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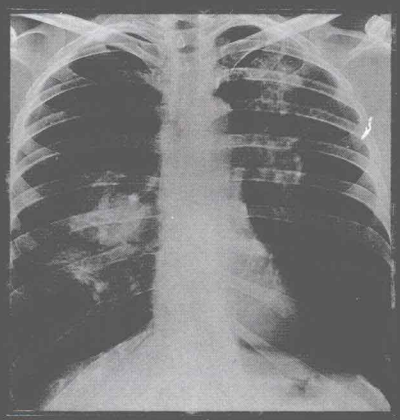
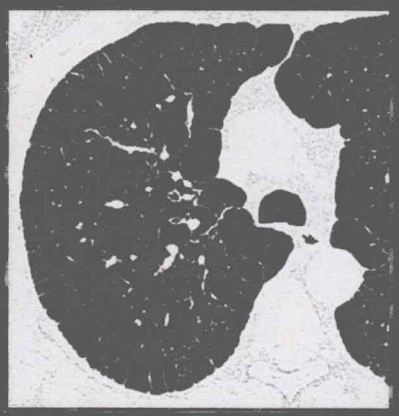
LUNG DISEASES

IN THE CHEST





IMAGING OF THE DISEASES OF THE CHEST

FOURTH EDITION

David M. Hansell

Professor of Thoracic Imaging, Department of Radiology, Royal Brompton Hospital, London, UK

Peter Armstrong

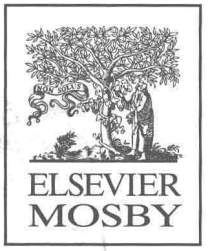
Professor of Radiology, Academic Department of Radiology, St. Bartholomew's Hospital, London, UK

David A. Lynch

Professor of Radiology and Medicine, Department of Radiology, University of Colorado Health Services Center, Denver, Colorado, USA

H. Page McAdams

Associate Professor of Radiology, Department of Radiology, Duke University Medical Center, Durham, North Carolina, USA



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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. Our aim has been to provide answers to the many queries that in our experience arise in the day-to-day practice of chest radiology.

From the preface to the first edition (1990)

Despite the advances in thoracic imaging that have occurred since the publication of the first edition, the scope and intention of this fourth edition remain the same.

A major change has come about with the retirement of Alan Wilson and Paul Dee and the stepping aside of Peter Armstrong as senior author. We are delighted that Page McAdams and David Lynch agreed to help with updating the text and renewing many of the illustrations - their expertise and input is readily evident throughout this new edition. The value of the contributions of Alan Wilson and Paul Dee to the preceding editions cannot be overstated. Their legacy is the authoritative bedrock that underpins many of the chapters.

This edition has been completely revised and the references brought up to date. For instance, the ramifications of the latest classification of the idiopathic interstitial pneumonias and the emerging role of PET in lung cancer staging are given particular attention.

In line with the ethos of earlier editions, we have chosen to present the clinical and pathologic features of the differing diseases in varying degrees of detail, based on our perception of the needs of readers. Complicated and rare entities are discussed in much more detail than commoner and well understood conditions. Once again, we hope that our efforts provide a useful resource for anyone who uses thoracic imaging in its many and varied forms.

David M. Hansell
Peter Armstrong
David A. Lynch
H. Page McAdams
2005

Acknowledgements

The number of individuals who have wittingly, or otherwise, helped with the production of this book increases with each succeeding edition, to the extent that it is now impossible to name them all; but that does not diminish our gratitude to them. Buried within some chapters are sections of enduring value from various contributors, who are not individually cited – their reward must be the knowledge that the quality of their contribution has ensured its continued inclusion.

Secretarial help with an endeavour of this size is crucial and we have been ably assisted by Jenni Hillsley, Lisa Bolt, Julie Jessop, Ann Willard and Brenda Baker, all of whom undertook the painstaking process of ensuring that the references were correctly transferred to an electronic database. Mary Anne Hansell sorted and renumbered the surviving figures from previous editions and in so doing spared the authors many hours of vexation.

Most of the numerous new illustrations were acquired in digital format direct from electronic image archives but the skills of medical photographers, notably the Medical Illustration Department of St Bartholomew's Hospital, were still occasionally required. It goes without saying that we have relied on many long-suffering colleagues to supply us with illustrations of some of the rarer conditions and we hope that they spot their prized cases in the following pages. Thanks are due to our publishers, in particular to Joanne Scott who was subjected to many impetuous requests and to Michael Houston for sanctioning most of them. We are also indebted to Nora Naughton and Sam Gear for their flexibility and efficiency at the eleventh hour.

Our wives Mary Anne, Carole, Anne, and Emma, know that we could not have completed this task without their fortitude and support.

David M. Hansell
Peter Armstrong
David A. Lynch
H. Page McAdams

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CHAPTER 1

Technical considerations

The chest radiograph remains the prime imaging investigation in respiratory medicine and the basic technique has changed little over the last 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, ultrasonography and positron emission tomography have complementary roles in a few situations. Refinements to CT scanning protocols continue as more specific applications are developed. Despite technological advances the radiation burden inherent in CT scanning remains relatively high. The trend towards routine narrow-collimation volumetric acquisition represents a significant radiation burden to the patient and the optimal protocol for each patient should always be considered.

CHEST RADIOGRAPHY

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 with the patient positioned so 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 at nearer residual volume (expiratory film) can dramatically 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, to a greater or lesser extent, 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 (Fig. 1.2) 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 of radiographic film 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 (Fig. 1.3). 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 means that there is a net reduction of image-degrading scattered radiation compared with a low-kilovoltage, nongrid technique. With 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

