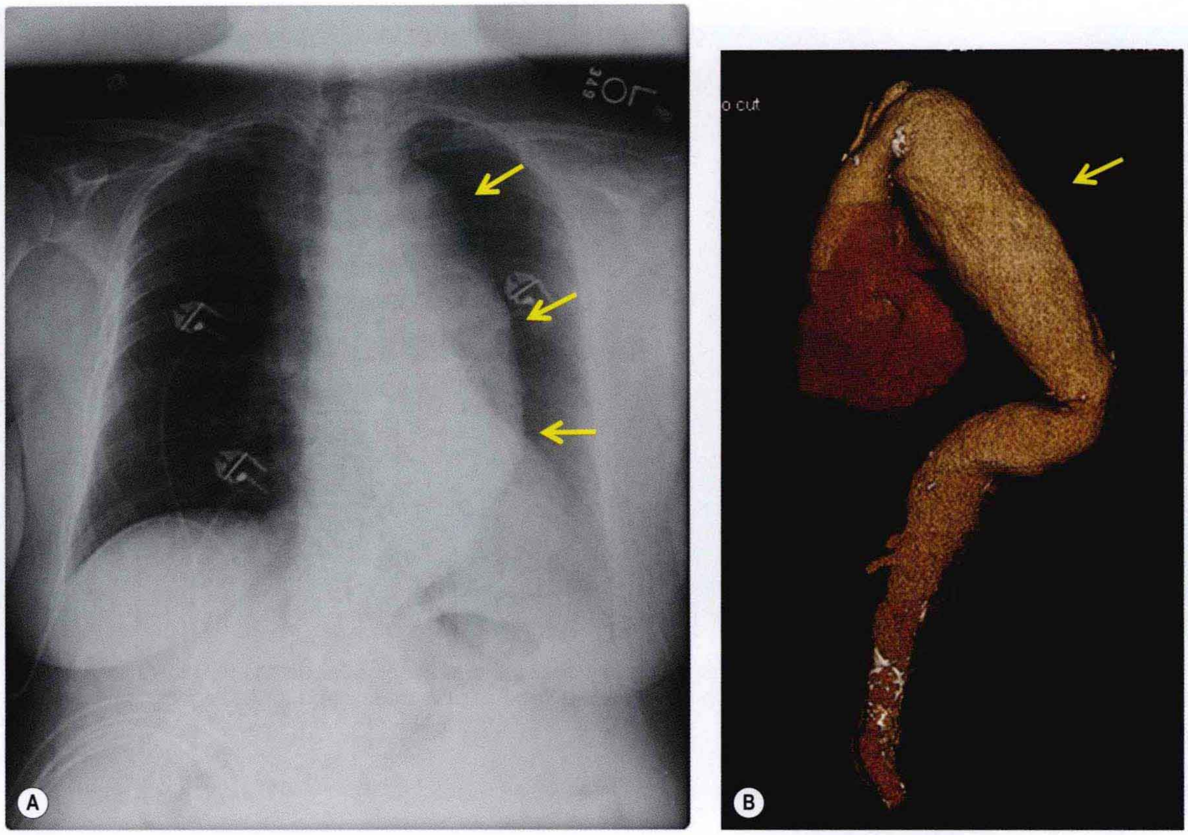


Fig. 1.11 Aortic aneurysm. **A**, Chest radiograph shows enlarged and tortuous aorta (arrows) in a patient with chest pain. **B**, Surface shaded reconstructions of emergency CT show large aneurysm of the aortic arch and the descending aorta (arrow).

