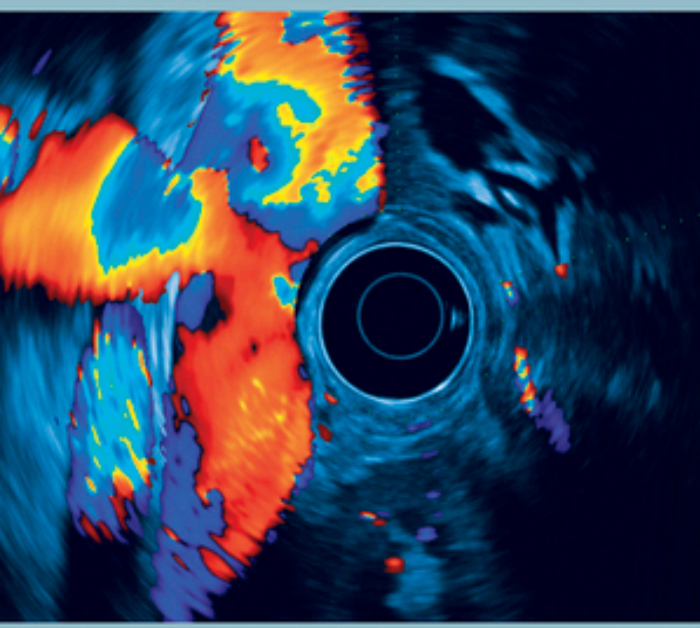
THIRD EDITION

Endoscopic Ultrasonography



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WILEY Blackwell

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Preface

Endoscopic Ultrasonography (EUS) was first conceptualized more than 30 years ago, during the early years of endoscopy, and was developed in an attempt to improve ultrasound imaging of the pancreas. Since the first prototype EUS scopes were released in the early 1980s, EUS has evolved into the "standard of care" for diagnosis and staging of a variety of gastrointestinal (GI) pathologies. In the last few years, it has also become an important therapeutic tool for assisting in complex interventional endoscopic techniques. EUS is now available at community hospitals throughout the word, and is no longer confined to academic medical centers.

Our hope is that *Endoscopic Ultrasonography* improves the training and dissemination of EUS by providing interested GI endoscopists with an authoritative yet practical approach to the role of EUS in the management of specific digestive disorders. This text allows the learner to understand the history of EUS, the fundamentals of ultrasound, and how best to utilize EUS in diagnostic and interventional procedures.

This third edition brings many new and exciting changes and additions to the text, including new chapters on how to learn EUS, elastography, therapeutic EUS, lung cancer, autoimmune pancreatitis, liver disease, biliary access, and pancreatic fluid drainage. We have continued to emphasize a practical, "how-to" approach to learning EUS.

Most of our contributors are either the "first-generation" pioneers of endosonography or the protégés of those pioneers. They have contributed significantly to clinical practice, research, and training in GI endosonography. Their collective experience in applying EUS to the management of GI diseases is unsurpassed. A tremendous amount of effort on the part of each individual author has led to this new third edition. They are the true masters of EUS. We are deeply grateful to them for their outstanding contributions.

This book is meant to introduce the new learner to the field of GI endosonograpy, as well as to update the current endosonographer on recent cutting-edge advances. The chapters combine well-referenced reviews with practical performance advice. We hope you enjoy the third Edition of *Endoscopic Ultrasonography*.

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We give our thanks and love to our parents, Francis and Evelyn Gress and John and Anita Savides, for the guidance, support, and love that created the opportunities we are fortunate to have had in life. We cannot thank enough our wives, Debra Gress and Wendy Buchi, for their unending support, understanding, and sacrifice during the many hours spent completing this text. We dedicate this book to our parents, wives, and especially children, Travis, Erin, Morgan, and Abby Gress, and Michael Savides, for their love, kindness and patience, which sustain us every day.

CHAPTER 1

Endoscopic ultrasonography at the beginning: a personal history

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The first report of endoscopic ultrasonography (EUS), to my knowledge, is that of DiMagno et al., published in 1980 [1]. These investigators described a prototype echoendoscope assembled by attaching a transducer to a duodenoscope. Although images were obtained only in dogs, this work established the feasibility of EUS. As with nearly all seminal advances in endoscopy, EUS was basically an amalgamation of existing technologies. But in 1980, the potential of this hybrid technology was scarcely apparent to anyone – probably including these first endosonographers, who did not expand on their demonstration of the feasibility of EUS.

For practical purposes, the inception of EUS as a clinical entity in the United States can be traced to a meeting I had with Mr. Hiroshi Ichikawa of the Olympus Optical Company. Neither of us can remember the exact date, but it was most likely 1981. Olympus was developing several new technologies, and Hiroshi offered me a choice between EUS and enteroscopy. The only other thing I recollect from that meeting is that, for some unknown reason, I did not ponder the choice very long before I selected EUS, largely because the idea of endosonography seemed especially intriguing; it offered a greater challenge, but also the promise of a much wider range of prospective applications. I certainly gave little thought to – indeed, did not appreciate – the formidable obstacles to the clinical realization of this potential, nor to the investment of time and effort I would need to reach this goal, which was much more distant than I realized. Hiroshi did, in fact, lay emphasis on the obstacles, warning that the instrumentation was in the early stages of development (a euphemism for crude, barely usable). Because of the scope and difficulty of the project, Hiroshi advised that Olympus proposed to work with two investigators in the United States (actually, the western hemisphere), the other being Dr. Charles Lightdale in New York City, as well as a few individuals in other countries. I already knew Charlie, and thought him an excellent choice. As it turned out, this was the beginning of a long and rewarding professional association, for which EUS became the basis. Thus, EUS in the United States began with me and Charlie Lightdale.

Given the technical sophistication of present-day EUS systems, it is important to recognize that during the early years, the viability of endosonography was far from certain. Until about 1985, there was substantial skepticism concerning the future of EUS, even among those of us most closely involved with and committed to its development. The ample tribulations facing the very small cadre of nascent endosonographers became strikingly evident with the arrival of the first EUS system, a prototype in the truest sense. Despite the obvious problems, however, I do not believe that any of us were ever truly discouraged; the best description of our mindset during these formative years might be "doggedly enthusiastic."

I began by writing a simple, all-encompassing protocol that would allow me to use the instrument as an investigational device in patients. The protocol, essentially, had no hypothesis, other than the assertion that EUS was going to be a good thing. It listed almost every possible indication I could conceive, and minimized the risks – which were unknown, in any case – to such a degree that I doubt it would be approved by any institutional research committee today.

The major problems that had to be addressed in the beginning divided into four categories: the technical limitations and deficiencies of the equipment, the development of efficient and safe techniques for the use of the echoendoscope in patients, interpretation of the ultrasound images, and the need to define and establish indications for EUS in clinical practice. More issues, some even more complicated, became evident over time.

The prototype echoendoscope itself was, by modern standards, incredibly cumbersome. The electronic (video) endoscope had not been introduced into clinical practice, so the prototype echoendoscope was a fiberoptic instrument; the optical (endoscopic) component consisted of an ocular lens and focusing ring, coupled to a coherent fiberoptic bundle, with another lens at the distal end of the insertion tube to focus an image on the bundle. The latter provided a limited, 80° field of view, oriented obliquely at an angle of 70° to the insertion tube. Of these two parameters, the narrow field of view was more of a limitation than the oblique orientation, which was not especially problematic for endoscopists accustomed to the side-viewing duodenoscope.

The ultrasound component of early echoendoscopes consisted of a transducer coupled to a rotating acoustic mirror at the distal tip of the insertion tube. The mirror was turned by means of an electric motor within a motor housing situated between a standard design control section and the insertion tube; thus the designation,

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"mechanical, sector-scanning echoendoscope." Because the mirror turned around the long axis of the insertion tube, the ultrasound scanning plane was oriented perpendicular to the insertion tube. In retrospect, this was the best choice, because it seemed to simplify the problems of image interpretation. But this arrangement also had its limitations; mainly that it was unsuitable for guiding a needle to a target. Needle aspiration was, in fact, attempted with the sector-scanning instrument, albeit unsuccessfully, because the width of the tissue within the circular scan was much too narrow.

Unfortunately, the ultrasound imaging sector provided by the first instruments was not a full 360°, but only 180°. To obtain a complete, circumferential sector scan of the surrounding tissue – a circumferential esophageal tumor, for example – it was necessary to rotate the insertion tube 180°, while maintaining the same scanning plane. This was a considerable feat, especially with the instrument deeply inserted, for example in the third part of the duodenum. In truth, it was largely impossible, because any application of torque to the insertion tube invariably altered the scanning plane. This was but one among many difficulties.

Owing to the mechanical components, principally the motor and its housing, the instrument was much heavier than a standard endoscope. I don't think I ever tried to weigh it, but it probably tipped the scale at more than one pound. Because EUS had no established clinical purpose, the first procedures can only be described as exploratory. Consequently, procedure length was determined largely by patient endurance, and with an especially tolerant patient, the weight of the instrument seemingly increased exponentially. After two or three examinations, it was often difficult (and painful) to straighten your left arm.

The combination of optical and acoustical components at the distal end of the insertion tube conferred other penalties, including some potential hazards. The diameter of the insertion tube was 13 mm; that is, substantially greater than that of the upper endoscopes of the time. To make matters worse, the distal end was rigid over a length of 4.5 cm; that is, the distance from the tip to the bending section. Together with the limited field of view, this increased the difficulty of inserting the instrument through the mouth and pharynx and into the esophagus. Although we assumed that the risk of complications with EUS was no greater than that with upper endoscopy, and informed our patients the same, in reality the risk of perforating the pyriform sinus was probably greater – a fact subsequently substantiated. Moreover, attempts at insertion of the large-diameter echoendoscope through a constricting tumor in the esophagus were no doubt associated with an appreciable risk of perforation.