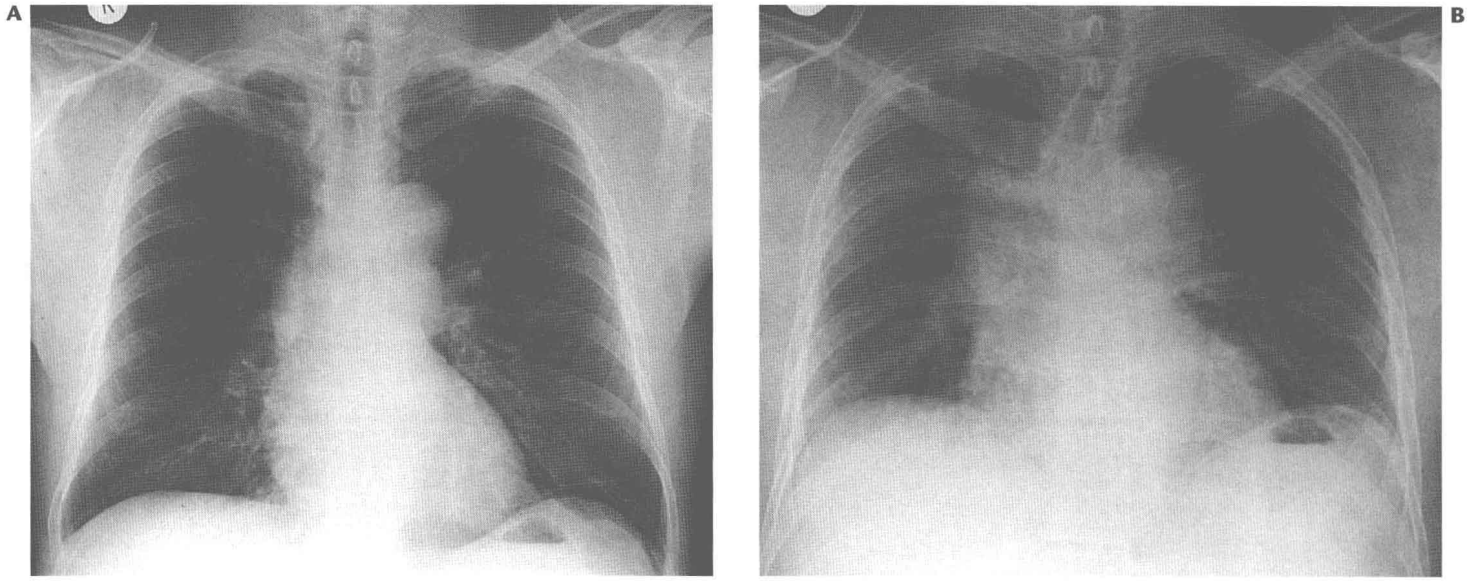

Fig. 1.1 **A**, A normal chest radiograph of an individual breath holding at full inspiration. **B**, By comparison, at full expiration, the mediastinum appears abnormally widened and there is the appearance of diffuse lung shadowing.

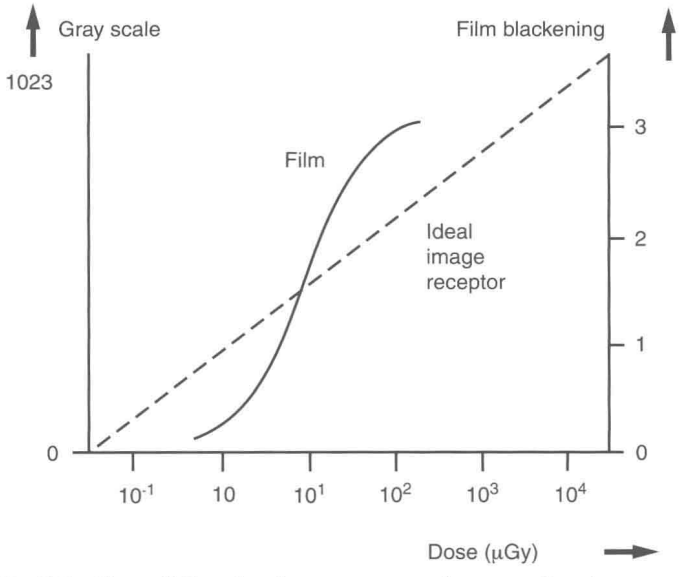

Fig. 1.2 The radiation dose/response curve of a conventional film–screen combination compared with the linear response (dotted line) of an ideal image receptor.

routine examination of the lungs and mediastinum, low-kilovoltage radiographs provide excellent detail of unobscured 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.

A major advance was the introduction of "faster" rare earth phosphor screens and the development of wide-latitude film. The improved light emission from rare earth phosphors over traditional calcium tungstate crystal screens resulted in shorter exposure times and thus sharper images. A significant improvement for chest imaging came with the development of an asymmetric combination consisting of a thin front screen and high-contrast film emulsion and, on the reverse side of the film base, a thicker back screen and a low-contrast film emulsion. With this combination the two ends of the wide spectrum of transmission of x-rays through the thorax could be accommodated. Such a film–screen combination reveals significantly more detail in the mediastinum and lung obscured by the diaphragm and heart[3,8,9] (Fig. 1.4). However, there may be some loss of detail in the unobscured lung.[10]

Because attenuation of x-rays by the mediastinum is up to 10 times greater than that by the lungs, several devices have been designed to produce a more uniformly exposed chest radiograph. One of the most widely used was the advanced multiple beam equalization radiography (AMBER) system.[5] The AMBER unit comprised a horizontally oriented scanning slit beam effectively divided into 20 segments, each modulated by an electronic feedback loop from 20 corresponding detectors on the far side of the patient. Such a system is particularly good at demonstrating pulmonary lesions behind the heart and diaphragm (Fig. 1.4).[11] Nevertheless, recent advances with digital acquisition technology mean that scanning equalization devices are now largely obsolete.

Extra radiographic views

The frontal and lateral projections suffice for most purposes. Other radiographic views are becoming much less frequently requested because of the ready availability of cross-sectional

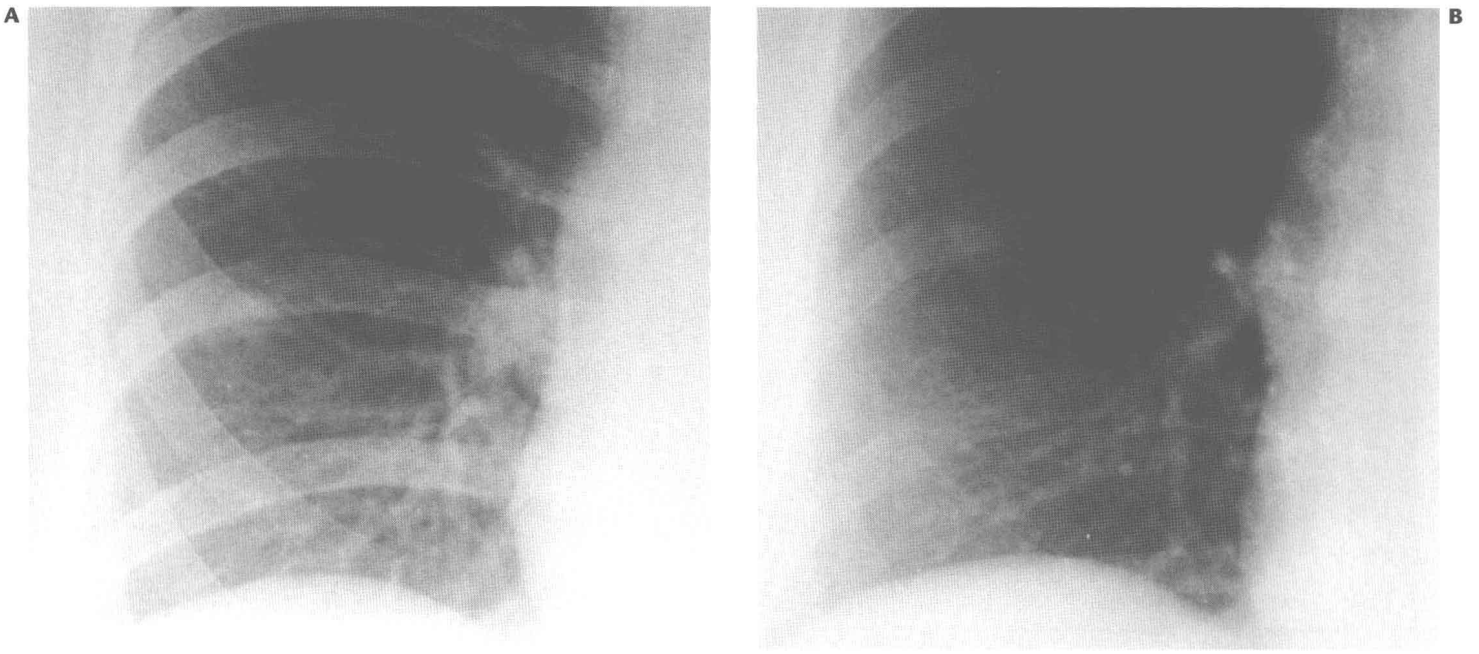

Fig. 1.3 The effect of kilovoltage on chest radiography. **A**, A low-kilovoltage (70 kVp) technique providing good detail of the ribs. A small nodule in the right mid zone is partially obscured by overlying ribs. **B**, By comparison, a high-kilovoltage (140 kVp) technique radiograph diminishes the visibility of the ribs and reveals the small carcinoid tumour in the right mid zone. (Courtesy of Dr MB Rubens, London)

