Basic Physics of Ultrasound in Transesophageal Echocardiography

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Abstract
The Transoesophageal echocardiography is proving a valuable tool to evaluate, diagnose and treat patients in perioperative and intensive care settings. This was previously limited to only cardiologist but now is accessible to most of us, especially in perioperative care. This article describes the basic physics of ultrasound involved in echocardiography. Understanding the basic physics will help the reader to obtain and interpret the images more effectively.

Keywords: Transoesophageal echocardiography, ultrasound, Doppler, artifact, perioperative care, intensive care.

INTRODUCTION

Since the introduction of transesophageal echocardiography (TEE) in 1980s, its clinical applications have been on the rise. During the initial period, the anaesthesiologists and intensivists did not have sufficient experience to evaluate, diagnose and treat patients using echocardiography. Now, this valuable tool can be used in the perioperative and intensive care setting, if the patient’s status warrants it, by anyone who has acquired the knowledge and experience in echocardiography. The first step towards this is a thorough understanding of the physics associated with echocardiography. This article deals with the basic physics of ultrasound involved in echocardiography and is applicable to other modes of ultrasonography as well.

In transthoracic echocardiography (TTE), the heart and great vessels are probed with ultrasound waves which are partially reflected by these structures. From these reflections, information about distance, velocity and density of the objects within the chest is derived. The echocardiography studies across lungs, bone and soft tissue especially in obese or emphysematous patients increases attenuation and hence a poor return of ultrasound signal and a poor quality image. In contrast, the sound waves emitted from a TEE transducer has to only pass through the oesophageal wall and pericardium to reach the heart.

This reduces the attenuation of the ultrasound signal, thus generating a stronger return signal with less distortion and better resolution ultimately enhancing the image quality. The development of high frequency transesophageal transducers along with real time blood flow measurement by Doppler ultrasound has revolutionised the cardiac monitoring in the perioperative and critical care setting.

ULTRASOUND

Sound waves are vibrations transmitted through a medium. These waves can cause compression and rarefaction or positive and negative pressure changes within the medium they pass. An ultrasound is a cyclic sound energy in the form of waves with a frequency greater than the upper limit of human hearing. The audible frequency range of humans is 20 to 20 kHz (cycles per second) and above this range, they are referred to as ultrasound frequencies. Ultrasound is produced by the piezoelectric crystals that rapidly change their shape producing compression and rarefaction on application of the electric current. Ultrasound frequency used for medical purposes is from 1 to 20 MHz. The TTE uses low-frequency transducers (2 to 4 MHz), which allows better penetration of the acoustic energy through the chest wall, but at the cost of reduced longitudinal resolution. TEE does not require such penetration hence uses higher...
frequency transducers (3.5 to 7 MHz) to produce better resolution imaging.

An ultrasound image is created in three steps. Firstly an ultrasound transducer emits brief pulses of sound (typically 3 to 5 cycles). Secondly the transducer listens for the returned echoes during the long period of silence following each pulse. Then lastly these echoes are interpreted.

Properties of ultrasound waves (Fig. 1)

Frequency (f)
The frequency is the number of cycles or wavelengths per unit of time. It is measured as cycles per second and the SI unit of frequency is hertz (Hz), named after the German physicist Heinrich Hertz. The higher the frequency, better the resolution but lower the penetration.

Wavelength (λ)
One wavelength equals the distance between two successive wave compressions or rarefactions. The wavelength depends on the size of the piezoelectric crystals in the transducer and the medium through which the sound wave travels.

Velocity (V)
Velocity of sound waves varies with the density and stiffness of the medium. Greater the stiffness, the faster sound waves will travel. Therefore the velocity of sound is low in liquids, highest in solids and very little in air. The average velocity of waves in soft tissue and blood is about 1540 m/sec.

Velocity (V) = frequency (f) × wavelength (λ)

Amplitude (A)
The amplitude is the extent of a vibratory movement measured as the distance from baseline to peak of the wave. Amplitude decreases as the wave travels (attenuation) leading to echoes from deeper structures being weaker than those from superficial structures.

Interaction of Ultrasound Waves with Tissues (Fig. 2)

Acoustic Impedance (Z)
When a sound wave encounters a material with a different density (acoustic impedance), various things can happen to the sound wave as mentioned below. Acoustic impedance depends on the density of the medium and propagation speed of the wave but as the speed between soft biological tissues varies only slightly, impedance depends mainly on the difference in density. The greater the difference between acoustic impedances, the larger will be the echo.