In addition to developing technique for the safe insertion of the echoendoscope, the learning curve for EUS imaging can only be described as long and steep, a line with a slope approaching straight up. According to Yogi Bera, "ninety percent of everything is half mental," and this was definitely true of EUS. The first quandary was the need to uncouple endoscopic imaging from ultrasonography. This related to the need for acoustic coupling; that is, the creation of a suitable interface between the tissue and the transducer (in this case, the acoustic mirror). We discovered in short order that ultrasound images can't be obtained through air. The obvious solution: remove the air. But this proved impractical, for several reasons. The alternative was to interpose water between tissue and "transducer," which could be accomplished in two ways: by placing a balloon over the transducer section of the instrument and filling it with water, or by filling the gut with water. However, it was not simply a matter of choosing between these two options. Depending on the circumstances, including location within the gastrointestinal tract, one or the other was usually a better choice. With the balloon method in particular, the endoscopic view was lost as the balloon was brought into contact with the gut wall, meaning that ultrasound imaging could only proceed by abandoning the endoscopic view. For technical reasons, therefore, EUS imaging was, of necessity, endoscopically blind. Although this decoupling might seem inconsequential today, it was a mental leap of faith in the early days, inasmuch as endoscopic dogma deemed "blind" use of an endoscope hazardous.

Use of the balloon with early-model echoendoscopes was so exasperating that it deserves a digressive paragraph of its own. The latex material that constituted the balloon was not of uniform quality, which made it nearly impossible to place the balloon on the echoendoscope without tearing it. When expanded, the balloon had an asymmetric bulge, and according to the instructions the bulge was to be placed over the transducer on the same side as the optical component; this was never accomplished. Assuming that the balloon could be maneuvered intact into the correct position, it was next necessary to tie it in place with small sutures. The design of the instrument was such that the proximal end of the balloon sometimes occluded the opening of the channel for air insufflation and water irrigation, which would not be evident until it was securely tied in place and tested. Subsequent attempts to nudge the balloon into proper position usually resulted in tearing. Since the objective was to create a water–tissue interface, it was necessary to remove all the air from the balloon (without breaking it). The balloon, if not placed exactly, could occlude the tiny-diameter channel provided for this purpose. Once all of the delicate parameters were attained, and the balloon was in gloriously correct position and functioning properly, the most maddening occurrence was rupture of the ill-fated bag in the middle of an examination, usually at the most inopportune moment. I dealt with some of these frustrations by persuading a gentleman from the biomedical engineering department (designated the "balloon man") to take on the task of balloon placement prior to each procedure.

During the examination, the balloon was filled with water via a Luer lock fitting located between the control section and the motor housing. Unfortunately, this design meant that the attached syringe protruded in perpendicular fashion. Accordingly, as the endosonographer moved his right hand from the control section to the insertion tube, he invariably broke the syringe. In order to fill the balloon, it was necessary to set a small lever on the motor housing to the balloon-filling position, clearly labeled as "B." The other choice was "G," which when selected channeled the water into the gut. Since it was not possible to see this lever, it was advisable to remember which position it was in. Otherwise, the balloon might be filled with water beyond its capacity.

One of the most gratifying aspects of endosonography, readily apparent at the very first examination, was the ability to obtain a structured image of the gut wall. Believe me, all of us knew intuitively and immediately that this was going to be very big. But the interpretation of these images was something else again. There was a natural tendency to assume, to hope, that the five-layer structure corresponded in exact fashion to the actual layers of the gut wall as seen microscopically in a histological section. This betrays a near total ignorance of the principles of ultrasound imaging, and over time it became evident that the physical basis for the endosonographic representation of the bowel wall is much more complex. For reasons unknown to me, the main ultrasound frequency selected for the first EUS systems was 7.5 MHz, a frequency that happens, under

the usual conditions, to render the wall structure of the stomach as five layers. I suspect that this choice of frequency was based on technical considerations, rather than experimental data. In any case, it took some time to work out the actual physical basis for the ultrasound images of the gut wall.

One thing that occurred to me during my first discussion of EUS with Hiroshi Ichikawa, and which probably influenced my choice of EUS as opposed to enteroscopy, was the possibility that EUS might have a positive impact on the problem of pancreatic cancer. By 1980, it was clear that endoscopic retrograde cholangiopancreatography (ERCP) could never alter the natural history of this disease, but perhaps EUS might provide an opportunity, under certain circumstances, for earlier detection and therefore improved survival. In retrospect, this was a worthy but naïve notion. Nevertheless, I resolved to pursue EUS of the pancreas. Charlie Lightdale, on the other hand, took a more sensible and practical path by studying the applications of EUS in staging esophageal cancer. Given the limitations of the first EUS systems, my focus on pancreatic imaging was not the wisest decision.

While my comprehension of the EUS image of the gut wall was next to zero, this knowledge was encyclopedic by comparison with my understanding of EUS of the pancreas. In truth, the only thing I could identify with certitude was a gallstone, and only if it was over 1 cm in diameter and solidly calcified. After a while, optimism becomes a poor substitute for know-how, and it was soon obvious that the only way to move forward was to seek the advice of a radiologist with expertise in ultrasonography. Many of the first endosonographers adopted a similar approach. And so, a radiologist by the name of Craig George came to my assistance. Our idea was that Craig would look over my shoulder during the EUS procedure and essentially interpret the images. By this time, we had a second-generation prototype EUS system. In contrast to the first prototype, the second system included an extremely bulky image processor with a tiny display screen, probably no more than 8 inches on the diagonal. Moreover, the quality of the image was poor, which made it necessary to get close to the screen to see anything. Furthermore, the screen was placed in the box such that it was only about 4 feet above the floor. So, Craig sat on a low stool in front of the box. But all of these limitations were inconsequential to me because Craig is a big guy with a correspondingly large head; most of the time the only thing I could see was the back of it. Somehow, we evolved a set of hand signals to deal with this problem. It worked like this: if Craig (face pressed to the screen) saw something he recognized, he would make certain motions with his hand, either the left or the right depending on the direction he wanted me to move the transducer, in an effort to obtain the best possible image (I always think of Craig whenever I watch a jet plane being guided to its parking place by the guy with the long, orange flashlights). When he got the image he wanted, Craig would hit the "freeze" button, quickly move his head out of the way so I could see it, and then place a camera in front of the screen to obtain a photograph (the permanent image in those days).

Although this arrangement was cumbersome, I learned most of what I know about pancreatic imaging, and the principles of ultrasonography, from Craig George. After about 6 months, our partnership gradually dissolved, partly because it was difficult to coordinate our schedules, but mostly because I had acquired, so I thought, enough knowledge to proceed on my own.

Until June 1982, the struggle to develop EUS was a lonely one; only a handful of endoscopists had any practical experience with EUS, and all were working essentially alone. This changed that June, when Olympus sponsored the first "International Workshop on Endoscopic Ultrasonography" at the Grand Hotel in Stockholm, Sweden – a time and venue selected to coincide with the World Congress of Gastroenterology. We met in a very small room, as there were, according to my notes, only about 15 active participants, including two invited guests with expertise in areas of digestive ultrasonography other than EUS, and excluding about a half dozen representatives from Olympus.

Keichi Kawai (Kyoto, Japan), who organized the meeting, asked me to speak on "Arrangement of Endoscopic Ultrasonography." I never did discover exactly what my assigned topic entailed. Nevertheless, compared to the many EUS meetings in which I participated in subsequent years, this first gathering was by far the most important. For, by the time of the meeting, each participant had discovered many things about EUS, but none had a complete picture, whether of its limitations or of its true potential. Thus, there was a remarkable and exhilarating exchange of information and ideas that, in retrospect, amounted by aggregation to a significant advance. I led a long discussion on EUS of the pancreas that solidified the concept of stationed withdrawal of the echoendoscope from the duodenum. Essentially, we made a list of the organs and structures that should be imaged at each station. But, most importantly, I think each of the dozen participants left the meeting with a revitalized sense of purpose, as well as a stronger sense of confidence in the future of EUS.

Another aspect of EUS that was clarified by the 1982 meeting was the incredible value of cooperation in the effort to establish EUS as a clinically useful technology. In many ways, the meeting revealed more about what we didn't know than what we did, and it showed how much had to be done before EUS could be considered clinically relevant. Shortly thereafter, and I think in response to the lessons learned at the meeting, Mr. Mark Donohue of Olympus asked me to help organize a small group of investigators that would meet two or three times each year. Our purpose was to grapple collectively with the problems of EUS and, in general, find ways to advance its development. In addition to myself, the original membership included Charlie Lightdale and Drs. H. Worth Boyce and Lok Tio. Over the eight or so years of its existence, the membership changed somewhat, but it was always strictly limited to no more than six (usually five). Together with two or three people from Olympus, the total number attending each meeting was never more than eight or nine. Naturally, when the existence of this group became known, albeit not widely, Olympus was besieged by individuals who felt they had the qualifications for membership. But, to the credit of Olympus, Mr. Donohue resisted all requests, in order to preserve the small-group dynamic. Because we could never dream up a better name, we called ourselves the "EUS Users Group."

I used to make an agenda for each "Users" meeting, based on input from the members and from Olympus. In retrospect, these lists of topics for discussion outline much of the developmental history of EUS from about 1982 to 1989. The subject matter divided into two major areas: technical development and the application of the technology to clinical practice, and training. During the earliest years, we did not recognize that there would be major issues and problems relating to the training of other endoscopists in EUS, or a need for the broader dissemination of information about EUS to the medical community at large. But as interest in EUS increased, it became glaringly evident that training constituted a most formidable problem, all the more so inasmuch as clinical relevance would never be achieved if EUS were performed by a small number of experts. This issue was further compounded by the high cost of the equipment (relative to that of standard endoscopes)

and the absence of reimbursement. In those days, furthermore, echoendoscopes were fragile, as well as expensive. The need for frequent maintenance and repair substantially increased the cost of operation. In the hands of an inexperienced operator, this fragility frequently pushed repair costs well beyond those normally anticipated by an endoscopy unit. All of these factors constituted a significant "cost barrier" to involvement with EUS.

