Section I Physics and Instrumentation

1

General Principles of Echocardiography

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Echocardiography is diagnostic imaging with ultrasound (sonography) of the heart. *Sonography* comes from the Latin *sonus* (sound) and the Greek *graphein* (to write). Diagnostic sonography is medical, real-time, two-dimensional (2D) and three-dimensional (3D) anatomic, motion, and flow imaging using ultrasound.

ULTRASOUND

Ultrasound is sound, a traveling pressure wave, of frequency higher than what humans can hear. Frequencies used in echocardiography range from about 2 MHz for adult transthoracic studies to about 10 MHz for higher-frequency applications such as harmonic imaging and pediatric and transesophageal studies. Higher frequencies produce images with improved detail resolution but with less penetration because the weakening of the ultrasound as it travels (attenuation) increases with increasing frequency. Ultrasound provides a live, noninvasive view of the heart. Echocardiography is accomplished with a pulse-echo technique.[1] Pulses of ultrasound, two to three cycles long, are generated by a transducer (Fig. 1.1) and directed into the patient, where they produce echoes at organ boundaries and within tissues. These echoes then return to the transducer, where they are detected and presented on the display of the sonographic instrument (Fig. 1.2). The ultrasound instrument processes the echoes and presents them as visible dots, which form the anatomic image on the display. The brightness of each dot represents the echo strength (amplitude), producing what is known as a *grayscale image*. The location of each dot corresponds to the anatomic location of the echo-generating object. Positional information is determined by knowing the path of the pulse as it travels and measuring the time it takes for each echo to return to the transducer. From a starting point at the top of the display, the proper location for presenting each echo is determined. Because the speed of the sound wave is known, the echo arrival time can be used to determine the depth of the object that produced the echo.

When a pulse of ultrasound is sent into tissue, a series of dots (one scan line, data line, or echo line) is displayed. Not all of the ultrasound pulse is reflected from any single interface. Rather, most of the original pulse continues into the tissue and is reflected from deeper interfaces. The echoes from one pulse appear as one scan line. Subsequent pulses go out in slightly different directions from the same origin. The result is a sector scan (sector image), which is shaped like a slice of pie (Fig. 1.3A). The resulting cross-sectional image is composed of many (typically 96 to 256) of these scan lines. For decades, sonography was limited to 2D cross-sectional scans (or slices) through anatomy such as that in Fig. 1.3A. 2D imaging has been extended into 3D scanning and imaging, also called *volume imaging,* as described in Chapter 2. This requires scanning the ultrasound through many adjacent 2D tissue cross sections to build up a 3D volume of echo information (see Fig. 1.3B), like a loaf of sliced bread in which each slice represents a 2D image and the loaf represents the 3D volume.

TRANSDUCER

The transducer used in echocardiography is a phased array that electronically steers the ultrasound beam in the sector format. It is energized by an electrical voltage from the instrument that produces each outgoing ultrasound pulse. The returning echo stream is received by the transducer and converted to an echo voltage stream that is sent into the instrument, ultimately appearing on the display as a scan line. This process occurs a few thousand times per second (called the *pulse repetition frequency* [PRF]). A coupling gel is used between the transducer and the skin to eliminate the air that would block the passage of ultrasound across that boundary. Transducers (see Fig. 1.1) are designed for transthoracic (see Section II) and for transesophageal (see Section III) imaging. The latter provides a shorter acoustic path (with less attenuation, allowing higher frequencies and improved resolution) to the heart that avoids intervening lung and ribs.

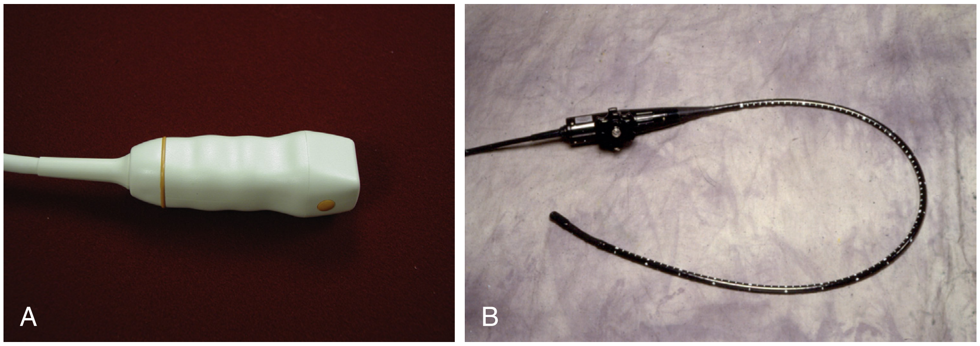

Figure 1.1. A, Transthoracic transducer. **B,** Transesophageal transducer.

