Lecture Five

ULTRASONIC FLOWMETERS

INTRODUCTION

The ultrasonic flowmeter, like the electromagnetic flowmeter, can measure instantaneous flow of blood. The ultrasound can be beamed through the skin, thus making transcutaneous flowmeters practical. Advanced types of ultrasonic flowmeters can also measure flow profiles. These advantages are making the ultrasonic flowmeter the subject of intensive development.

TRANSDUCERS

The transducer used in ultrasonic flowmeter uses a piezoelectric material that converts power from electric to acoustic form. Lead zirconate titanate is a crystal that has the highest conversion efficiency. It can be molded into any shape by melting. As it is cooled through the Curie temperature, it is placed in a strong electric field to polarize the material. It is usually formed into disks that are coated on opposite faces with metal electrodes and driven by an electronic oscillator.

The resulting electric field in the crystal causes mechanical constriction. The piston-like movements generate longitudinal plane waves, which propagate into the tissue. For maximal efficiency, the crystal is one-half wavelength thick. Any cavities between the crystal and the tissue must be filled with a fluid or watery gel in order to prevent the high reflective losses associated with liquid-gas interfaces.

Because the transducer has a finite diameter, it will produce diffraction patterns. Figure 4 shows the outline of the beam patterns for several transducer diameters and frequencies. In the near field, the beam is largely contained within a cylindrical outline and there is little spreading. The intensity is not uniform, however: There are multiple maximums and minimums within this region, caused by interference. The near field extends a distance d_{nf} given by:

$$d_{nf} = \frac{D^2}{4\lambda}$$

where D = transducer diameter and λ = wavelength.

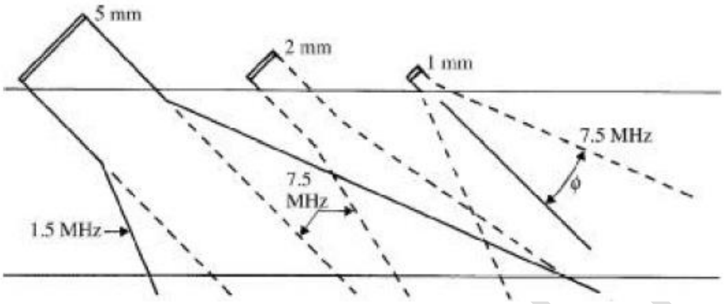

Figure 4 Near and far fields for various transducer diameters and frequencies. Beams are drawn to scale, passing through a 10 mm-diameter vessel. Transducer diameters are 5, 2, and 1 mm. Solid lines are for 1.5 MHz, dashed lines for 7.5 MHz.

In the far field the beam diverges, and the intensity is inversely proportional to the square of the distance from the transducer. The angle of beam divergence ϕ, shown in Figure 4, is given by:

$$\sin\phi = \frac{1.2\lambda}{D}$$

Figure 4 indicates that we should avoid the far field because of its lower spatial resolution. To achieve near-field operation, we must use higher frequencies and larger transducers.

Several factors are considered in order to select the operating frequency. For a beam of constant cross section, the power decays exponentially because of absorption of heat in the tissue. The absorption coefficient is approximately proportional to frequency, so this suggests a low operating frequency. However, most ultrasonic flowmeters depend on the power scattered back from moving red blood cells. The backscattered power is proportional to f^4, which suggests a high operating frequency. The usual compromise dictates a frequency between 2 and 10 MHz.

TRANSIT-TIME FLOWMETER

Figure 5 shows the transducer arrangement used in the transit-time ultrasonic flowmeter. The effective velocity of sound in the vessel is equal to the velocity of sound, c, plus a component due to û, the velocity of flow of blood averaged along the path of the ultrasound. For laminar flow, û = 1.33 ū, and for turbulent flow, û = 1.07 ū, where ū is the velocity of the flow of blood averaged over the cross-sectional area. Because the ultrasonic path is along a single line rather than averaged over the cross sectional area, û differs from ū. The transit time in the downstream (+) and upstream (-) directions is

$$t = \frac{Distance}{Conduction\ velocity} = \frac{D}{c \pm \hat{u} cos\theta}$$