There was a certain division within the "Users Group" as to the best approach to the problem of training. We were unanimous concerning the value of didactic teaching, and to this end we organized a number of short symposia. However, we fully recognized that this was no substitute for so-called "hands-on" instruction. With respect to the latter, one viewpoint held that short periods of training, ranging from a few days for an accomplished endoscopist to 6 months for the less experienced, would be adequate to "get started." I and some others felt that a "quick and dirty" approach was doomed to failure; we advocated much more formal and prolonged training. The caveat of this approach, however, was that EUS might never become established. As late as 1988, the programs with the capability for training numbered only five; that is, the members of the group. Even if we trained 10 endosonographers per year, it would take many years before EUS became widely available. In retrospect, I think I was right: it took better training and a lot more time than anyone expected.

It was fortunate that EUS was introduced during the decade of the 1980s, a period when endoscopists were under less pressure to be ultra-efficient and financially productive. The commitment to screening colonoscopy, for example, had not yet arisen, even as a concept. Had the introduction of EUS been attempted 10 years later, the probability that it would become an established procedure would have been substantially reduced. In those earlier times, gastrointestinal endoscopy was less of a mass-produced commodity, and not something akin to a chest radiograph or complete blood count. It is true that we were somewhat mesmerized by technology, but this was always integral to the overriding desire to improve patient care.

The establishment of EUS as a clinical procedural entity stands as a tribute to the perseverance of a relatively small group of people, as well as to the resolve of the Olympus company. Although this was not generally known, EUS also constituted a substantial cost barrier for the company. I was never privy to the actual financial data, but Mr. Donohue once told me that EUS was a financial loss for more than a decade. That any company would invest so much time and talent for so long, despite an uncertain prospect of financial gain, is remarkable. There is a story, which admittedly be apocryphal, that Mr. Ichizo Kawahara, then the director of the Medical Instrument Division of Olympus, was once asked why the company persisted in its efforts to develop EUS despite the obstacles and the uncertain chance for success. He is said to have replied, "Because the doctors want it." This, I believe, also reveals the different nature of those times.

I think I became fully convinced that EUS was here to stay with the introduction of the Olympus/Aloka UM2 system, which occurred around 1986. The GF-UM2 echoendoscope was still a fiberoptic instrument, but the EU-M2 display unit was markedly improved. In particular, it offered a 360-sector display, a gigantic improvement with respect to pancreatic imaging. This was followed by a gradual but steady flow of technical improvements. This, together with the continuing addition of more and better data, solidified a lasting place for EUS in clinical practice. It took a lot longer than I had imagined, but it was gratifying to have played a part.

Reference

1 DiMagno EP, Buxton JL, Regan PT, et al. Ultrasonic endoscope. Lancet 1980;I:629–631.

CHAPTER 2

Basic principles and fundamentals of EUS imaging

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An understanding of the fundamental mechanisms of ultrasound (US) is useful to both the new and the experienced endosonographer. It is not necessary to be a physicist or an engineer to appreciate some basic principles of US imaging and Doppler US. These principles can guide the endosonographer in both obtaining the best representation of a tissue structure with endoscopic ultrasounography (EUS) and interpreting the images thus produced. Knowing these fundamental concepts also aids in the recognition and avoidance of artifacts.

In this chapter, the principles of US imaging will be reviewed. An emphasis will be placed on their practical application to endosonography, rather than on the derivation of formulas and equations, which will soon be forgotten.

How US images are made

Sound is mechanical energy that is transmitted as a wave through a fluid or solid medium [1, 2]. Unlike electromagnetic waves (e.g., radio, light, and X-ray), sound waves cannot be transmitted through a vacuum. The energy must be transmitted via its impact on the molecules of the transmitting medium.

The periodicity or frequency of sound waves per unit of time varies widely and is measured in the number of cycles of the wave that are formed in 1 second, termed a *hertz (Hz)*. Each wave cycle has both a positive and a negative pressure component. US is higher in frequency than can be heard by the human ear (Figure 2.1). The frequencies of waves commonly used in medical imaging are between 3.5 and 20 million Hz, usually abbreviated as 3.5–20 MHz. Even higher-frequency waves can be used in microscopy to define tissue ultrastructure.

The high-frequency sound waves used in imaging have some interesting properties that affect how they are used. Unlike lower-frequency audible sound waves, which travel well through air, high-frequency sound is more readily absorbed and attenuated by air, and is strongly reflected at the boundary between tissue and air. This is why gas-filled lungs and bowel limit the use of transcutaneous US in imaging of mediastinal and retroperitoneal structures.

How US waves are made

Sound waves are made by applying an oscillating pressure to a medium. A radio speaker vibrates at variable speeds or frequencies to create sound waves in air, which we hear as sound. Higher-frequency US waves are made by crystals that vibrate to transmit a US pulse within a body fluid or tissue. These crystals are made from a special ceramic material, because this can be made to vibrate at a high frequency when a high-frequency alternating polarity charge is applied to it. This property is termed *piezoelectric* and is also responsible for the crystal's ability to detect sound waves returning from the tissue and convert them back into an electrical signal.

US transducers are composed of either one large crystal or, more commonly, multiple crystals aligned in an array. These transducers change an electrical signal to a sound wave and also receive the reflected sound wave back from the tissue. US transducers typically emit a series of waves or a pulse, and then stop transmitting while they wait to detect the returning echo.

What happens when US waves encounter tissue

US waves propagate through tissue at a speed that is determined by the physical properties of the tissue [3, 4]. The speed of transmission is largely determined by the stiffness of the tissue: the stiffer it is, the faster the speed. For soft tissue, the variation in speed is only approximately 10%, ranging from 1460 m/s in fat to 1630 m/s in muscle [5–7].

US waves are reflected back to the transducer when the sound wave encounters a tissue that is difficult to pass through. For example, water easily transmits US, but air and bone do not. A sound wave that travels through a water-filled structure like the gallbladder is likely to reach the opposite gallbladder wall unless it encounters a gallstone, which will it back to the transducer. Other solid tissues reflect sound waves to a variable extent, depending on the tissue properties. Fat and collagen are more reflective to US than are muscle and lean solid organs. Sound waves are also reflected when they encounter a boundary or interface between two tissues with different acoustical properties (see next section).

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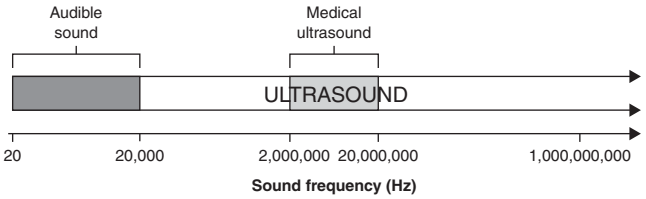

Figure 2.1 Frequencies of audible sound and US.

How images are made from reflected US waves

Sound waves that are reflected by tissue components back to the transducer are detected by the same piezoelectric crystals that created them. These crystals then translate the waves back into electrical signals for processing into an image.

The transducer detects the returning echo as a function of the time that passed from when the sound pulse was emitted. The amount of time it takes for an echo to return is a function of the speed of sound in the tissue and the distance from the transducer of the part of the tissue from which the sound wave is being returned. Because the speed of sound in lean tissue varies only by approximately 10%, the time between transmission and return of an echo is a good marker for the distance the sound wave has traveled. Thus, for medical imaging, distance or the location of a reflector within a tissue can be approximated by the delay observed in the return of a US pulse.

The returning waves or echoes can be displayed in a number of ways or modes. The simplest display plots the intensity or amplitude of echoes according to the time at which they are detected. This is termed A-mode and is infrequently used for medical imaging. If the amplitude of the returning signals is displayed as the brightness of a dot on the image, a B-mode image is created. If the transducer is moved across the tissue or if the transducer contains numerous crystals, a two-dimensional image is created out of the dots, which reflect echo amplitude; one dimension is the location or depth of the reflector causing the echo, while the other is the span of tissue being imaged (Figure 2.2).

The precise time at which a returning echo is detected is also a function of the orientation of the target tissue and the transducer. A more accurate representation of tissue structure is obtained when the US wave propagates in a direction that is perpendicular to the target. The reflected wave is then perpendicular to the transducer as well. If the US wave encounters the target from another angle or tangentially, then the returning wave is detected later and is thus displayed on the image at a distance that overestimates its actual position (see section on Imaging Artifacts).

How transducer properties affect the image

US frequency and axial resolution

When high US frequencies are used, more waves can be transmitted per unit of time and the duration of the pulse of US energy can be proportionately reduced. This allows the US transducer to receive returning echoes more often. The result is a better ability to discriminate between two points in the target tissue that are within the direction of the US beam. This distance between distinguishable points in the direction of the US beam is termed "axial" or "range" resolution (Figure 2.3). In general, the higher the US frequency, the better the axial resolution. Most endoscopic US systems have axial resolutions that are approximately 0.2 mm. However, tissue penetration is also reduced with higher US frequencies (Table 2.1).