imaging, particularly CT. Nevertheless, an additional view may occasionally solve a particular clinical problem quickly and cheaply.

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 movement of fluid in the pleural space. If a pleural effusion is not loculated, it gravitates to the dependent part of the pleural cavity. 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 free pleural fluid. A lateral shoot-through radiograph may be used to advantage to show a small anterior pneumothorax in recumbent patients in intensive care (Fig. 1.5).[12]

A lordotic view is now rarely used, but is included here for completeness. It is performed by angling the x-ray beam 15° craniad, either by positioning the patient upright and angling the beam up or by leaving the beam horizontal and leaning the patient backward. In this way the lung apices are better demonstrated, free from the superimposed clavicle and first rib. The lordotic view may be useful for distinguishing a pulmonary shadow from incidental calcification of the costochondral junctions (Fig. 1.6). With the exception of identifying rib fractures and confirming the presence of a rib lesion, such as a lytic metastasis, 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 department. In many medical centers the proportion of portable to departmental chest radiographs has gradually increased over the years. However, the many disadvantages of portable radiography should not be forgotten.

The shorter focus–film distance results in undesirable magnification. High-kilovoltage techniques cannot be used because portable machines are unable to deliver 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 extremely long exposure times. Radiation exposure of nearby patients and staff is a further consideration.

Positioning of bed-bound patients is difficult, and the resulting radiographs are often of 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, especially phosphor plate computed radiography, represents a significant advance.

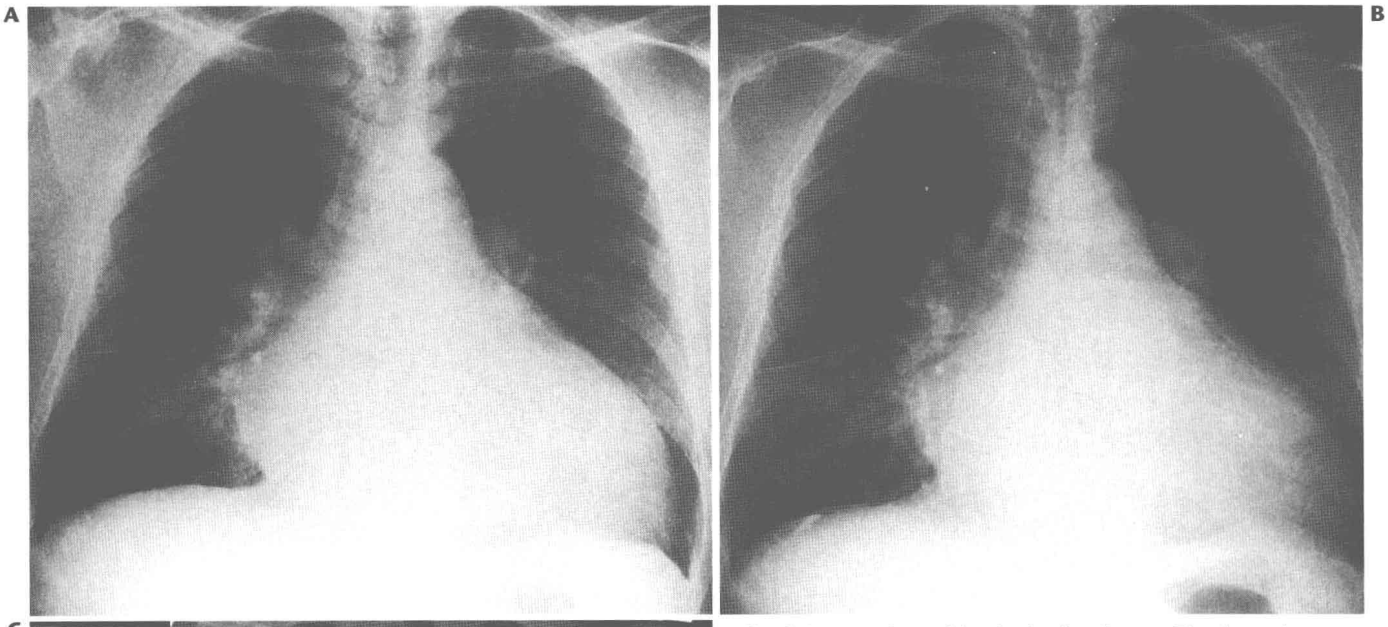

Fig. 1.4 A patient with mitral valve disease. The three chest radiographs were taken at 140 kVp using: **A**, a conventional film and rare earth screen combination; **B**, INSIGHT asymmetric film; C, an AMBER scanning equalization unit. Note the differences in mediastinal detail between the radiographs produced with the three techniques.

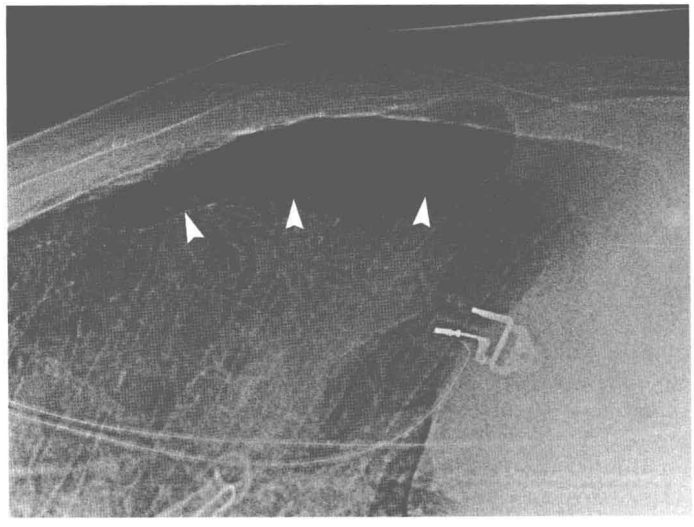

Fig. 1.5 A lateral shoot-through computed radiograph of a patient on the intensive care unit. The supine AP radiograph did not reveal a pneumothorax, but on this lateral radiograph an anterior pneumothorax is seen. The visceral pleural edge of the lung (arrowheads) has fallen away from the anterior chest wall.