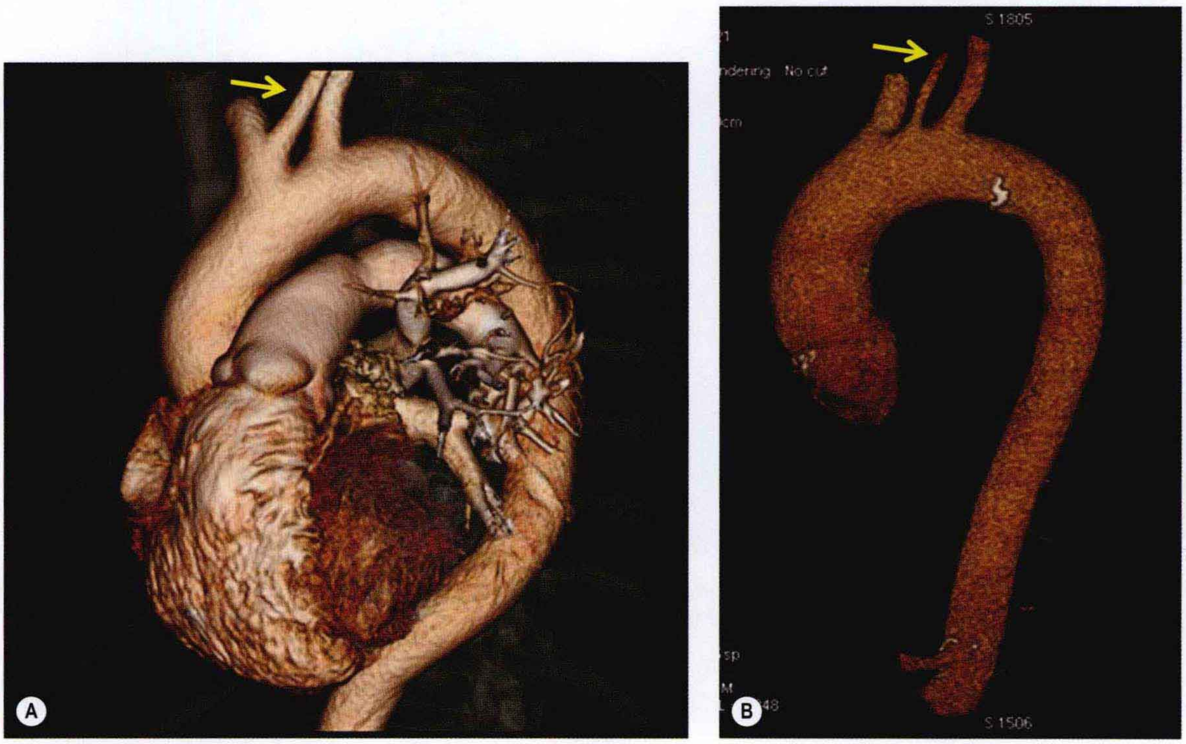


Fig. 1.12 **A**, Surface shaded reconstruction of the aorta shows asymptomatic variant of left common carotid artery arising from the brachiocephalic trunk (arrow). **B**, Normal origin of the vessel (arrow) is shown in a different patient.