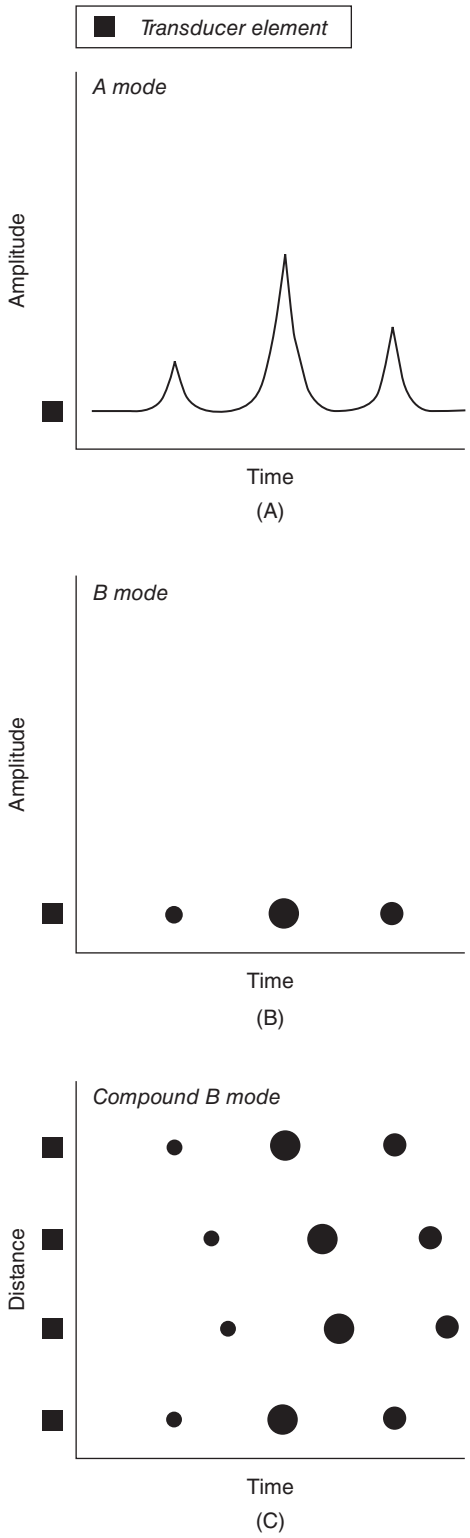

Figure 2.2 The basic types of US image. (A) An A-mode image plots the amplitude of a returning echo versus the time at which it returns relative to the transmitted US wave. Because the velocity of sound through soft tissue is relatively constant, the time taken for an echo to return can be converted into the distance or depth within the tissue at which the echo originated. (B) A B-mode image displays the amplitude of an echo as the brightness of a dot. (C) When multiple transducers are used or when a single transducer is moved over an area, the multiple single-line B-mode images can be converted into a rectilinear or compound scan.

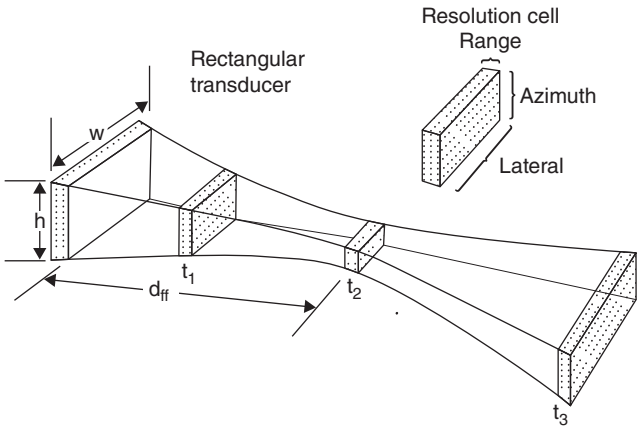

Figure 2.3 The resolution in three dimensions (resolution cell) for a pulse of US energy as it propagates from a rectangular-shaped transducer of defined width (w) and height (h). The duration of the pulse, defining the axial or range resolution, stays the same as the wave propagates and is illustrated at three times: t_1, t_2, and t_3. Changes in the beam pattern produce changes in the lateral and azimuthal resolutions at the three time points. The near–far field transition point (d_{ff}) is the point with the smallest-resolution cell (in this case, illustrated at time t_2) and offers the best overall resolution. Source: Kimmey MB, Martin RW 1992 [4]. Fundamentals of endosonography. Gastrointest Endosc Clin North Am 2:560, WB Saunders. Reproduced with permission of Elsevier.

Table 2.1 Effect of US frequency on axial resolution and tissue penetration.

US frequency (MHz)	Axial resolution (mm)	Tissue penetration (cm)
5	0.8	8
10	0.4	4
20	0.2	2

Transducer size and lateral resolution

The lateral resolution makes it possible to distinguish between two points in the lateral dimension (see Figure 2.3). The magnitude of this resolution is dependent on the diameter of the transducer. In general, larger transducers have poorer lateral resolution. The lateral resolution is not constant, but varies according to the distance of the target reflector from the transducer. The location of the best lateral resolution is often referred to as the focal zone of the transducer, and is the point at which the beam is focused and the lateral resolution is optimized. With most US endoscopes, this distance is between 2 and 3 cm from the transducer.

The frequency of a US transducer also affects the lateral resolution. Small-diameter transducers used on catheter probes are especially vulnerable to this effect. With other variables being equal, higher-frequency small-diameter transducers have a narrower focal zone over a broader distance from the transducer than do lower-frequency transducers of the same diameter (Figure 2.4). This is the primary reason why catheter probes are made with higher-frequency (12–20 MHz) transducers.

Attenuation and tissue penetration

"Attenuation" refers to the loss of strength of the US beam over time or distance traveled. The degree of attenuation is dependent on the properties of both the US transducer and the tissue, but the most important factor is the US frequency. Higher US frequencies are maximally attenuated and hence do not penetrate as far into the tissue. Higher frequencies are also attenuated to a greater degree by specific tissue components, such as fat. For example, a lipoma within the gastrointestinal (GI) wall can attenuate a 12 or 20 MHz US beam so effectively that no US energy reaches the deep aspect of the lesion (Figure 2.5). The entire lipoma therefore may not be represented on the US image. In such situations, a lower-frequency US transducer might be preferable.

Since all tissue attenuates US to some degree, returning echoes from deeper tissue structures will have lower amplitude than those from more superficial structures. This is due to attenuation of both the transmitting US wave and the returning echo. Medical US imaging systems compensate for this effect by amplifying the echoes that return to the transducer later (Figure 2.6). Amplification of these echoes from deeper tissue structures is called time gain compensation (TGC). TGC can be controlled by the sonographer by changing settings on the US processor. The goal is to make similar tissue have the same US appearance, irrespective of location within the tissue.

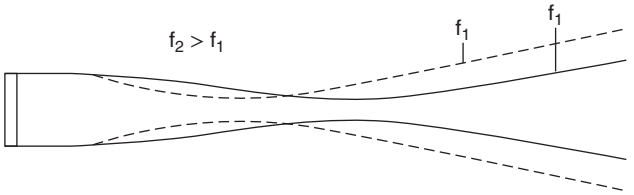

Figure 2.4 Effects of US frequency (f) on the beam pattern of a transducer. For the same size transducer, a beam (solid lines) with a higher US frequency (f_2) produces a near–far field transition point that is further from the transducer and causes a narrower beam width in the far field. A beam (dashed lines) with a "lower frequency" (f_1) is illustrated for comparison. Source: Kimmey MB, Martin RW 1992 [4]. Fundamentals of endosonography. Gastrointest Endosc Clin North Am 2:561, WB Saunders. Reproduced with permission of Elsevier.

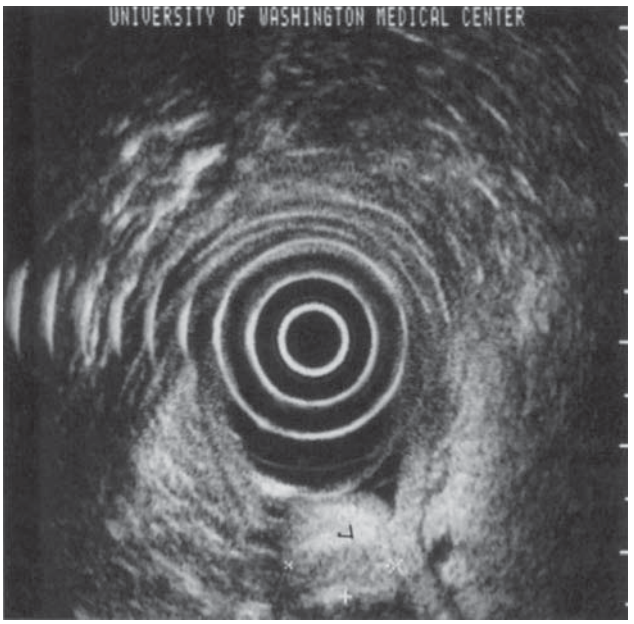

Figure 2.5 A duodenal lipoma (L) strongly attenuates the 12.5 MHz US beam, producing an acoustic shadow (arrows) in the tissue deep to the lipoma.

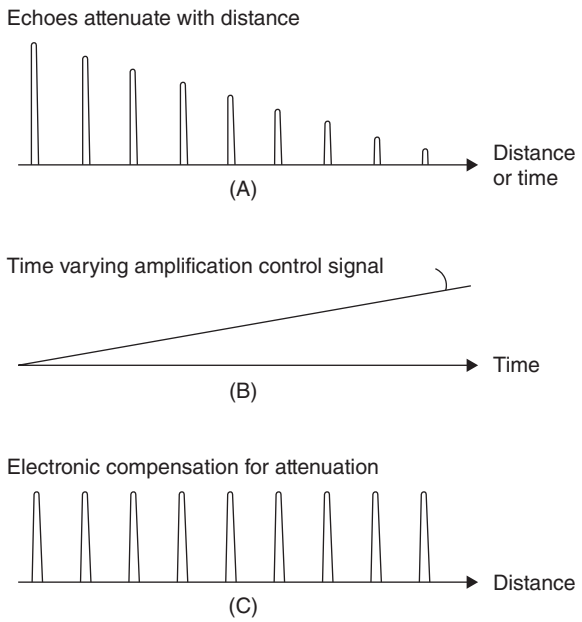

Figure 2.6 Time-varying gain (TVG) compensation. The vertical axis represents the amplitude of the received echoes (A, C) and the control signal (B). (A) US echoes with the same amplitude at the reflection site are received by the transducer as lower-amplitude signals according to how far the reflector is from the transducer, because of attenuation of both the transmitted and the reflected US waves. (B) The received echo can be electronically amplified according to when it is received. As shown by the linear increase, echoes from similar reflectors have the same amplitude at all distances from the transducer. Source: Kimmey MB, Martin RW 1992 [4]. Fundamentals of endosonography. Gastrointest Endosc Clin North Am 2:563, WB Saunders. Reproduced with permission of Elsevier.