INSTRUMENT

The echocardiographic instrument has a functional block diagram as shown in Fig. 1.2B. The beam former drives the transducer and receives the returning echo streams, amplifying (this is called *gain*) and digitizing them. Attenuation compensation occurs in the reception side of the beam former. This is an amplification process that equalizes echo strengths (amplitudes) by increasing gain with depth. This compensates for the weakening of echo amplitude with depth caused by attenuation. The signal processor, among other functions, detects the amplitude of each echo voltage. Echo amplitudes are stored digitally in the image memory, which is part of the image processor. Upon completion of a single scan (one frame of a real-time presentation), the stored image is sent to the flat-panel display. The echo information enters the image memory in ultrasound scan-line sector format but is read out to the display in horizontal-line display format, with each horizontal line on the display corresponding to a row of echo data in the image memory.

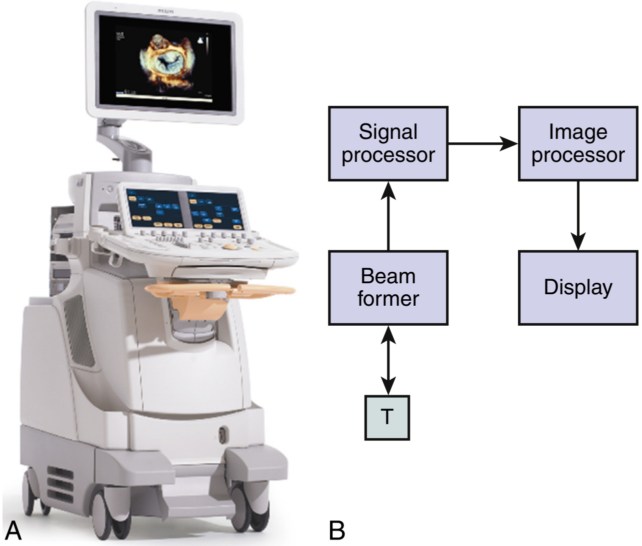

Figure 1.2. A, Echocardiographic instrument. **B,** Block diagram of an echocardiographic instrument. *T,* transducer.

ARTIFACTS

In imaging, an artifact is any presentation that does not correctly display the structure or function (tissue motion and blood flow) imaged. Artifacts are caused by problematic aspects of the imaging process. They can hinder correct interpretation and diagnosis. These artifacts must be avoided or managed properly when encountered to avoid the errors they can cause. Some artifacts are produced by improper equipment operation or settings (e.g., incorrect gain and compensation settings). Other artifacts are inherent in the sonographic methods and can occur even with proper equipment and technique. The assumptions inherent in the sonographic process include the following:

- Sound travels in straight lines.
- Echoes originate only from objects located on the beam axis.
- The amplitude of returning echoes is related directly to the reflecting or scattering properties of the objects that produces them.
- The distance to reflecting or scattering objects is proportional to the round-trip travel time at a speed of 1.54 mm/µs.

If any of these assumptions are violated, artifacts occur. Fig. 1.4 and Video 1.4 provide examples of artifacts in echocardiography.

VIRTUAL BEAMFORMING

Two alternative fundamental principles of operation are now present in the array of commercially available echocardiographic equipment. The first, termed *conventional echocardiography*, has been the operating principle for more than 50 years and is described earlier in this chapter. Recently, a second principle has appeared, termed *virtual-beam echocardiography*.[1,2] These two operating principles are fundamentally different, and there are implications for the resulting anatomic images and motion information presented. Virtual-beam sonography improves nearly every aspect of echocardiographic, anatomic imaging, motion, and flow presentation. Rather than a one-to-one relationship of pulse to scan line, characteristic of conventional echocardiography, virtual-beam echocardiography acquires each image with a few broad, unfocused beams and then through massive, high-speed, retrospective computation accomplished with graphics processing units determines the amplitude (and Doppler shift if needed) of the echo at each pixel location (Fig. 1.5). This computational, retrospective beamforming capitalizes on the fact that an echo arrives at the elements of the transducer in a sequence uniquely characteristic of the location from which it originated. The challenge is to sort out each echo from the mixed-up combination in