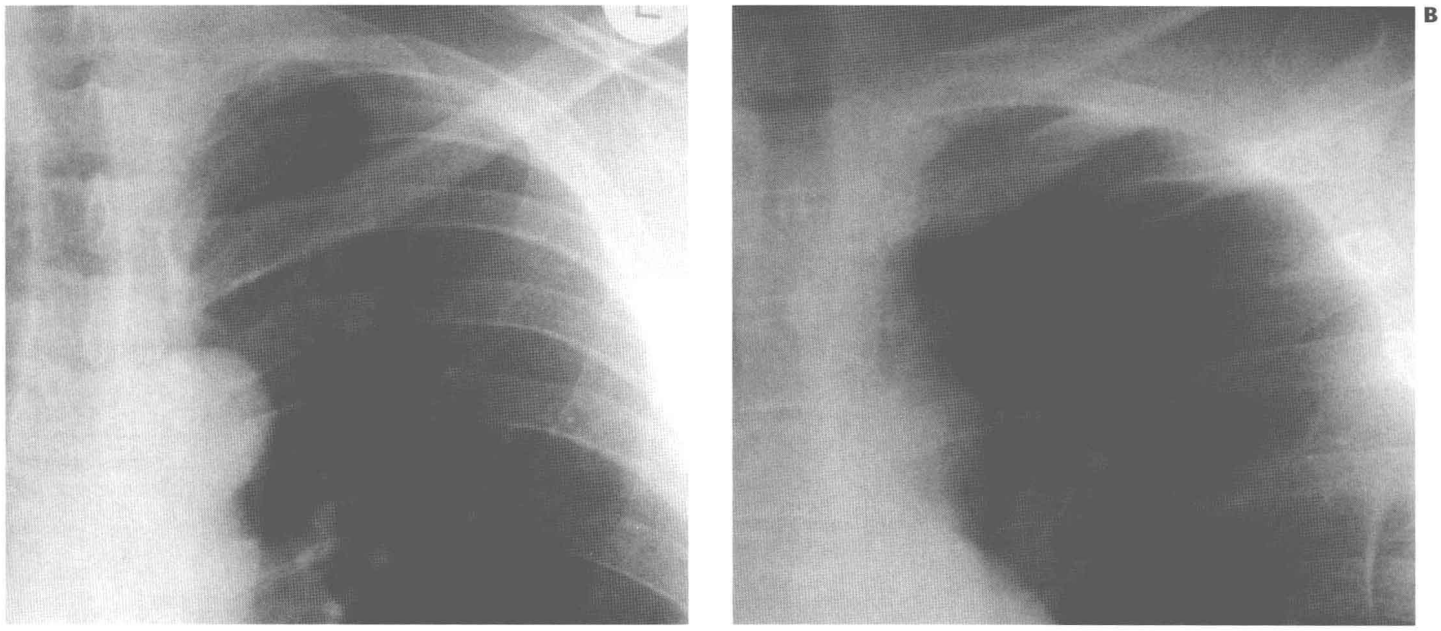

Fig. 1.6 Use of the lordotic view. **A**, A selective view of the left apex showing a small opacity projected over the anterior end of the left first rib. **B**, A lordotic view confirms that the opacity is intrapulmonary, rather than part of a calcified costochondral cartilage.

DIGITAL CHEST RADIOGRAPHY

It has long been recognized that conventional film as a means of image capture, storage, and display represents something of a compromise,[13] and over the years it has become apparent that digital image acquisition, transmission, display, and storage can, with advantage, be applied to projectional chest radiography.[14] The earliest work on digital chest radiography used digitization of conventional film radiographs by means of optical drum scanners or laser scanners. A great deal of useful information that helped to establish the parameters for clinically acceptable digital radiographs derived from observer performance studies of digitized conventional film.[15–17]

A well-established digital system employs conventional radiographic equipment but uses a reusable photostimulable phosphor plate (Europium-doped barium fluorohalide)[18] instead of a conventional film–screen combination. Phosphor plate computed radiography has been used successfully for many years, particularly as a substitute for portable film radiography.[19,20] The phosphor plate is a large-area detector housed in a "filmless" cassette. The phosphor plate stores some of the energy of the incident x-ray photons as a latent image. When the plate is scanned with a focused laser beam, the stored energy is emitted as light that is detected by a photomultiplier and converted to a digital signal (Fig. 1.7). The digital information can then be manipulated, stored and displayed in whatever format is desired (Fig. 1.8). The phosphor plate can be reused once the latent image has been erased by exposure to white light. Most currently available computed radiography systems produce a digital radiograph with a 2 K × 2 K matrix (with a pixel size of 0.2 mm) and a gray scale of up to 1024 discrete levels. The fundamental requirement of segmenting the image into a finite number of pixels entailed much work to determine the relationship between pixel size, which affects spatial resolution, and lesion detectability.[16,21,22] Digital radiography probably cannot match conventional film radiography for the detection of extremely subtle pneumothoraces[23] and early interstitial lung disease.[24] Although an image composed of pixels of the smallest size possible would seem desirable, there is a direct inverse relationship between the pixel size and the cost and ease of data handling. In fact, there is not always a measurably significant difference in observer performance between 2 K and 4 K formats.[25] In general, pixel size is ultimately a practical compromise between image fidelity and ease of data processing and storage.

Single-shot dual-energy imaging, a technique in which the bony thorax can be removed to reveal a "soft tissue image", can be performed relatively simply with phosphor plate computed radiography by separating two phosphor plates by a thin copper filter.[26,27] The resulting images, which separate the anatomic information into bone and soft tissue components, although remarkable, have not gained widespread clinical acceptance, probably because dual-energy imaging is competing with computed tomography to provide similar information.

There has been much interest in systems using selenium as a broad area detector, particularly the commercially available Thoravision system (Philips Medical Systems).[28] This device consists of a large drum coated with amorphous selenium which, on exposure to x-rays, stores an electrostatic charge on the selenium surface. The charge pattern is then recorded by electrometer probes while the drum is rapidly rotated. Amorphous selenium is an exceedingly efficient x-ray detector which is superior to conventional film–screen and phosphor plate radiography. Phantom and clinical studies have shown that selenium detector radiography is at least as good,[29,30] or superior[31,32] to conventional radiography, or indeed phosphor plate radiography.[33,34] The overall appearance of chest radiographs obtained with a selenium detector system seems to be readily accepted by chest and general radiologists.[35] Considerable dose reduction is possible with selenium detector radiography: in studies of both chest phantoms and patients, there was no significance difference in diagnostic performance