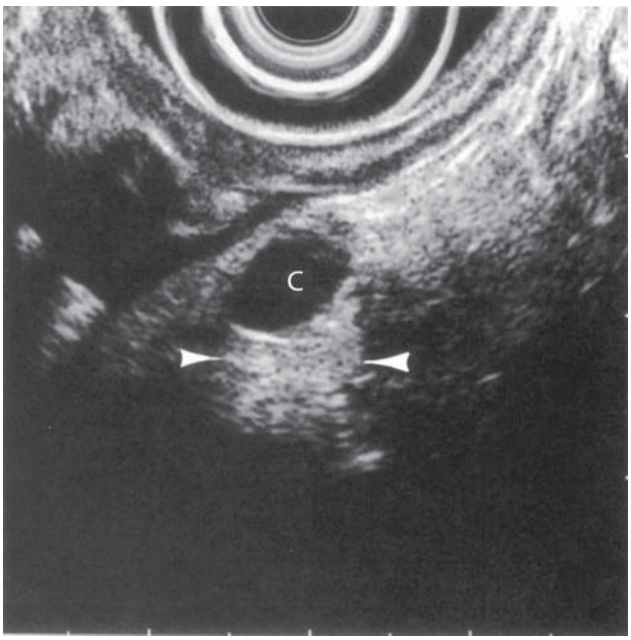

Figure 2.7 Fluid within this small pancreatic cyst (C) does not reflect much of the US beam, leading to more echoes being seen in the tissue deep to the cyst (between arrows). This is the through-transmission artifact.

Knowledge of attenuation can also be useful in image interpretation. Most bodily fluids (blood, urine, and bile) attenuate a US beam very little. Thus, when imaging a fluid-filled structure, more US energy is transmitted to the tissue deep to the structure than to the tissue deep to the adjacent solid tissue. There are then more returning echoes from the tissue deep to the fluid-containing structure, making this tissue brighter on the image. This through-transmission enhancement can be used to help distinguish between fluid-filled and solid structures. For example, images of a cyst will show brighter echoes in the area of tissue deep to the cyst (Figure 2.7).

How tissue properties affect images: the GI wall

The composite image of a tissue depends on the properties of the tissue and on the US transducer and system used. US imaging of the GI tract wall is a good example of how these various factors interact.

Frequency dependence

Early reports of imaging of the GI wall with transcutaneous US transducers described a three-layered structure. The layers represented luminal contents (echo rich), the wall itself (echo poor), and the surrounding tissues (echo rich). The axial resolution of these low-frequency (3–5 MHz) systems was too poor to detect the different components of the wall itself. With the development of endoscopic US systems with higher frequency (7.5–12 MHz) and better-resolution transducers, the GI wall was usually imaged as a five-layered structure, due to the different US properties of the mucosa, submucosa, and muscularis propria [8]. Most recently, 20 MHz catheter-based EUS systems routinely image the GI wall as a seven- or nine-layer structure, due to their better resolution, which allows the muscularis mucosae and the intermuscular connective tissue of the muscularis propria to be distinguished [9, 10].

Higher US frequencies also produce brighter echoes from specular reflectors (see next section). This also contributes to the improved resolution seen with higher-frequency US systems.

Specular and nonspecular reflectors

There are two types of tissue reflector that are sources of echoes on US images. These are termed "nonspecular" and "specular" reflectors. Echoes from nonspecular reflectors are produced by tissue components that scatter the US wave. Echoes from specular reflectors are produced when the US wave encounters two adjacent tissues with different acoustical properties. The US image is a composite of echoes from both types of reflector. For example, the US image of a mixture of oil and water is homogeneous and echo-rich. Echoes are reflected from nonspecular reflectors caused by the small oil droplets mixed in the water. After separation of the oil and water, however, only a thin echoic line is seen from the specular reflector at the interface between the oil and the water.

Nonspecular reflectors (scatterers)

Fat and collagen are the most reflective tissue components of the GI wall. These tissue components are responsible for the bright layer seen in the center of the GI wall on EUS images. The submucosa is a dense network of collagen fibrils that provide structural support and allow for sliding of the overlying mucosa during motility. There is sometimes fat present in the submucosa, as well. The other bright layer on EUS images of the bowel wall comes from tissue just deep to the muscularis propria. In most areas of the body, this is from fat

in the subserosa. In the esophagus, which is not covered by serosa, the bright layer is caused by fat in the mediastinum. In the rectum, fat and collagen in the pelvis create the bright layer.

Specular reflectors (interface echoes)

Early interpretations of US images of the GI wall associated the echo-poor second layer with the muscularis mucosae. However, careful measurements later demonstrated that this US layer was much too thick to be the muscularis mucosae [8]. Further measurements also suggested that the central echoic layer was too thick to be the submucosa and the deep, echo-poor (or fourth) layer was too thin to represent the muscularis propria. These observations were reconciled by considering the contribution to the image of specular reflectors produced at the interface between tissue layers of the bowel wall [8].

The thickness of an interface echo is determined by the pulse length or axial resolution of the US transducer. The beginning of an interface echo corresponds with the location of the interface, so that the thickness of the interface echo itself will co-locate with the most superficial aspect of the deeper tissue layer. Thus, an interface echo will add thickness to a more superficial echo-rich layer like the submucosa, but subtract from the apparent thickness of a deeper echo-poor layer like the muscularis propria. When layer measurements are corrected for the presence of interface echoes, an accurate interpretation of the images is possible (Figure 2.8).

These principles can also be applied to the interpretation of the seven- or nine-layered images of the GI wall that are obtained with higher US frequencies. Better axial resolution and thinner interface echoes allow the muscularis mucosae to be visualized as a thin echo-poor layer superficial to the submucosa. The interface echo between the lamina propria and the muscularis mucosae divides the mucosa into four layers: an interface echo at the mucosal surface, the lamina propria, an interface echo between the lamina propria and muscularis mucosae, and the remainder of the muscularis mucosae that was not obscured by the interface echo [9, 10]. The additional three layers in a nine-layered GI wall are caused by the division of the muscularis propria into inner circular and outer longitudinal components by a line of nonspecular echoes from a thin layer of connective tissue (Figure 2.9).

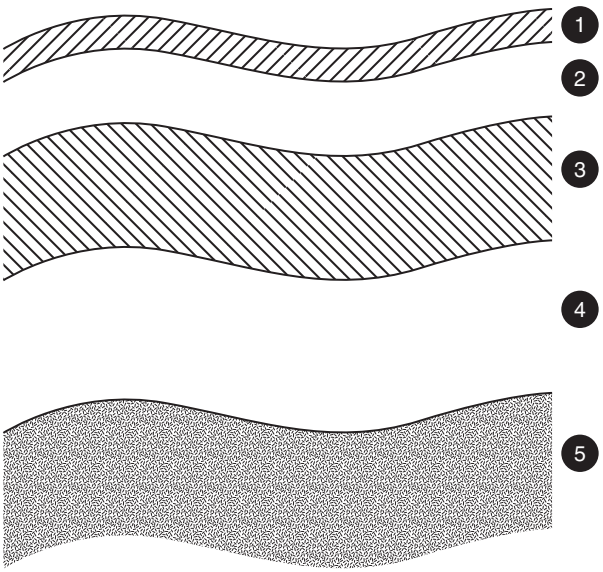

Figure 2.8 The five layers of the normal GI wall, as imaged with most endoscopic ultrasound equipment. From the mucosal surface at the top, layer 1 is produced by the interface between luminal fluid and the mucosal surface. Layer 2 is from the remainder of the mucosa. Layer 3 is from the submucosa and its interface with the muscularis propria. Layer 4 is the remainder of the muscularis propria. Layer 5 is from subserosal fat and connective tissue.

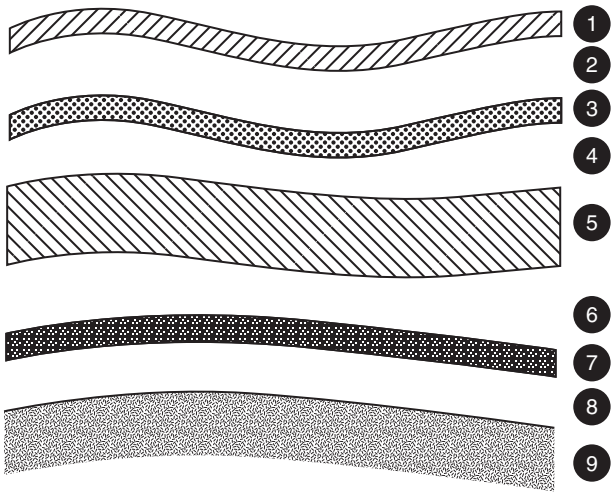

Figure 2.9 High-frequency US transducers may image the GI wall as a nine-layered structure. From the mucosal surface at the top, layer 1 is produced by the interface between luminal fluid and the mucosal surface. Layer 2 is from the remainder of the lamina propria. Layer 3 is from the interface of the lamina propria and the muscularis mucosae. The remainder of the muscularis mucosae is visualized as a hypoechoic fourth layer only if the muscularis mucosae is thicker than the pulse length or axial resolution of the US transducer used. Layer 5 is from the submucosa and its interface with the muscularis propria. Layer 6 is the remainder of the inner circular component of the muscularis propria. The intermuscular connective tissue produces a thin echoic layer 7. The outer longitudinal component of the muscularis propria is responsible for layer 8. Layer 9 is from subserosal fat and connective tissue.

Detection of tissue movement: doppler imaging

When a US wave encounters a moving object, its US frequency is shifted. This frequency change is termed the *Doppler shift*, and the use of this principle in detecting tissue movement is called *Doppler imaging*. Movement of red blood cells within blood vessels is the most common application of Doppler imaging. The direction of the frequency shift can also be used to determine the direction of the movement (i.e., toward or away from the transducer).

A few special principles of Doppler physics need to be recalled to optimize use of this technique. First, the Doppler frequency shift is maximal when the US wave encounters the moving objects at a tangential rather than a perpendicular angle. This is contrary to the principle of US imaging that tissue structure is reproduced most faithfully by a US wave that is perpendicular to the tissue. It is therefore often necessary to move the transducer in real time to simultaneously obtain optimal imaging and Doppler information.

