

# DOPPLER ULTRASOUND:

## Introduction

Ultrasound technology has evolved dramatically since its introduction to medicine in the late 1940s, originating from sonar technology developed during World War II. When combined with the Doppler effect—a physical phenomenon describing frequency changes in moving objects—ultrasound becomes an extraordinarily powerful diagnostic tool. Unlike static imaging methods such as X-rays or CT scans, Doppler ultrasound captures dynamic information about blood flow, providing both anatomical and functional data in real time. This dual capability has transformed clinical practice across numerous medical specialties. The safety profile of ultrasound, combined with its cost-effectiveness, portability, and real-time imaging capability, has established it as a first-line imaging modality for vascular and cardiac assessment. This report explores the scientific foundations, technological components, clinical methodologies, and practical applications of Doppler ultrasound, culminating in a detailed case study demonstrating its diagnostic value in managing peripheral arterial disease.

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## 1. Fundamentals of Ultrasound Technology

### 1.1 Definition and Physical Principles

Ultrasound refers to acoustic waves with frequencies exceeding the upper limit of human hearing, typically above 20 kHz. In medical imaging applications, ultrasound frequencies range from 1 MHz to 15 MHz, with the specific frequency selected based on the target tissue depth and desired image resolution. Ultrasound waves are longitudinal mechanical waves, meaning particles in the medium vibrate parallel to the direction of wave propagation. These waves require a physical medium to travel through—they cannot propagate through a vacuum. In soft tissue, ultrasound travels at approximately 1540 meters per second, a value standardized across ultrasound machines for image reconstruction. The fundamental interaction between ultrasound and tissue creates the basis for diagnostic imaging. When ultrasound waves encounter boundaries between tissues with different acoustic impedances, partial reflection, refraction, and absorption occur. The reflected waves return to the transducer, where they are detected and converted into electrical signals for processing and display.



Figure 1: Modern Doppler ultrasound system used in clinical practice for real-time blood flow assessment



Figure 2: Advanced color Doppler ultrasound system showing comprehensive imaging capabilities

## 1.2 Frequency Selection and Tissue Penetration

The relationship between frequency, image resolution, and tissue penetration represents a fundamental trade-off in ultrasound imaging. Higher frequencies produce images with superior spatial resolution suitable for imaging superficial structures, whereas lower frequencies penetrate deeper into tissue but yield lower resolution images. Medical professionals must select appropriate frequencies based on the specific clinical question and anatomical target.

High-frequency transducers operating between 10 and 15 MHz provide excellent image resolution but exhibit limited penetration, typically reaching only 2 to 3 centimeters beneath the skin. These transducers excel at imaging superficial structures such as the thyroid gland, breast tissue, tendons, and other structures located near the body surface. Conversely, low-frequency transducers operating between 2 and 5 MHz penetrate much deeper, reaching 10 to 15 centimeters or more, making them ideal for abdominal imaging, cardiac evaluation, and obstetric assessment where visualization of deeper structures is essential.

Modern ultrasound systems often incorporate broadband transducers capable of operating across multiple frequency ranges, allowing clinicians flexibility in optimizing image quality for different clinical scenarios. Advanced signal processing techniques including harmonic imaging, spatial compounding, and speckle reduction have further enhanced image clarity and diagnostic accuracy.

Clinical Application	Frequency Range	Penetration Depth	Primary Use
Superficial vessels	10-15 MHz	2-3 cm	Thyroid, tendons, breast
Intermediate vessels	5-10 MHz	5-8 cm	Peripheral arteries and veins
Abdominal organs	2-5 MHz	10-15 cm	Liver, kidneys, pancreas
Deep vascular structures	1-2 MHz	15-20 cm	Large vessels, fetal imaging

Table 1: Relationship between transducer frequency, penetration depth, and clinical applications

## 1.3 Safety Considerations

Ultrasound has established an excellent safety record in medical practice, particularly when compared to ionizing radiation modalities such as X-ray and computed tomography. The mechanical nature of ultrasound, combined with the low energy levels used in diagnostic imaging, eliminates the risks associated with radiation exposure. This safety profile makes ultrasound particularly valuable for serial examinations, pediatric imaging, and fetal assessment where repeated exposure or protection of developing tissues becomes paramount.

The acoustic output power is carefully regulated according to international safety standards established by organizations such as the FDA and international regulatory bodies. Two primary safety metrics guide clinical practice. The Mechanical Index quantifies the likelihood of producing cavitation—the formation of microscopic gas bubbles that could damage tissue. The Thermal Index measures the potential temperature increase in tissue resulting from ultrasound energy absorption.

Clinical practice maintains these indices within proven safe limits during diagnostic ultrasound examination.