The difference between upstream and downstream transit times is

$$\Delta t \cong \frac{2D\hat{u}cos\theta}{c^2}$$

Thus, the average velocity $\hat{u}$ is proportional to Δt. A short acoustic pulse is transmitted alternately in the upstream and downstream directions. Unfortunately, the resulting Δt is in the nanosecond range, and complex electronics are required to achieve adequate stability. Like the electromagnetic flowmeter, the transit-time flowmeter and similar flowmeters using a phase shift principle can operate with either saline or blood as a fluid, because they do not require particulate matter for scattering. However, they do require invasive surgery to expose the vessel.

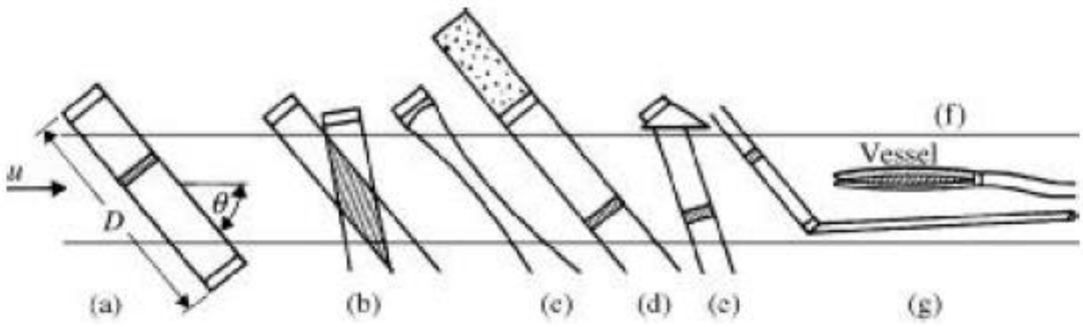

Figure 5 Ultrasonic transducer configurations (a) A transit-time probe requires two transducers facing each other along a path of length D inclined from the vessel axis at an angle 0. The hatched region represents a single acoustic pulse traveling between the two transducers. (b) In a transcutaneous probe, both transducers are placed on the same side of the vessel, so the probe can be placed on the skin. Beam intersection is shown hatched. (c) Any transducer may contain a plastic lens that focuses and narrows the beam. (d) For pulsed operation, the transducer is loaded by backing it with a mixture of tungsten powder in epoxy. This increases losses and lowers Q. Shaded region is shown for a single time of range gating. (e) A shaped piece of Lucite on the front loads the transducer and also refracts the beam. (f) A transducer placed on the end of a catheter beams ultrasound down the vessel. (g) For pulsed operation, the transducer is placed at an angle.

CONTINUOUS-WAVE DOPPLER FLOWMETER

When a target recedes from a fixed source that transmits sound, the frequency of the received sound is lowered because of the Doppler effect. For small changes, the fractional change in frequency equals the fractional change in velocity.

$$\frac{f_d}{f_o} = \frac{u}{c}$$

Where: f_d = Doppler frequency shift, f_0 = source frequency, u = target velocity, c = velocity of sound.

The flowmeter shown in Figure 6 requires particulate matter such as blood cells to form reflecting targets. The frequency is lowered twice. One shift occurs between the transmitting source and the moving cell that receives the signal. The other shift occurs between the transmitting cell and the receiving transducer.

$$\frac{f_d}{f_o} = \frac{2u}{c+u} \cong \frac{2u}{c}$$

The approximation is valid, because c=1500m/s and u=1.5 m/s. The velocities do not all act along the same straight line, so we add an angle factor:

$$f_d = \frac{2f_o u cos\theta}{c} \dots \dots \dots (1)$$

where θ is the angle between the beam of sound and the axis of the blood vessel, as shown in Figure 6. If the flow is not axial, or the transducers do not lie at the same angle, such as in Figure 5(b), we must include additional trigonometric factors.