There are two basic methods for performing Doppler measurements: *pulsed Doppler* and *continuous-wave Doppler*. Continuous-wave Doppler requires two transducers: a transmitting transducer and a receiving transducer. The transmitting transducer delivers a continuous fixed-frequency US wave into the tissue. The receiving transducer then receives the signal. If there is movement

in the tissue, the transmitted and received signals will differ, and when the two signals are summed together, the result will be a waveform that contains a *beat frequency* that is equivalent to the Doppler shift frequency. Continuous-wave Doppler is unable to give information regarding the location at which the Doppler shift is detected; therefore, pulsed Doppler was developed to obtain depth information regarding where the motion causing the Doppler shift is occurring. In pulsed Doppler, a single transducer is used to send a US pulse intermittently, so that detection of the returning Doppler wave is not limited by further transmitting waves. This leads to a more reliable detection of the depth of the moving object. For example, pulsed-wave Doppler probes have been shown to reliably detect the location of blood vessels in the GI wall [11].

Doppler information can be displayed in a number of ways. The Doppler shift of *moving* blood is approximately 15 000 Hz. Because this is within the range of human hearing, the signal can be amplified into an audible signal. The Doppler signal can also be superimposed on a B-mode scan so that the location of the moving objects can be determined by looking at the B-mode image. This is called *duplex scanning* and is commonly used in EUS. The presence of a Doppler signal is good evidence that a cystic anechoic structure on B-mode imaging is a blood vessel. The direction of the Doppler shift can also be codified with color, in a technique called *color Doppler*. Red is commonly used to represent flow toward the transducer, and blue to represent flow away from the transducer. *Power Doppler* is the most recent advancement in Doppler US imaging and is the most sensitive method for detecting blood flow. For power Doppler imaging, pulsed Doppler is used to obtain the Doppler signal. However, power Doppler evaluates the strength of the Doppler signal and discards any information regarding the velocity or direction of motion.

New techniques in EUS imaging

Contrast-enhanced EUS imaging

Intravenous injection of a US contrast agent (UCA) – gas-filled microbubbles that are 2–5 μm in diameter – results in enhancement of vascular structures on US imaging if an appropriate imaging technique and processing are used. This is a relatively well developed imaging technology for cardiac imaging and transabdominal applications; however, the technology for EUS imaging is still in development [12]. The use of UCAs has enhanced the diagnostic capabilities of US imaging by improving the ability to image smaller-caliber blood vessels, improving identification of tumors, and enhancing visualization of the cardiac wall [13–15]. Potential applications in EUS include evaluation of vascular invasion for tumor staging, differentiating benign and malignant lymph nodes [16], discriminating between focal pancreatitis and pancreatic carcinoma [17, 18], and localizing vascular tumors such as insulinomas [19].

Elastography

Elastrography is a method used to assess the stiffness of tissue in response to compression, by comparing the backscattered US signal from tissue in a compressed and a noncompressed state [20]. This method is being evaluated for use in diagnosing disease processes that cause the stiffness of tissue to change, such as cirrhosis, inflammation, and malignancy. It is analogous to the physical examination technique of palpation. For example, malignant tumors are often firm when palpated on physical examination. Elastography is a form of palpation that uses US to detect regions that have different stiffness relative to the surrounding tissue. With external compression, the US signal that is received from the region of interest will be different than the signal received when the region of interest is not compressed. The two signals are compared using image processing algorithms to produce an *elastrogram*. For external imaging applications, the US transducer can be used to apply compression to the region of interest, typically in a repetitive motion (compression–relaxation). For endoscopic applications, it can be difficult to apply compression to a region of interest using the EUS transducer; therefore, the compressions to the region of interest can be made by vascular pulsation or respiratory motion. EUS elastography should improve the diagnostic capabilities of EUS and help to improve localization of lesions and diagnostic yields on biopsy [21].

Imaging artifacts

There are a number of artifacts that should be recognized when performing EUS imaging. Artifacts are echoes seen on an image that do not reliably reproduce the actual tissue structure. Failure to recognize artifacts can lead to image misinterpretation and errors in *patient management*. This section will highlight some common artifacts and discuss how to recognize or, if possible, avoid them.

Reverberation artifacts

Strong echoes are produced when a US wave encounters solid nontissue objects. The most common example of this is reverberation of the US beam from the casing of the transducer. This produces a characteristic series of echoes at equal intervals, radiating out from the transducer – the ring artifact (Figure 2.10). It is seen more commonly with the radial scanning echoendoscope than with the curvilinear array (CLA) instrument, and in some situations can interfere with the near-field image. Reducing overall and near-field gain helps to minimize this artifact. Moving the transducer away from the area

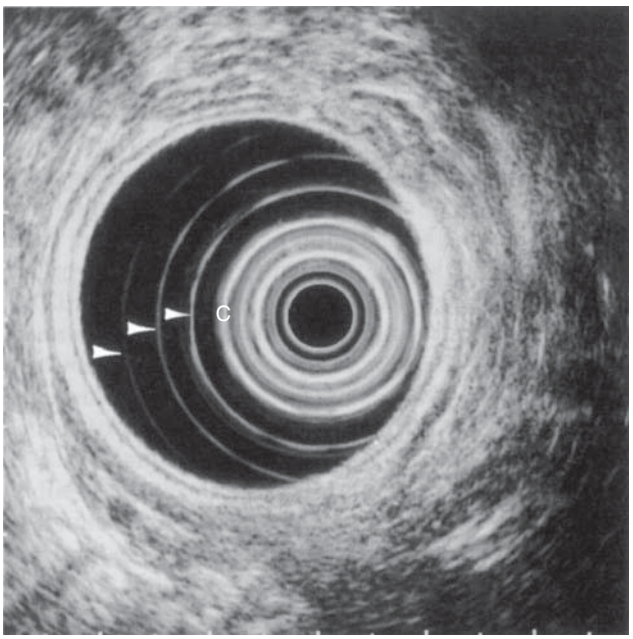

Figure 2.10 The plastic casing (C) around the US transducer produces a strong reverberation of the US beam between the transducer and the casing. This results in a series of circular rings (arrows) of equal spacing and diminishing amplitude around the transducer.

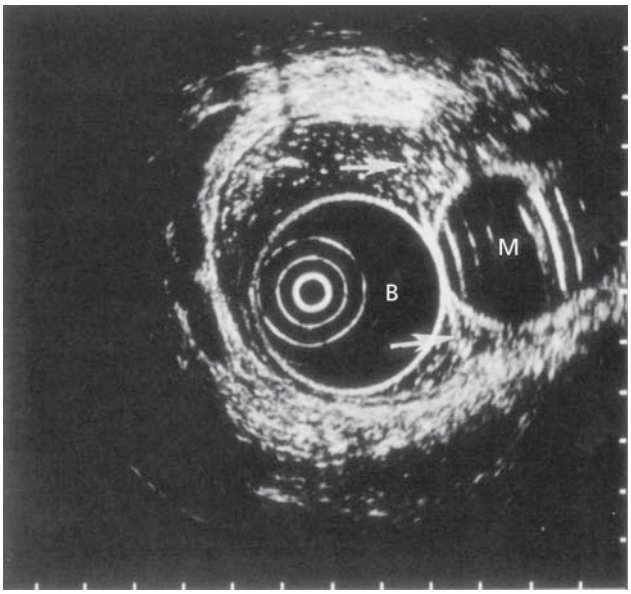

Figure 2.11 Mirror image (M) of the US transducer and water-filled balloon (B), produced by reverberation between the transducer and the air–water interface (arrow) within the gastric lumen.

of interest by filling the balloon or bowel lumen with water may help move the artifact away from the area of interest.

Another problem created by reverberation is the mirror-image artifact [22]. In this situation, US waves bounce off of an interface between water and air (Figure 2.11). This is typically seen when imaging within a partially water-filled organ such as the stomach or rectum. The US waves bounce back and forth between the transducer and the air–water interface, creating a mirror image of the transducer on the opposite side of the air–water interface (Figure 2.12). This effect is similar to observing both a mountain and its inverted reflection in a lake. The artifact is easily recognized and can be avoided by removing air and adding more water into the lumen.

Tangential scanning

As previously discussed, distances, and therefore tissue thicknesses, are most accurate when the US wave is perpendicular to the area of interest. When the US wave is tangential, tissue layers appear artificially thickened (Figure 2.13). This artifact can result in tumor "overstaging," especially in the esophagus and gastroesophageal (GE) junction, and particularly when the radial scanning US endoscope is used (Figure 2.14). To avoid this problem, the endoscope should be carefully maneuvered so that the US wave is perpendicular to the tissue. The normal wall layers should appear symmetric and of uniform thickness. When imaging abnormal tissue, care must be taken that the findings are reproducible and are not altered by small deflections of the endoscope tip.

Attenuation artifacts

Other artifacts are caused by attenuation of the US wave, but attenuation artifacts facilitate image interpretation in some cases. For example, lack of transmission of US through a