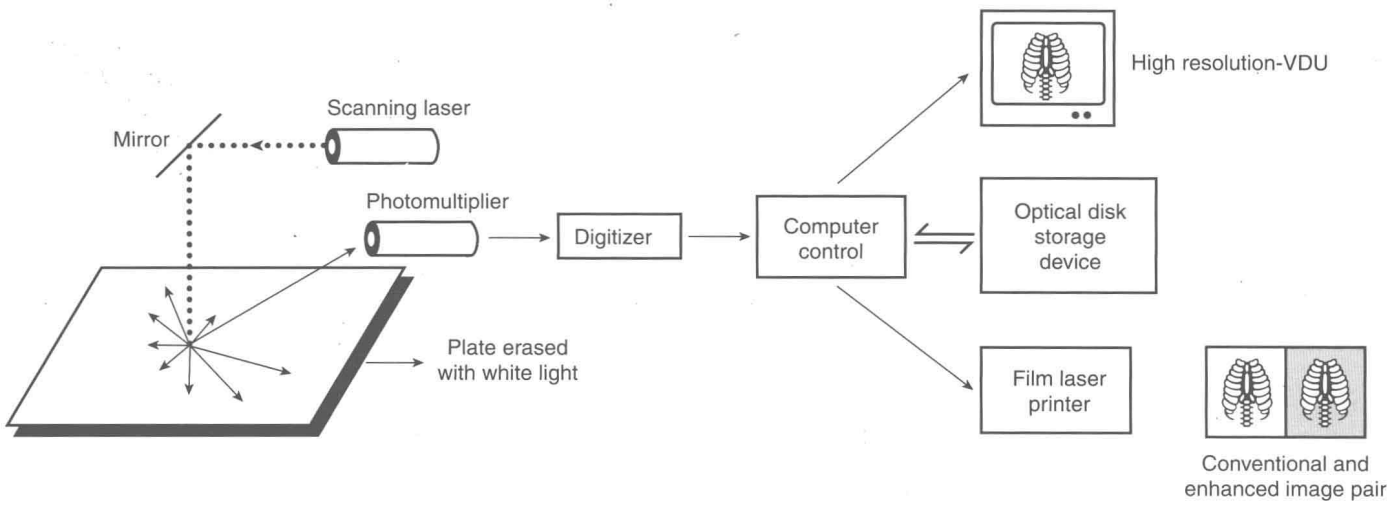

Fig. 1.7 The components of a phosphor plate computed radiography system.

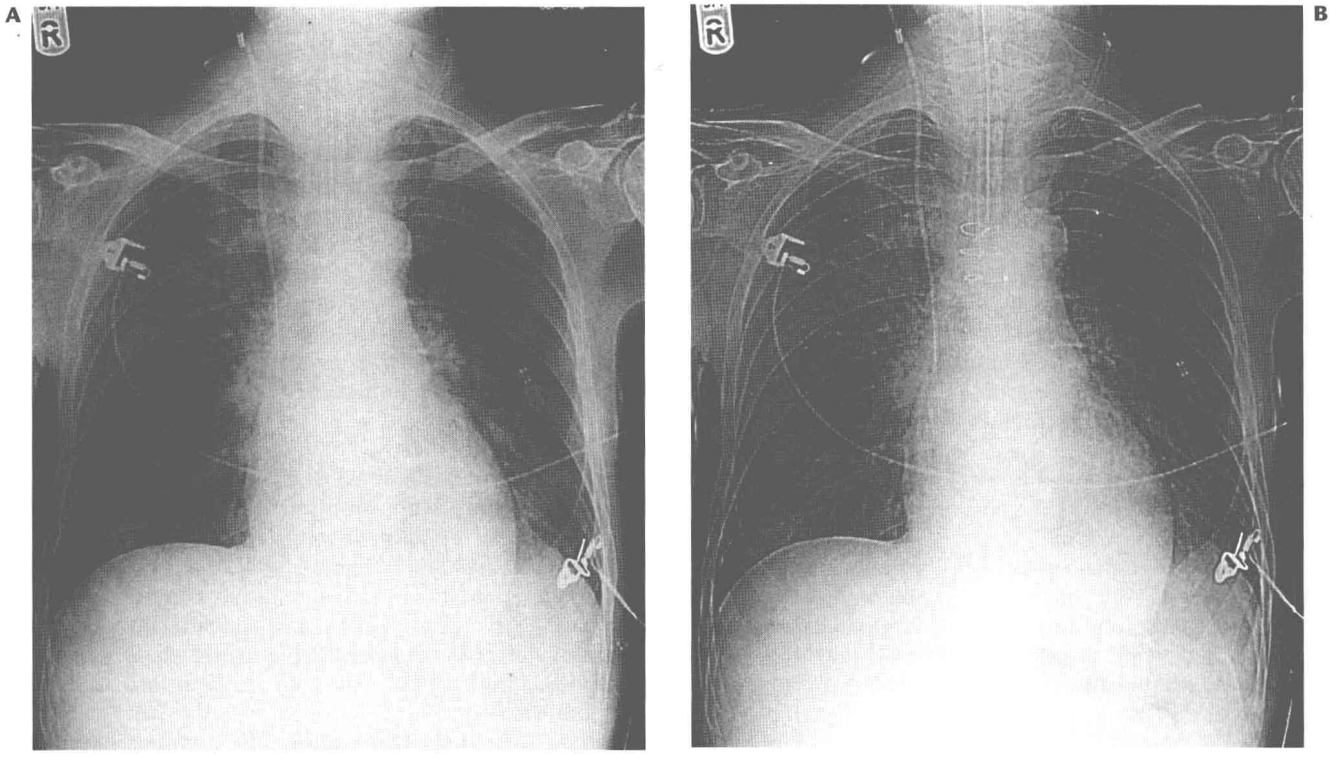

Fig. 1.8 A portable radiograph (phosphor plate) – **A**, manipulated to resemble a conventional radiograph and **B**, digitally processed to give wider latitude and edge enhancement.

between radiographs obtained with conventional exposures and those taken with up to a two-thirds reduction.[36,37] However, if an anti-scatter grid is used the dose needed for optimal images becomes comparable to conventional radiography.[38,39] In contrast to conventional film–screen chest radiography, a high kilovoltage technique may be detrimental and the recommended optimum is in the range 90–110 kVp.[40] Selenium detectors are susceptible to "memory artifact" whereby direct radiation may cause a lingering silhouette on subsequent images, unless sufficient time is given for the artifact to resolve.[41] By the nature of the drum design, selenium detector radiography is currently limited to departmental use, and the new "direct" digital acquisition devices will probably supplant selenium technology.

There is considerable commercial activity in developing direct digital acquisition of images using new solid-state thin-film transistor flat-panel x-ray detectors.[14,42] An increasing array of "flat-panel radiography detectors" are now being marketed – the term is used to denoted the property of an active matrix with direct readout (integrated into a flat panel).[43] At present, many of these panel detectors are too heavy (and expensive) to be used for portable radiography. Clinical experience with flat-panel detectors for thoracic imaging is relatively limited but early

phantom and clinical studies suggest that their favorable quantum efficiency[44–49] and performance compared with phosphor plate computed radiography[50–52] will encourage their further implementation. The report of observer preference for flat-panel chest radiographs, compared to state-of-the-art film–screen and phosphor plate chest radiographs, coupled with an overall decrease in radiation dose of 50% is promising.[53] Whether further technological developments and reduced manufacturing cost of flat-panel detectors will result in the demise of the robust, relatively inexpensive, and now mature phosphor plate technology remains to be seen.