Figure 6 shows the block diagram of a simple continuous-wave flowmeter. The oscillator must have a low output impedance to drive the low-impedance crystal. Although at most frequencies the crystal transducer has a high impedance, it is operated at mechanical resonance, where the impedance drops to about 100Ω.

The ultrasonic waves are transmitted to the moving cells, which reflect the Doppler-shifted waves to the receiving transducer. The receiving transducer is identical to the transmitting transducer. The amplified radio-frequency (RF) signal plus carrier signal is detected to produce an audio-frequency (AF) signal at a frequency given by (1).

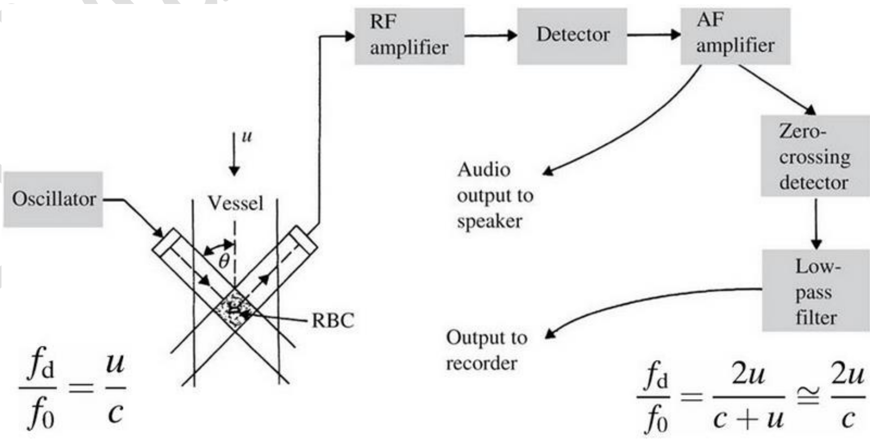

Figure 6 Doppler ultrasonic blood flowmeter. In the simplest instrument, ultrasound is beamed through the vessel walls, backscattered by the red blood cells, and received by a piezoelectric crystal.

PULSED DOPPLER

Continuous-wave flowmeters provide little information about flow profile. Therefore, several instruments have been built that operate in a radar-like mode. The transmitter is excited with a brief burst of signal. The transmitted wave travels in a single packet, and the transmitter can also be used as a receiver, because reflections are received at a later time.

The delay between transmission and reception is a direct indication of distance, so we can obtain a complete plot of reflections across the blood vessel. By examining the Doppler shift at various delays, we can obtain a velocity profile across the vessel. To achieve good range resolution, the transmitted-pulse duration should ideally be very short. To achieve a good SNR and good velocity discrimination, it should be long. The usual compromise is an 8MHz pulse of 1ps duration, which produces a traveling packet 1.5mm long, as shown in Figure 5(d). The intensity of this packet is convolved with the local velocity profile to produce the received signal.

Thus, the velocity profile of the blood vessel is smeared to a larger-than-actual value. Because of this problem, and also because the wave packet arrives at an angle to normal, the location of the vessel walls is indistinct. It is possible, however, to mathematically "deconvolve" the instrument output to obtain a less smeared representation of the velocity profile.

There are two constraints on pulse repetition rate f_r. First, to avoid range ambiguities, we must analyze the return from one pulse before sending out the next. Thus

$$f_r > 2f_0$$

thus:

$$u_m(cos\theta)R_{max} < \frac{c^2}{8f_0}$$

which shows that the product of the range and the maximal velocity along the transducer axis is limited. In practice, measurements are constrained even more because of (1) spectral spreading, which produces some frequencies higher than those expected, and (2) imperfect cutoff characteristics of the low-pass filters used to prevent *aliasing* (generation of fictitious frequencies by the sampling process).

Because we cannot easily start and stop an oscillator in 1ps, the first stage of the oscillator operates continuously. The transmitter and the receiver both use a common piezoelectric transducer, so a *gate* is required to turn off the signal from the transmitter during reception. A one-stage gate is not sufficient to isolate the large transmitter signals from the very small received signals. Therefore, two gates in series are used to turn off the transmitter.