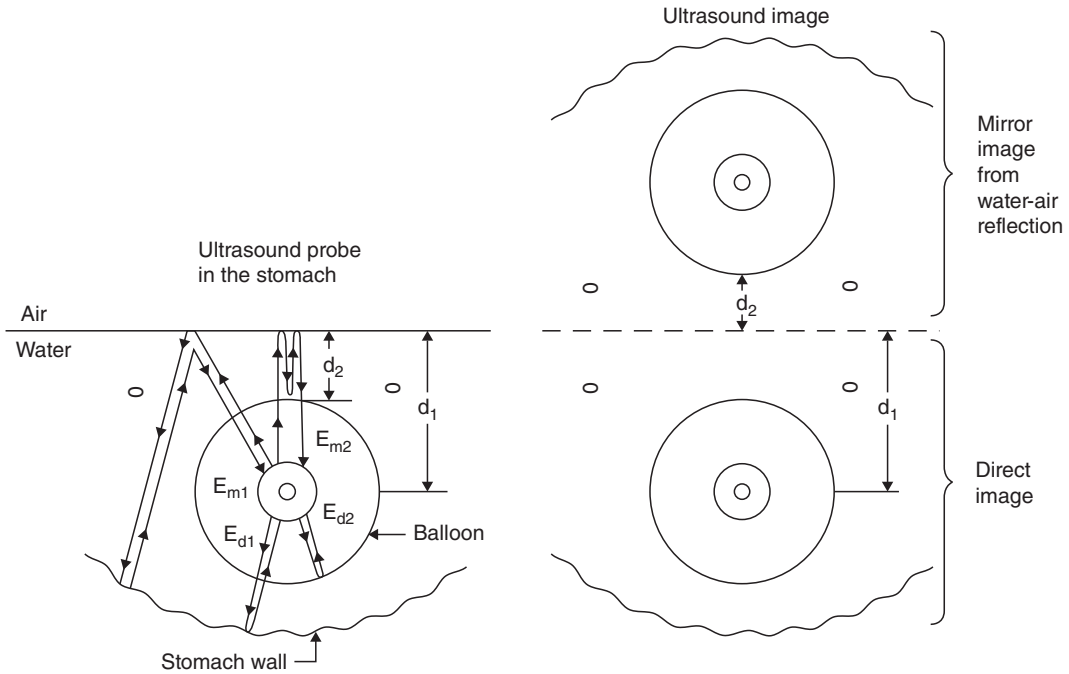

Figure 2.12 Production of a mirror-image artifact by reverberation of echoes from an air–water interface. The air–water interface reflects so strongly that US energy is redirected back to the transducer, just like light is redirected by a mirror. In the illustration at the left, the echoes E_{m1} and E_{m2} result from a double reflection, from the air–water interface and the stomach wall or balloon (or transducer case), respectively. The US processor records the position of the echo according to the time it receives the signal; the double reflection path takes longer and therefore causes the echo to appear further away from the transducer, as if it were a reflection in a mirror (diagram at left). The echoes received by the transducer directly (e.g., E_{d1} and E_{d2}) are displayed on the image in the expected location. The distance from the transducer to the air–water interface (d) and the distance from the balloon or transducer case to the interface (d_2) are also illustrated. (Reproduced from Kimmey MB, Martin RW. Fundamentals of endosonography. Gastrointest Endosc Clin North Am 1992;2:570, with permission from WB Saunders.)

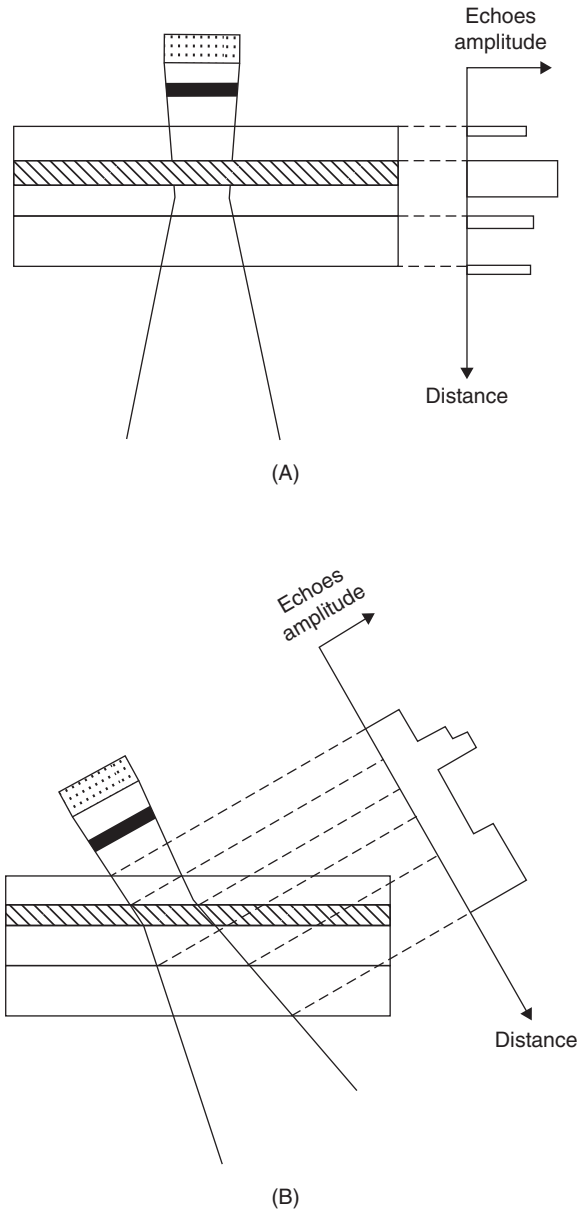

Figure 2.13 Why artifactual layer thickness increases with tangential scanning. (A) Amplitude and spatial duration of the echoes from the interfaces and specular reflectors in the normal GI wall when the US beam is at right angles to the wall. The diagonally-hatched region represents a tissue type with nonspecular echoes (e.g., the submucosa); the remaining echoes are produced by interfaces between tissue layers (specular echoes). The duration of the interface echoes is the same as the duration of the US pulse or the range resolution of the system (illustrated as a black rectangle in the beam). The echoes (displayed at the right) are spatially separated and distinguishable from one another. (B) When the US beam is not perpendicular to the wall, both the lateral and range resolution affect the duration of the echoes from each layer. In the extreme situation illustrated here, echoes from each layer overlap and cannot be distinguished individually. (Reproduced from Kimmey MB, Martin RW. Fundamentals of endosonography. Gastrointest Endosc Clin North Am *1992;2:572,* with permission from WB Saunders.)

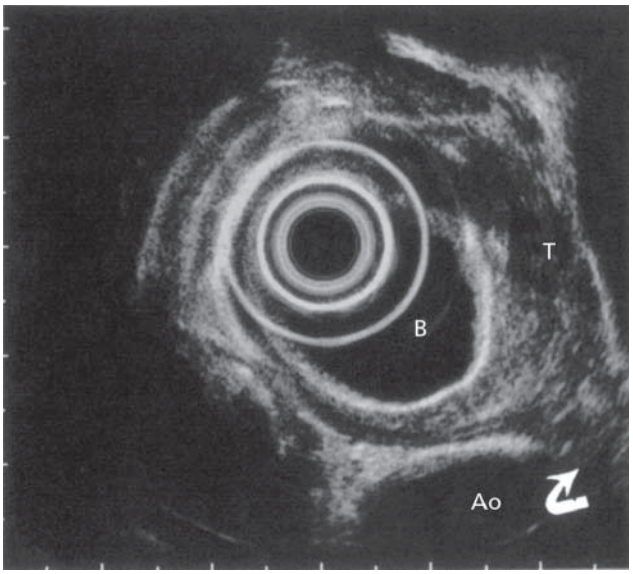

Figure 2.14 EUS image of an esophageal cancer (Tj), appearing to show invasion of the descending aorta (Ao) at the arrow. This is an artifact caused by nonperpendicular or tangential scanning. A clue to this is the location of the water-filled balloon (B): the transducer and balloon should be positioned in the center of the esophagus, with the transducer in the center of the balloon to avoid this artifact and avoid tumor over-staging.

gallstone or pancreatic duct stone is a key feature of cholelithiasis, choledocholithiasis, and pancreaticolithiasis. Soft tissue can also attenuate US waves, making it difficult to image deep into the tissue, especially when high-frequency transducers such as those on catheter probes are used. This can limit the ability to image the deep aspects of tissue masses.

Another common artifact is caused by attenuation by air bubbles. Bubbles develop in several unwanted locations, including the oil surrounding the transducer within the transducer housing, the water in the balloon on the outside of the transducer housing, water placed into the GI lumen, and air within the lumen itself. The transducer casing should be inspected for air bubbles prior to each procedure; removing these bubbles requires a minor repair by the manufacturer. Air bubbles in the balloon can be avoided by using degassed water and by repetitive filling and suctioning of the balloon prior to use. Air in water placed into the lumen can be avoided by using degassed water and by having the patient drink a simethicone "cocktail" before the procedure [23].

Side-lobe artifacts

These artifacts are characterized as nonshadowing echoes within an otherwise anechoic or fluid-filled structure [24]. They can be confused with biliary sludge in the gallbladder or with a mass within a pancreatic cyst (Figure 2.15). Side-lobe artifacts are caused by low-amplitude components of the transmitted US beam that are not perpendicular to the target. If these echoes are reflected by solid tissue outside the fluid-containing target, they may be displayed by the US processor as having come from the fluid-filled structure. When imaging solid tissue, low-amplitude side-lobe echoes are obscured by the echoes from the solid tissue and do not pose a problem in image interpretation. However, when an anechoic structure is being imaged, these echoes become visible and can artifactually suggest the presence of a solid component.

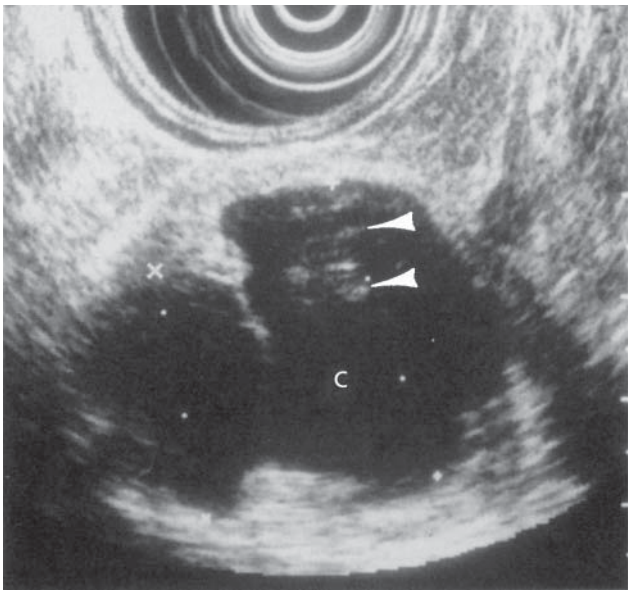

Figure 2.15 Pancreatic cyst (C) with apparent echoes (arrows), suggesting a solid component. These echoes are caused by side-lobe artifacts and are recognized because they are not consistently imaged when the transducer is maneuvered into another imaging plane.

They are easily recognized because they disappear with transducer movement and are eliminated by scanning from other angles.

Doppler artifacts

Artifacts associated with Doppler imaging can lead to signals being detected when no flow is present and, conversely, a lack of signal when flow is present. Flow can be artifactually seen when the Doppler gain is set too high. Under those conditions, bowel wall and transmitted cardiac and respiratory motion can be amplified and give the appearance of flow. However, this false signal is usually easy to recognize, because the Doppler signal is diffuse and is not localized to a specific structure.