An unequivocal advantage of digital acquisition systems over conventional film radiography is the exactly linear photoluminescence–dose response, which is a full order of magnitude greater than that of conventional film (Fig. 1.2). This extremely wide latitude, coupled with the facility for image processing, produces diagnostically acceptable images over a wide range of exposures. The ability to retrieve an image of diagnostic quality from a suboptimal exposure, which with conventional film would have resulted in an uninterpretable radiograph, has led to the increasing implementation of digital systems. Nevertheless, although overall optical density is maintained on an underexposed computed radiograph, the decreased signal-to-noise ratio may result in a loss of diagnostic content.[54] Early claims by manufacturers of phosphor plate computed radiography systems that a significant radiation dose reduction could be achieved seem to have been exaggerated[55]; indeed, in the intensive care setting the dose may actually be higher than that needed with a flat film–screen combination.[56] In addition, there are several artifacts, peculiar to computed radiography, that may hamper accurate interpretation.[57]

Numerous observer performance studies have shown that computed radiography can equal conventional film–screen radiography in virtually any specific task.[58–61] For this, however, postprocessing of the digital image may have to be used to match the digital radiograph to the task.[62] For example, unsharp masking improves the detection of central lines and other devices on intensive care portable radiographs.[63,64] A problem inherent in all forms of digital manipulation is that enhancement of the image for one purpose degrades it for another.

Many studies suggest that 2 K × 2 K monitors are adequate for making primary diagnoses of digital chest radiographs.[65–67] Indeed, for the interpretation of chest radiographs obtained on a coronary care unit, 1 K × 1 K monitors are probably sufficient.[68] There are many differences between the image appearance, radiation dose, unit cost, and applicability of the various techniques now available for obtaining a chest radiograph. Studies that have attempted to compare, for example, storage phosphor, selenium and film–screen systems[69–71] do not address every one of these factors and the final choice of system depends as much on local circumstances as objective image quality.[72]

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, slight 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 number and geometry of these detectors are variable. The signal from the x-ray detectors is reconstructed by a computer, and the resulting images are either laser-printed or, increasingly, displayed on a workstation. The speed with which a CT scanner acquires a single sectional image depends on the time the anode takes to rotate around the patient. The rotation time of the latest machines can be as fast as 0.42 s, allowing the acquisition of up to 38 images per second. Electron beam ultrafast CT scanning technology dispenses with a rotating mechanical anode: the patient is surrounded by a tungsten target ring, and a focused electron beam sweeps around the tungsten ring at high speed to produce an x-ray beam. Such machines are capable of acquiring an image in 100 ms or less.

Continuous volume (formerly referred to as spiral or helical) scanning has altered routine CT scanning protocols since the 1990s.[73] 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 breath hold.[74,75] The resulting "corkscrew" of information is then reconstructed, most frequently as a contiguous set of axial images, similar to conventional single-slice CT sections. To achieve this, some interpolation is needed because direct reconstruction results in nonorthogonal images of nonuniform thickness. Continuous volume CT scanning has several advantages over conventional CT of the thorax: (1) rapid scan acquisition in one or two breath holds; (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 nodule detection; and (4) potential for multiplanar or three-dimensional (3D) reconstructions.[76–79] With the advent of multidetector (4–16 detector rings) technology, the possible variation in scanning protocols has become vast, as has the potential number of images generated in a single examination.[80] Multidetector computed tomography (MDCT) not only permits shorter acquisition times and greater coverage and image resolution, but also provides data sets that increasingly exploit multiplanar reformations and sophisticated 3D renditions[81] (Fig. 1.9). The potentially huge number of images generated by some protocols, for example more than 350 contiguous 1 mm sections in CT pulmonary angiography, represents a challenge in terms of efficient interpretation and the logistics of image storage 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 form is still to be realized.[82]

General considerations

The CT image is composed of a matrix of picture elements (pixels). There is 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 and 1 mm across. By selecting a specific area of interest, the operator can achieve optimal spatial resolution for that region (targeted

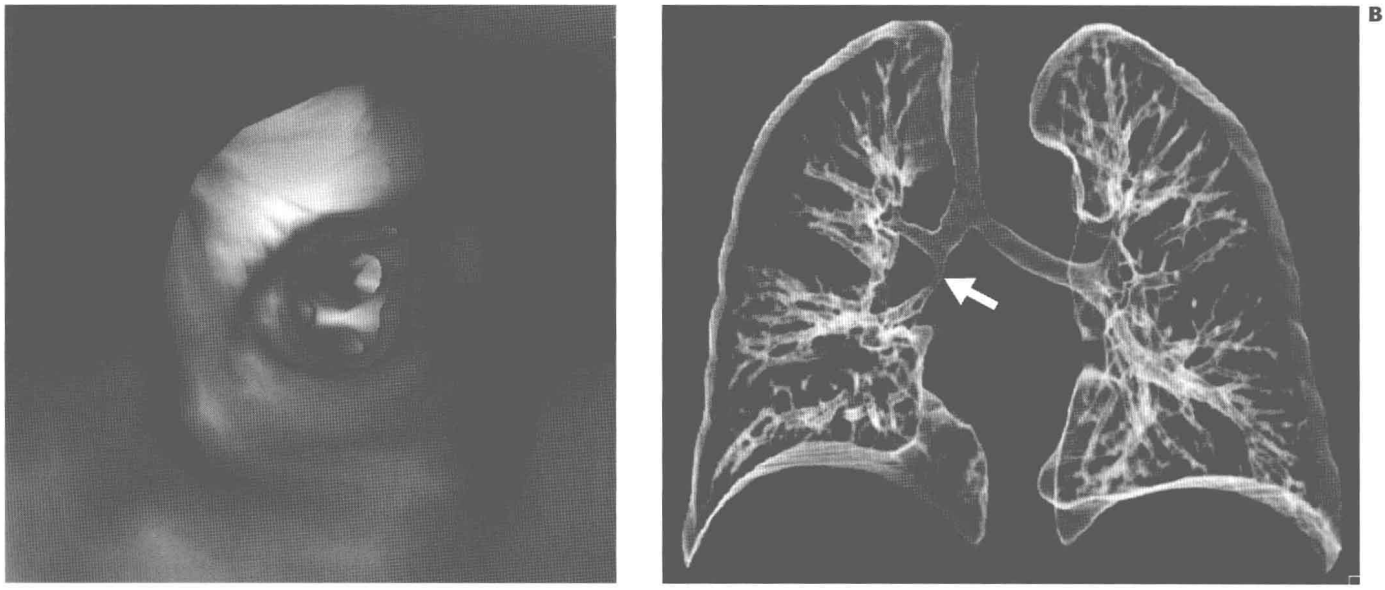

Fig. 1.9 Two examples of three-dimensional reconstructions from continuous volume CT data. **A**, A simulated endobronchial view (virtual bronchoscopy) derived from a routine CT protocol showing two segmental bronchi in the distance. **B**, A volume reduced reconstruction showing narrowing of the bronchus intermedius (arrow) caused by a central lung cancer.

reconstruction of the raw data); extra information is available that is not displayed when the whole body section is viewed at once. Targeted reconstruction is used only when the finest morphologic detail is required.