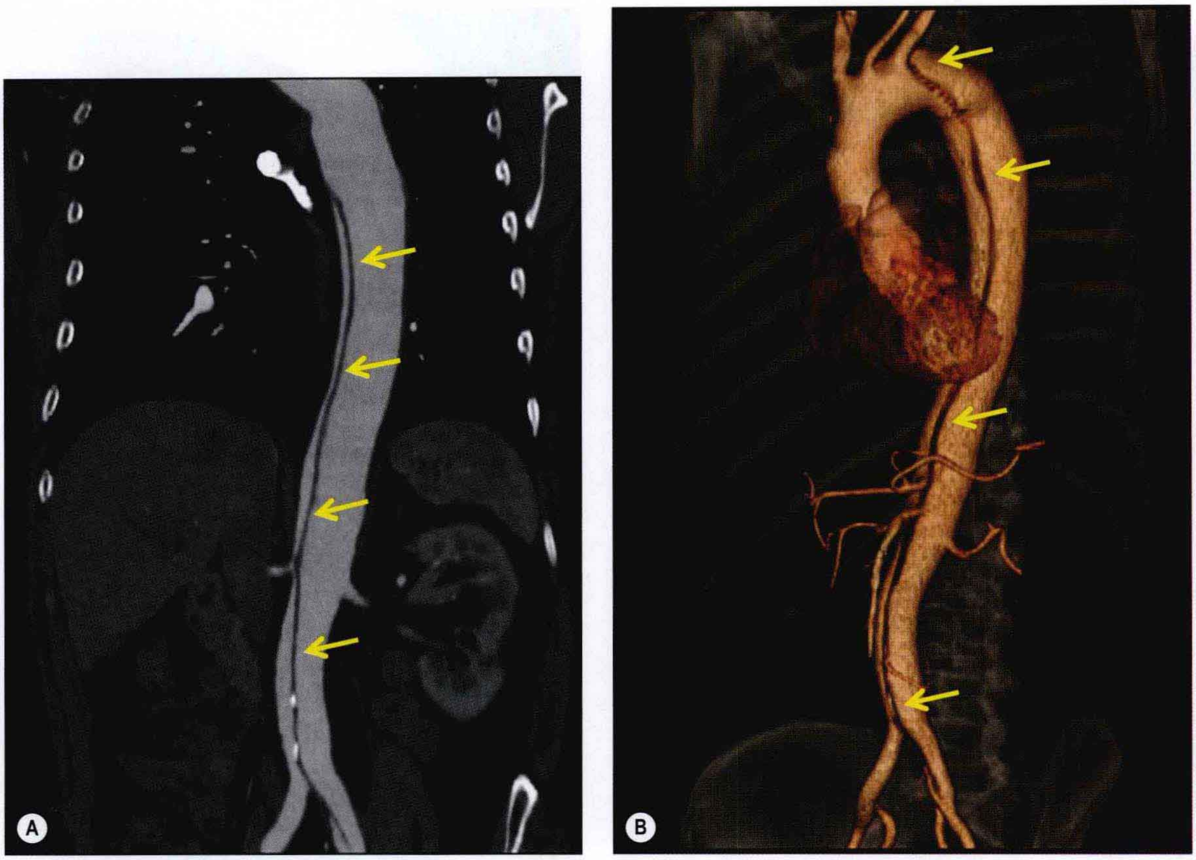


Fig. 1.13 Aortic dissection. **A**, Curved array reformat shows intima (arrows) in a patient with aortic dissection. **B**, Surface shaded display in the same patient shows the intima membrane in its entire length (arrows).

and transmission. The technique of volumetric MDCT scanning of the thorax continues to be refined, and the full potential of acquiring and analyzing data in a truly volumetric manner is still to be realized.[70,71]

General considerations

The CT image is composed of a matrix of picture elements (pixels). There are a fixed number of pixels within the picture matrix so that the size of each pixel varies according to the diameter of the circle to be scanned. The smaller the scan circle size, the smaller the area represented by a pixel and the higher the spatial resolution of the final image. In practical terms the field-of-view size should be adjusted to the size of area of interest, usually the thoracic diameter of the patient. Depending on the field-of-view size, the pixel size varies between 0.3 mm and 1 mm across. By selecting a specific area of interest, the operator can achieve an even increased spatial resolution for that region (targeted reconstruction of the raw data). In a clinical context, targeted reconstruction is used only when the finest morphologic detail is required.

Sometimes there is a marked difference in the appearance of CT images acquired on different scanners. This is the result of differences in the software reconstruction algorithms that 'smooth' the image to a greater or lesser extent by averaging the density of neighboring pixels. Smoothing is used to reduce the conspicuity of image noise and improve contrast, but it has the drawback of reducing the definition of fine structures. The lung is a high-contrast environment, and smoothing here is less necessary than in other parts of the body. Higher spatial resolution algorithms, which make image noise more conspicuous, are generally more desirable, and it has been recommended that they should be applied to both standard thick sections and high-resolution CT (HRCT).[72,73]

Acquisition parameters

Although a single CT section appears as a two-dimensional image, it has a third dimension of depth. Thus each pixel has a volume, and the three-dimensional element is referred to as a voxel. The average radiographic density of tissue within each voxel is calculated, and the final CT image consists of a representation of the numerous voxels in the section. The single attenuation value of a voxel represents the average of the attenuation values of all the structures within the voxel. The thicker the section, the greater the chance of different structures being included within the voxel and so the greater the averaging that occurs. The most obvious way to reduce this 'partial volume' or 'volume averaging' effect is to use thinner sections (Fig. 1.14).

The entire thorax is now usually examined with contiguous sections. MDCT has brought section thickness down to a range of 0.75–2.5 mm. Additional dedicated thin sections are sometimes required to clarify partial volume effects or to study areas of anatomy that are oriented obliquely to the plane of scanning. Specific examples of the use of thin sections to display differential densities, which would otherwise be lost because of the partial volume effect, is the demonstration of small foci of fat within a hamartoma, or of calcifications within a pulmonary nodule. Thin sections of 1–1.5 mm thickness are also used to study the fine morphologic detail of the lung parenchyma (HRCT). Apart from the evaluation of diffuse lung disease, when sampling of a few parts of the lung (traditionally with sections taken at 10–30 mm intervals) is adequate, contiguous section scanning is necessary to allow accurate interpretation in most clinical situations.

For volumetric CT scanning, consideration needs to be given to the speed of table travel, volume of interest, duration of scanning (usually within one breathhold), and reconstruction interval. Pitch is defined as the distance traveled by the table per gantry revolution