False-negative Doppler signals can occur if the US beam is perpendicular to the target. Doppler shift is best detected with a US beam that is less than 60° incident to the target. Doppler can also miss low levels of venous flow if the US processor's wall filter is improperly set. This filter is meant to reduce noise from vessel wall motion, but can sometimes indiscriminately delete clinically important low-frequency echoes.

Table 2.2 Use of US principles to optimize image quality.

Principle	Practice
US frequency affects penetration depth	Use lower US frequency for distant targets
US frequency affects axial resolution	Use the highest US frequency that provides adequate penetration
Lateral resolution varies with distance from the transducer	Position the transducer so that the target is in the optimal focal zone
Attenuation is greater with higher US frequencies	Use lower frequencies for fatty and fibrous structures
The same tissue type should appear the same throughout the US image	Adjust the TGC on the US processor
Air transmits high-frequency US poorly	Eliminate air bubbles in the water-filled balloon and in the lumen
Images are more reliable if the US beam is perpendicular to the tissue	Recognize and avoid tangential scanning artifacts
Doppler shift is greatest with a tangential US beam	Adjust the transducer position to optimize the Doppler signal

Conclusion

The principles of US discussed in this chapter can be used to facilitate better endosonographic scanning and produce images that more accurately reproduce tissue structure. The importance of a standardized pre-procedure checklist and consistent procedure technique cannot be overemphasized. The basic steps in achieving an optimal examination, based on the principles discussed in this chapter, are summarized in Table 2.2.

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CHAPTER 3

Learning EUS anatomy

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Endoscopic Ultrasonography (EUS) is different than regular endoscopy in that it is a planar anatomy-based procedure. However, EUS anatomy is somewhat difficult to learn, as the planes generated are not often described in traditional anatomy learning material. Beyond this, there are other factors which increase the difficulty of becoming proficient at EUS anatomy. First, the images are generated by ultrasound, so one must be able to interpret an ultrasound image. Next, there are often patient features (obesity, hiatal hernias, variant anatomy) that can complicate placing an echoendoscope into a position in which it can generate the desired images.

General principles of EUS

EUS anatomy is easier to interpret if one considers a few basic concepts.

The first has to do with understanding the nature of ultrasonography. The transducer on the tip of the echoendoscope makes the sound waves and receives the echoes. The transducer has quartz (piezoelectric) crystals. An electric current applied to these crystals causes the crystals to vibrate and produce sound waves that travel outward. These waves are reflected back at various intensities, depending on what is in their path, and when they return to hit the crystals, the crystals emit electrical currents. The probe has an acoustic lens to help focus the emitted sound waves. Fat and air tend to strongly reflect sound waves, leading to bright (hyperechoic) images. Fluid tends to conduct sound waves, leading to dark (hypoechoic) images. Fluid-filled structures (arteries, veins, ducts) can generally be well seen and can be used as guides to finding organs and lesions of interest.

Endosonography is facilitated if one has a general knowledge of vascular and ductal anatomy, as these fluid-filled structures provide a "roadmap" of the regional anatomy. Figure 3.1 shows the major vascular and ductal structures of interest during an EUS exam. Familiarity with these structures simplifies EUS procedures.

Another important concept is that the echo endoscope may not go where the endoscopist thinks it is going. One can get lost while pushing in an endoscope, assuming that it is moving in a caudal direction when it is actually moving in a cephalad, anterior, or lateral direction. Rather than trying to figure out where one is by assuming a course, it is often better to trace a known structure (particularly a vessel or duct) to the desired location.

Finally, personal evaluation of computed tomography (CT) and abdominal ultrasound images helps one become better at endosonographic anatomy. One becomes better at EUS by becoming better at reading CT scans and transabdominal ultrasounds.

Echo endoscopes

There are two basic arrays of piezoelectric crystals on an echo endoscope: the radial array, which encircles the tip of the endoscope, and the linear array, which is parallel to the endoscope. The anatomic planes generated during EUS are quite different when one uses a radial versus a linear array probe. Although the early echo endoscopes were primarily radial array, the majority of EUS applications (such as fine-needle aspiration) currently use linear-array technology.

Regional anatomy

The esophagus and extraesophageal spaces

Esophageal EUS tends to follow traditional cross-section CT anatomy (radial-array exams approximate transaxial views (Figure 3.2), while linear-array images follow coronal and sagittal planes (Figure 3.3)). Extraesophageal EUS anatomy is the easiest to learn.

The esophagus runs a relatively straight course and is partially bordered by vascular structures, which provide excellent endosonographic images. If one is familiar with the aorta, the branches on the

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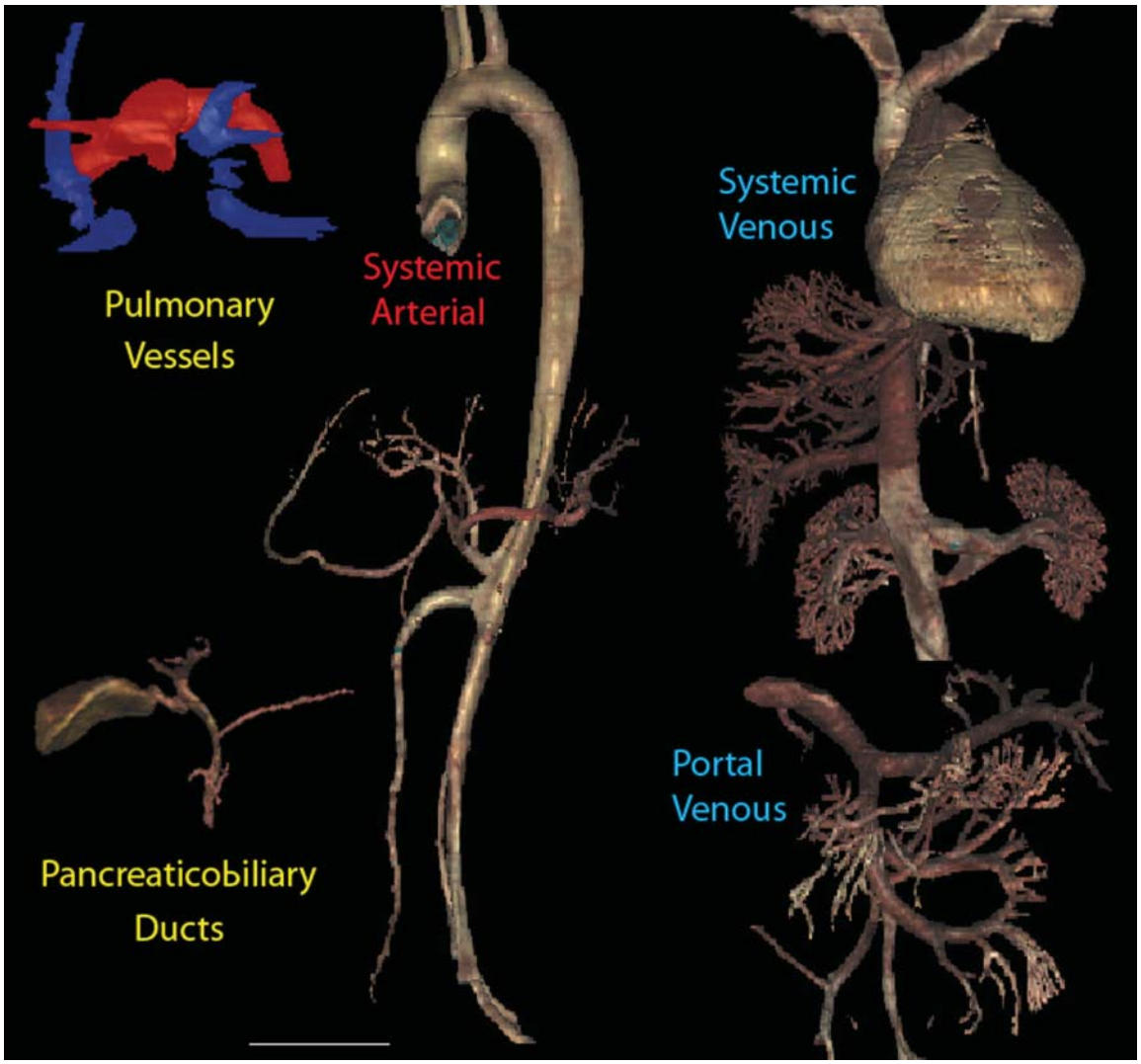

Figure 3.1 The major vascular and ductal structures of interest during an EUS exam. Taken from the TolTech dissector program using University of Colorado Visible Human data.

aortic arch, the azygos vein, and the heart, the other regional structures fall into place.

Esophageal radial array anatomy is very similar to routine transaxial CT anatomy from the thyroid to the diaphragm, and placing the aorta at 5 or 6 o'clock will approximate transaxial CT images (Figure 3.2). The thyroid, mediastinal nodes, vertebral column, and cardiac structures are usually clearly evident.

Linear-array exams are easiest after identification of the aorta. The mediastinum can be fully evaluated as the instrument is rotated. From the level of the aortic arch, the left subclavian and left carotid arteries are seen. Moving towards the stomach reveals the aortopulmonary window, subcarina space, azygos arch, and cardiac structures such as the great pulmonary vessels, left atrium, mitral valve, and left ventricle (Figure 3.3). The aorta can then be followed into the abdomen, down to the celiac artery.

Knowing the vascular anatomy allows one to use vessels to guide one's way to lesions. Figure 3.4 shows the major vessels of the chest and their relation to the esophagus.

The stomach and the extragastric spaces

The extragastric spaces can be a challenge to examine in full detail. There are many factors that can alter images, including hiatal hernias, different amounts of intraabdominal fat, and various orientations of the stomach within the abdomen. EUS anatomy reference material will usually show ideal images from ideal patients, but in practice most patients will not be ideal. In addition, the stomach does not confine the echo endoscope to any specific path. It is important to be able to find landmarks and work outwards from them, tracing known structures.