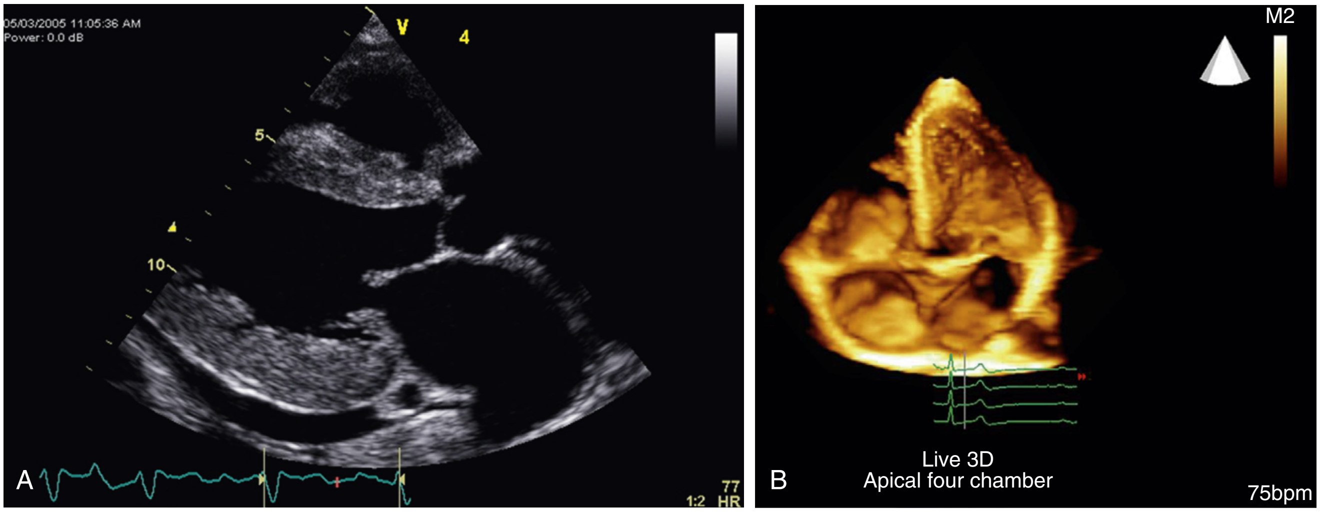

Figure 1.3. A, Two-dimensional cardiac sector image. **B,** Three-dimensional cardiac image.

Video 1.4. Echocardiography artifacts. Example of a scan that illustrates two artifacts in echocardiography: a grating lobe producing an artifactual mitral valve and a mirror image producing an artifactual aortic valve.

Video 1.6. Raindrop waves. Multiple water waves from raindrops illustrate the challenge of sorting out the source of each wave when received at the edge of the video.

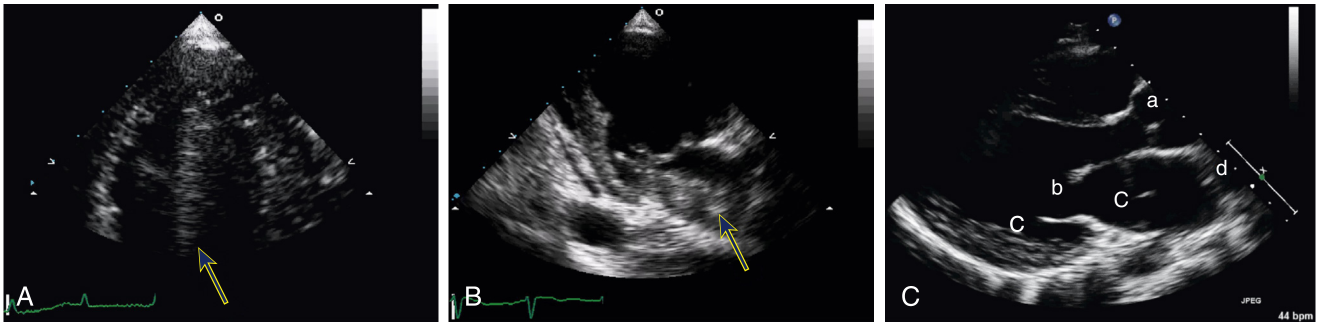

Figure 1.4. A, Comet-tail artifact *(arrow)*. **B,** Grating-lobe artifact *(arrow)* mimicking a mass in the atrium. **C,** Aortic valve (a), mitral valve (b), grating lobe (c) duplication of (b), and mirror-image (d) duplication of (a). See also Video 1.4.