Sometimes a striking difference is apparent in the characteristics and "look" of the final CT image between different scanners. This is generally 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 such smoothing is thus less necessary. 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 contiguous and high-resolution CT.[83,84] Because of the inherent high contrast between soft tissue structures and aerated lung, it is possible to reduce the photon flux by up to tenfold (for example, a decrease of 200 to 20 mAs) and preserve the diagnostic yield for the detection of pulmonary nodules,[85] a factor which has been exploited in the context of screening for lung cancer. Theoretically, even lower dose protocols may be practicable for the detection of high-contrast lesions. In one report, 6 mA was achieved with the use of aluminum filtration, and there was no significant difference in the detection rate of artificial focal lesions between this ultralow-dose protocol and a low-dose (50 mA) protocol.[86]

Acquisition parameters

Although a single CT section appears 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 various 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" effect is to use thinner sections (Fig. 1.10).

When the whole thorax is examined, contiguous sections are employed. Thinner sections are required to clarify partial volume effects or to study areas of anatomy that are oriented obliquely to the plane of scanning. A specific example of the use of narrow sections (less than 3 mm) to display differential densities (which would otherwise be lost because of the partial volume effect) is the demonstration of small foci of fat or calcium that are sometimes seen within a hamartoma. Thin sections of 1–1.5 mm thickness are used to study the fine morphologic detail of the lung parenchyma (high-resolution CT). Apart from the evaluation of diffuse lung disease, when sampling of a few parts of the lung (traditionally with sections taken at 20 or 30 mm intervals) is adequate, contiguous section scanning is necessary to allow accurate interpretation in most 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 breath hold) and reconstruction interval. Pitch is defined as the distance travelled by the table per gantry revolution divided by the section thickness (collimation). A potential source of confusion arises from the two definitions of pitch used in the context of multidetector CT: it should be borne in mind that either the section thickness or the total z-axis length of the detector array may be used. The latter definition is most frequently quoted in the literature. It also should be emphasized that definitions of

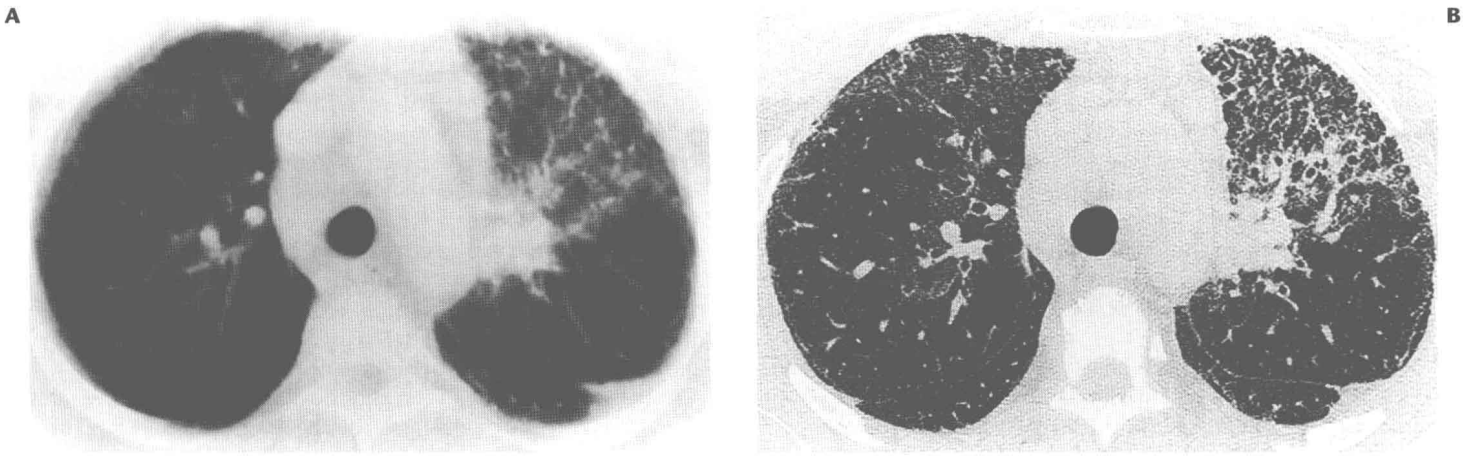

Fig. 1.10 The effect of CT section thickness. **A**, 10 mm section collimation with a standard reconstruction algorithm showing indistinct abnormalities in the lungs of a patient with pulmonary fibrosis and lymphangitis carcinomatosa. **B**, 1.5 mm section with a high spatial resolution reconstruction algorithm at the same anatomic level. Both the normal structures (for example the oblique fissures and airways) and the abnormal features including a reticular pattern and thickened interlobular septa are more clearly seen (courtesy of Dr SR Desai, London).

acquisition parameters and protocols for MDCT may, because of unique detector array designs, be specific to a given manufacturer.

A typical pitch of 1 describes the situation, assuming a gantry revolution in one second, in which the table travels at 10 mm/s with 10 mm collimation. During a 10 second breath hold, 10 cm in longitudinal axis will be covered. If the travel speed is increased to 20 mm/s, the pitch will be increased to 2 and twice the distance will be covered. In general, the useful range of pitch for thoracic work is between 1 and 2.[87] When detection of small pulmonary nodules is the primary aim, a pitch of less than 1.5 is recommended.[88] Conversely, when radiation dose is a major consideration scanning at a higher pitch reduces the radiation burden to the patient.[89] Although the spatial resolution of spiral CT images in the transaxial plane is nearly comparable to conventional CT, there is some image degradation because of broadening of the slice profile, inherent in all spiral CT scanning; this results in additional partial volume averaging in the longitudinal (z-) axis.[90] The faster the table feed, the broader the slice profile. The use of a 180° interpolation algorithm produces a slice profile close to the nominal section thickness, although this causes a slight increase in image noise.[91] Greatly increased z-axis resolution is a feature of MDCT, with isotropic imaging (identical resolution in all planes) being the ultimate goal pursued by manufacturers.

A higher pitch and increased section thickness together enable greater coverage at the expense of increased partial volume effects. However, this can be partly ameliorated by reducing the reconstruction increment, thus producing a larger number of overlapping images.[92,93] The ability retrospectively to reconstruct axial images with considerable overlap by choosing a small reconstruction interval is a major advantage of spiral CT.[94] Nowadays, with the increasing implementation of MDCT, adequate anatomical coverage in a single breath hold is becoming less of an issue.