Attenuation
Attenuation occurs due to the absorption of ultrasound energy by conversion to heat. This also occurs because of reflection, refraction and scattering. The attenuation of the sound wave is increased at higher frequencies, so in order to have better penetration of deeper tissues, a lower frequency is used. The time gain compensation (TGC, one of the settings of the ultrasound machine) compensates for attenuation of ultrasound signals from distant objects.

Reflection
Part of the sound wave is reflected back to the transducer and is detected as an echo. The time taken for the echo to
travel back to the transducer is measured and is used to calculate the depth of the tissue interface causing the echo.

Refraction and Scattering

Refraction is the bending of a wave beam when it crosses the interface of two materials with different propagation velocities at an oblique angle, similar to light waves (Snell’s law). This does not happen when the incident ray is perpendicular. Refraction can sometimes produce artifacts but are usually not a problem with TEE.

Resolution

Resolution is the ability to distinguish between two objects by the ultrasound waves. Spatial resolution is the ability to distinguish two objects that are located close to each other. Spatial resolution can be axial or lateral. Axial resolution is the ability to distinguish two different objects at different depths from the transducer along the axis of the ultrasound beam. Lateral resolution is the ability to distinguish two objects that are side by side. Temporal resolution is the ability to locate objects at a particular instant in time and is expressed as frames per second.

Transducers

The ultrasound waves are produced by a transducer. A transducer is a device that converts one form of energy into another form. In this case, the electrical pulses produce mechanical movement (oscillations) of a crystal thus producing ultrasound waves, and this is called the piezoelectric effect. Piezoelectric crystals rapidly change shape on application of the electric current and create electricity when the ultrasound strikes them. Hence a piezoelectric crystal converts electrical energy into ultrasound when transmitting, and converts ultrasound into electrical energy when receiving. The echocardiography transducers are commonly made from a substance with piezoelectric properties such as lead-zirconate-titanate-5 (PZT-5). Transducers regulate the duration and frequency of emitted ultrasound waves. The ultrasound waves reflected back from the tissues are received by the transducers and an image is constructed. They may be single plane, biplane, and multiplane, phased array transducer (the beam is formed by firing an array of small, closely spaced transducer elements in a sequence) with 64 elements or paediatric transducers with 26 elements.

Modes of Imaging

A-Mode (Amplitude)

The simplest single dimension mode is the A-mode or Amplitude mode. It is not used any longer in echocardiography as it is of limited clinical usefulness.

M-Mode (Motion) (Fig. 3)

The basic form of ultrasound imaging in echocardiography is M-mode or the motion mode, where the density and position of all tissues in the path of the narrow ultrasound beam are displayed as a scroll on a video screen. It is a timed motion display. Cardiac tissue is always in motion, so this provides a continuously changing tissue section study. It is used for precise timing of cardiac events but not used as the primary imaging technique as only a limited heart tissue is observed at any one time and this requires considerable interpretation. It is often used with colour flow Doppler for timing of abnormal flows. M-mode images are updated about 1000 times/sec so that more subtle changes in motion or size can be appreciated compared to the 2D echo. Also all the quantitative measurements can be performed without numerous analysis stations.

Two-dimensional Mode or B-mode (Brightness)

2D mode is the mode generally used in TEE, which includes rapid, repetitive scanning within the area in a sector. For this mode, a mechanical transducer or a phased array transducer (electronic) is used. In a 2D scan the images are updated 30 to 60 times/sec, thus producing a real time image of the heart. The image resembles an anatomical section and is easily interpreted. Best images are obtained when...
beam is perpendicular to scanned object. The Doppler beam can be superimposed on the 2D image to get audio and visual interpretation of the reflected signal and is used to measure blood velocities.

**Real time Three-dimensional (RT-3D)**

Recently, RT-3D cardiac echocardiography imaging has become popular, especially for perioperative noninvasive imaging of intracardiac lesions such as valvular and congenital heart disease. It has been shown to be more accurate in imaging of mobile structures like left ventricular thrombi and myocardial wall motion. In clinical practice the role of RT-3D echocardiography continues to evolve.