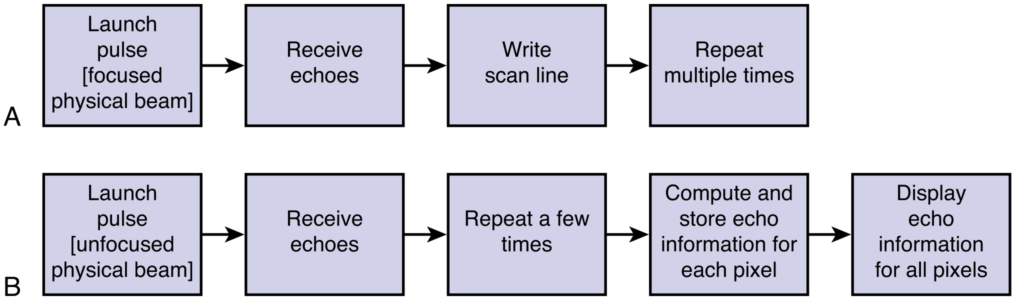

Figure 1.5. A, Sequence of operations for conventional, pulse-echo echocardiography. **B,** Sequence of operations for virtual-beam echocardiography.

which they arrive at the transducer, similar to waves from raindrops combining as they travel (Fig. 1.6 and Video 1.6). When the computational process for a frame is complete, the results are sent to the display to show that frame. Virtual beamforming produces images that are in focus throughout, with higher frame rates, increased sensitivity and penetration, and improved contrast resolution compared with conventional echocardiography (Fig. 1.7 and Video 1.7).

Artificial intelligence, machine learning, and deep learning are invading echocardiography, taking on some optimization functions that formerly were the responsibility of the echocardiographer, and automating some functions. This transition to instrument optimization and automation will continue, will alter the role of echocardiographers, and will improve the accuracy and efficiency of diagnostic echocardiography.

Advanced features and techniques, including 3D echocardiography, Doppler principles, tissue Doppler imaging, speckle tracking, and strain, are covered in the following chapters in this section. Expansion of all the topics covered in this chapter can be found in Kremkau.[1,2]

Please access ExpertConsult to view the corresponding videos for this chapter.

Figure 1.6. Raindrops falling on a puddle are an example of the computational challenge with virtual beamforming. Note that the wave emanating from each drop travels away to be combined with all the others. If they can be sorted out as they arrive at any edge of the photo, the location of their origins and strengths can be individually determined based on their arrival characteristics. This is analogous to what the retrospective computational process accomplishes virtual beamforming. See also Video 1.6.

Video 1.7. Conventional and virtual beamforming. The superior quality of the virtual beamforming example *(right)* is seen compared with conventional beamforming *(left)*.