Intravenous contrast enhancement

The high contrast on CT between vessels and surrounding air in the lung, as well as between vessels and surrounding fat within the mediastinum, means that intravenous contrast enhancement is needed only in specific instances, for example for the detection of emboli within the pulmonary arteries. The exact timing of the injection of contrast media depends most on the time the CT scanner takes to scan the thorax. With fast spiral CT scanning the circulation time of the patient becomes an important factor. However, general guidance about the time of arrival of contrast medium from the antecubital vein to various structures is possible.[95] In normal individuals arrival time in the superior vena cava is 4 seconds, pulmonary arteries 7 seconds, ascending aorta 11 seconds, descending aorta 12 seconds, and inferior vena cava 16 seconds. Occasionally small bubbles of air, introduced with the contrast medium infusion, will be seen, particularly in the main pulmonary artery.[96]

The contrast medium rapidly diffuses out of the vascular space into the extravascular space, so that opacification of the vasculature following a bolus injection quickly declines and nonvascular structures such as lymph nodes steadily increase in density over time. Because of these dynamics, there is a time at which a solid structure may have exactly the same density as an adjacent vessel. The timing and duration of the contrast medium infusion must therefore be taken into account when interpreting a contrast-enhanced CT examination. Rapid scanning protocols with automated injectors improve contrast enhancement of vascular structures at the expense of enhancement of solid lesions because of the rapidity of scanning. With rapid CT scanning it is possible to achieve good opacification of all the thoracic vascular structures with a total dose of less than 100 ml contrast (iodine content of 150–350 mg/ml) at a rate of about 2 ml/s.[97] Some CT scanners generate streak artifact centered on the high-density bolus of contrast, usually as it passes through the superior vena cava. This beam-hardening artifact may be troublesome if it obscures detail in the adjacent pulmonary arteries, particularly in patients being investigated for pulmonary embolism (Fig. 7.16). One solution is to reduce the iodine concentration and use a high volume of dilute contrast at an increased flow rate.[98] A reduction in both the streak artifact and amount of contrast needed can also be achieved by employing a power injector to "push" a smaller volume of contrast with a following bolus of saline solution.[99] One protocol recommended for general thoracic CT scanning is

100 ml of 150 mg iodine/ml (300 mg iodine/ml diluted 50:50) injected at a rate of 2.5 ml/s after a 25 second delay.[100] However, Loubeyre et al have shown that satisfactory enhancement of the hilar vasculature can be obtained with a relatively modest amount of contrast (60 ml of 250 mgI/ml at 3 ml/s).[101] For the examination of inflammatory lesions it may be necessary to delay scanning by at least 30 seconds to allow contrast to diffuse into the extravascular space. Each CT examination must be carefully tailored to the clinical problem; the protocol needed for the evaluation of an empyema with a single-slice CT is very different to that required for the investigation of pulmonary embolism using MDCT.

Consideration must be given to the consequences of accidental extravasation of contrast medium: the flow rate used, within reason, is not predictive of the likelihood of extravasation,[102] nevertheless large volumes (more than 100 ml) introduced into the soft tissues of the forearm by an automated power injector may be associated with severe complications, including compartment syndrome and tissue necrosis; in the event of extravasation of a large volume of contrast, urgent surgical advice should be sought.[103]

Window settings

The average density of each voxel is measured in Hounsfield units (HU); these units have been arbitrarily chosen so that zero is water density and –1000 is air density. The range of Hounsfield units encountered in the thorax is wider than in any other part of the body, ranging from aerated lung (approximately –800 HU) to ribs (+700 HU). The operator uses two variables to select the range of densities to be viewed: window width and window center or level.

The window width determines the number of Hounsfield units to be displayed. Any densities greater than the upper limit of the window width are displayed as white, and any below the limit of the window are displayed as black. Between these two limits the densities are displayed in shades of gray. The median density of the window chosen is the center or level, and this center can be moved higher or lower at will, thus moving the window up or down through the range (Fig. 1.11). The narrower the window width, the greater the contrast discrimination within the window. No single window setting can depict this wide range of densities on a single image. For this reason, thoracic work requires at least two sets of images, usually to demonstrate the lung parenchyma and the soft tissues of the mediastinum. Furthermore, it may be necessary for the operator to adjust the window settings to improve the demonstration of a particular abnormality. Standard window widths and centers for thoracic CT vary between institutions, but some generalizations can be made: for the soft tissues of the mediastinum and chest wall a window width of 300–500 HU and a center of +40 HU are appropriate. For the lungs a wide window of approximately 1500 HU or more at a center of approximately –600 HU is usually satisfactory.[104] For skeletal structures the widest possible window setting at a center of 30 HU is best. Allowing observers to adjust window settings, compared with images at fixed window settings, does not appear to improve performance in terms of identifying fine lung structures or detecting diffuse lung disease.[105]

Window settings have a profound influence on the visibility and apparent size of normal and abnormal structures. The most accurate representation of an object appears to be achieved if the value of the window level is halfway between the density of the structure to be measured and the density of the surrounding tissue.[106,107] For example, the diameter of a pulmonary nodule, measured on soft tissue settings appropriate for the mediastinum, will be grossly underestimated.[108] It is also important to remember that when inappropriate window settings are used, smaller structures (for example, peripheral pulmonary vessels) are proportionately much more affected than larger structures. The optimal window settings for the postprocessed data, for example minimum intensity projection images or 3D volume rendered images, cannot be prescribed and are largely a matter of observer preference.

In the context of HRCT the window settings have a substantial effect on both the appearance of the lungs and the apparent dimensions of, for example, the thickness of bronchial walls[109,110] (Fig. 1.12). Alterations of the window settings may sometimes make detection of parenchymal abnormalities impossible in cases in which there is a subtle increase or decrease in attenuation of the lung parenchyma. Although no absolute window settings can be given because of machine variation and individual preferences, uniformity of window settings from patient to patient will aid consistent interpretation of the lung images. In general, a window level of –500 to –800 Hounsfield units (HU) and a width of between 900 and 1500 HU is usually satisfactory. Modifying the window settings for particular tasks is often desirable: for example in looking for pleuro-parenchymal abnormalities in asbestos-exposed individuals, a wider window of up to 2000 may be useful. Conversely, a narrower window width of approximately 600 HU may usefully emphasize the subtle density differences encountered in patients with emphysema or small airways disease. There does not appear to be any significant diagnostic gain, or otherwise, in allowing observers to adjust window settings freely.[105]

Indications and protocols

There is no single protocol which can be recommended for every clinical eventuality without being prohibitively excessive in terms of radiation dose, time taken, or data acquired. The optimal protocol is one that makes a difference to patient outcome by providing clinically relevant information at the lowest possible radiation dose. There is a constant tension between the desire for a comprehensive examination and the unnecessary exposure of the patient to ionizing radiation.[111] Despite its obvious benefits, MDCT encourages indiscriminate "catch-all" protocols, a problem exacerbated by unfocused clinical requests, for example: "Dyspnea, Rule out bronchiectasis, Nodule on CXR – ?neoplasm". A combination protocol for a general lung examination using MDCT might include contiguous 5 or 3 mm sections from which thin sections (3 or 1.25 mm respectively) can be extracted. Clearly the radiation dose of contiguous 3 mm sections, compared to a tailored protocol appropriate to the scenario above (e.g. HRCT 1 mm sections interspaced 20 mm and contiguous 3 mm sections through the putative nodule), could be regarded as unacceptably high, especially in a young patient.

Attempts to contain and, wherever possible reduce, the radiation dose of a CT examination should be a constant consideration.[111,112] As a practical and simple example, Wildberger et al found that with a Siemens Volume Zoom MDCT (Siemens,