**DOPPLER (FIG. 4)**

During echocardiography using the Doppler principle (Christian Doppler in 1842) velocity of the bloodstream in the heart and great vessels can be measured.

Doppler effect is the shift in the frequency and wavelength of waves, which results from a motion of the source, motion of the observer, or motion of the medium. So the velocity (V) of the blood flow can be calculated by measurement of the relative change in the returned ultrasound frequency when compared to the transmitted frequency. One of the limitations is that the ultrasound beam should be as parallel to the blood flow as possible to avoid any errors in the measurements.

Doppler Equation

\[ V = \frac{(C \times f_d)}{(2 \times f_o \times \cos \theta)} \]

V: Velocity of the moving blood
C: Constant (velocity of sound in blood)

\( f_d \): Frequency shift
\( f_o \): Frequency of transmitted ultrasound
\( \cos \theta \): Angle between the ultrasound beam and the direction of blood flow, best when below 20° (Note that, \( \cos 0° = 1 \) and \( \cos 90° = 0 \)).

There are different types of Doppler’s used in echocardiography with various advantages and disadvantages.

**Continuous Wave Doppler (CW) (Figs 5 and 7)**

The advantage of CW Doppler is its ability to measure high blood velocities accurately (for example in aortic stenosis). But the disadvantage is, its lack of selectivity or depth discrimination, because it reflects the ultrasound data from along the course of the beam.

**Pulse Wave Doppler (PW) (Figs 6 and 7)**

One main advantage of pulsed Doppler is its ability to provide Doppler shift data selectively from a small volume of sample along the ultrasound beam (for example mitral valve inflow). The location of this sample volume is operator controlled. The main disadvantage of PW Doppler is its inability to accurately measure high blood flow velocities (velocities above 1.5 to 2 m/s), known as aliasing.

**Pulse Repetition Frequency (PRF) (Hz)** is the number of pulses in one second. PRF is inversely related to imaging depth.

Aliasing is an artifact that lowers the frequency components when the pulse repetition frequency (PRF) is less than 2 times the highest frequency of a Doppler signal.
Nyquist Frequency (NF) is the maximum frequency that can be sampled without aliasing. Therefore NF = PRF/2.

**Colour Doppler (Figs 5 and 6)**

Colour Doppler is a technique for visualising the velocity of blood within an image plane. It measures the Doppler shifts in a few thousand sample volumes located in an image plane. For each sample volume, the average Doppler shift is encoded as a colour and displayed on top of the 2D image. Conventionally, the direction and velocity of blood is depicted in red for the movement toward the transducer and blue for movement away from the transducer (BART, meaning blue away and red towards).

**ARTIFACTS**

Many of the important imaging artifacts result from the interplay of the ultrasound system, the patient, and the interpreting echocardiographer.

**Reverberation**

Reverberation artifact causes evenly spaced lines at increasing depths and is caused by sound reflecting back and forth between the surface of the probe and a strong reflector close to the surface.

**Sidelobe**

The ultrasound probe cannot produce a pulse that travels purely in one direction. Pulses also travel off at specific angles. Sidelobe beams are generated from the edges of the transducer and project in a different direction from the main beam. These echoes are much weaker than those of the main beam but if a very strong reflector is encountered, they may be strong enough on returning to the transducer to be displayed prominently on the image.

**Shadowing**

Acoustic shadowing is the loss of data below a dense object because the majority of the sound energy has been reflected. This occurs typically in prosthetic valves.

Other artifacts include mirror image and ring-down artifacts. By knowing the anatomy, using different planes and settings for imaging, it is possible to recognise the artifacts.

**COMPLICATIONS**

So far ultrasound waves have shown no adverse effects on biological tissues. The main concern is localised heating leading to burns. Heat from transducers can cause esophageal burns. Most transducers have built-in thermometers to prevent transmission when temperature reaches 40°C. But burns can occur even with lower temperatures in hypothermic patients. Freezing the image display will halt transmission of waves.

TEE has been generally found to be safe, but the commonest complications are related to the injury to the mucosa (oropharyngeal, oesophageal and gastric) and bleeding.

In conclusion, TEE provides instantaneous clinical information in the perioperative and intensive care setting. Understanding the basic physics will help the reader to obtain and interpret the images more effectively.

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