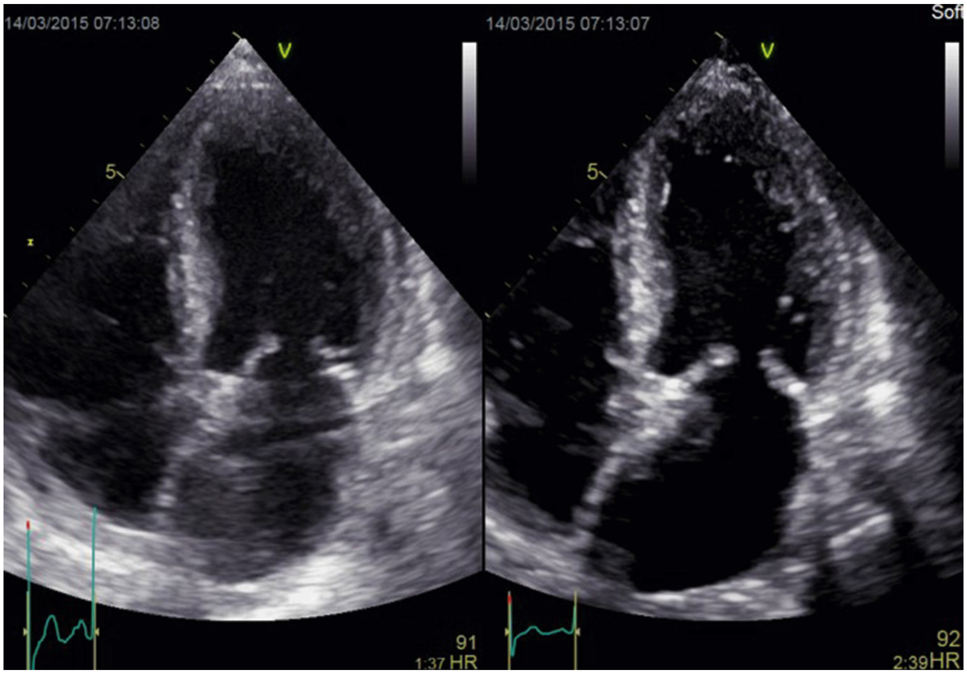

Figure 1.7. Comparison of conventional pulse-echo imaging *(left)* with virtual beamforming *(right)*. See also Video 1.7.

REFERENCES

1. Kremkau FW. *Sonography: Principles and Instruments*. 10th ed. Philadelphia: Elsevier/Saunders; 2020.
2. Kremkau FW. Your new paradigm for understanding and applying sonographic principles. *J Diag Med Sonography*. 2019;35(5):439–446.

2

Three-Dimensional Echocardiography

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The milestone in the history of three-dimensional echocardiography (3DE) has been the development of fully sampled matrix-array transthoracic transducers based on advanced digital processing and improved image formation algorithms that allowed the operators to obtain on-cart transthoracic real-time volumetric imaging with short acquisition times and high spatial and temporal resolution. Further technological developments (i.e., advances in miniaturization of the electronics and in element interconnection technology) made it possible to insert a full matrix array into the tip of a transesophageal probe for real-time volumetric imaging. In addition to transducer engineering, improved computer processing power and the availability of dedicated software packages for both on- and offline post-processing have allowed 3DE to become a practical clinical tool.

COMPARISON BETWEEN TWO- AND THREE-DIMENSIONAL ECHOCARDIOGRAPHY ULTRASOUND TRANSDUCERS

The backbone of the 3DE technology is the transducer. A conventional two-dimensional (2D) phased array transducer is composed of 128 piezoelectric elements, electrically isolated from each other, arranged in a single row (Fig. 2.1). Ultrasound wave fronts are generated by firing individual elements in a specific sequence with a delay in phase with respect to the transmit initiation time. Each element adds and subtracts pulses to generate a single ultrasound wave with a specific direction that constitutes a radially propagating scan line (Fig. 2.2). Because the piezoelectric elements are arranged in a single row, the ultrasound beam can be steered in two dimensions—vertical (axial) and lateral (azimuthal)—while resolution in the z-axis (elevation) is fixed by the thickness of the tomographic slice, which in turn is related to the vertical dimension of the piezoelectric elements.

Currently, 3DE matrix-array transducers are composed of approximately 3000 individually connected and simultaneously active (fully sampled) piezoelectric elements with operating frequencies ranging from 2 to 4 MHz and 5 to 7 MHz for transthoracic and transesophageal transducers, respectively. To steer the ultrasound beam in 3D space, a 3D array of piezoelectric elements needs to be used in the probe; therefore, piezoelectric elements are arranged in rows and columns to form a rectangular grid (matrix configuration) within the transducer (see Fig. 2.1, *right*). The electronically controlled phasic firing of the elements generates a scan line that propagates radially (y or axial direction) and can be steered both laterally (x or azimuthal direction) and in the elevation plane (z direction) to acquire a volumetric pyramid of data (see Fig. 2.1, *right*). Matrix-array probes can also provide real-time multiple simultaneous 2D views at high frame rates in predefined or user-selected plane orientations (Fig. 2.3). The main technological breakthrough that allowed manufacturers to develop fully sampled matrix