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## 2. The Doppler Effect and Its Medical Application

### 2.1 Historical Background and Physical Principle

Christian Doppler, an Austrian physicist, first described the Doppler effect in 1842, noting that observers perceived different frequencies depending on their motion relative to a wave source. He observed that when a wave source and observer move toward each other, the observer perceives a higher frequency as the waves become "compressed," whereas movement away from the source produces a lower perceived frequency as the waves become "stretched." This same principle applies universally to all wave phenomena, including sound, light, and other forms of electromagnetic radiation.

The practical application of the Doppler effect to medical ultrasound emerged decades after Doppler's initial description. When an ultrasound wave of known transmitted frequency encounters moving red blood cells within a vessel, the reflected wave returns to the transducer with a slightly altered frequency. This frequency difference, termed the Doppler shift, directly relates to the velocity of blood flow. By measuring this frequency shift, clinicians can quantify blood flow velocity with remarkable precision, enabling assessment of vascular patency, stenosis severity, and hemodynamic function.

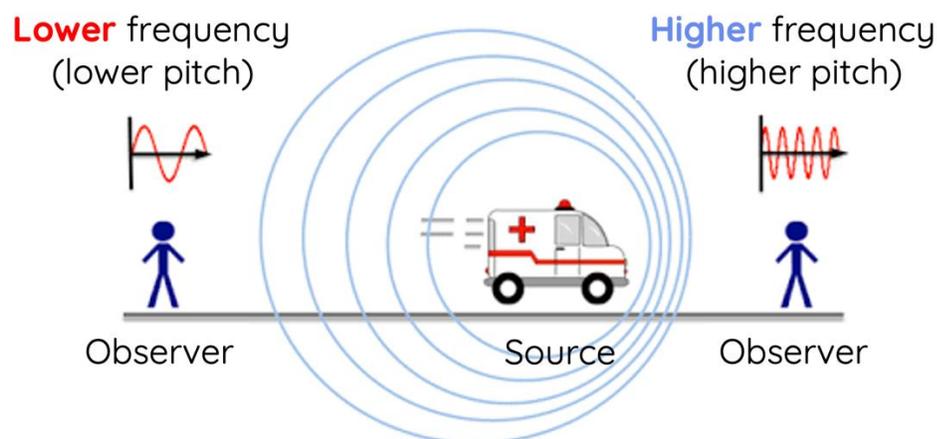


Figure 3: Illustration of the Doppler effect demonstrating frequency compression during approach and frequency extension during recession



Figure 4: Ultrasound machine displaying spectral Doppler waveforms for comprehensive hemodynamic analysis

## 2.2 Mathematical Foundation of Doppler Measurements

The mathematical relationship between the Doppler frequency shift and blood flow velocity forms the foundation for all quantitative Doppler measurements. This relationship demonstrates exquisite sensitivity to velocity changes while also revealing important limitations that clinicians must understand and compensate for during clinical practice.

### The Basic Doppler Equation: Frequency Shift Calculation

The primary equation expressing the relationship between Doppler frequency shift and moving blood is expressed as:

$$\Delta f = \frac{2f_0 v \cos \theta}{c}$$

This equation describes how the transmitted ultrasound frequency undergoes a frequency shift when reflected from moving red blood cells. The transmitted ultrasound beam must first travel to the moving red blood cells, and the reflected beam must return to the transducer. This double encounter with the moving scatterers produces the factor of 2 in the numerator.

The variables in this equation possess specific meaning and measurement constraints:

- $\Delta f$  represents the Doppler frequency shift measured in Hertz, quantifying the difference between transmitted and received frequencies
- $f_0$  denotes the transmitted ultrasound frequency measured in megahertz, typically ranging from 2 to 15 MHz depending on the clinical application
- $v$  represents the velocity of blood flow measured in centimeters per second, with typical arterial velocities ranging from 20 to 200 cm/s

- $\theta$  represents the angle in degrees between the ultrasound beam direction and the direction of blood flow, ranging from 0 to 180 degrees
- $c$  represents the speed of sound in tissue, standardized at approximately 1540 meters per second

The cosine function in this equation introduces a critical dependency on beam-to-flow angle that profoundly affects measurement accuracy. When the ultrasound beam aligns perfectly parallel to blood flow ( $\theta = 0^\circ$ ), the cosine equals 1, and the equation yields maximum accuracy. As the angle increases, the cosine value decreases, reducing the measured frequency shift and producing systematic underestimation of true velocity.

## Velocity Derivation: Converting Frequency Shift to Speed

Clinicians cannot directly measure frequency shifts; instead, ultrasound systems measure frequency shifts and convert them to velocity units using the derived formula:

$$v = \frac{\Delta f \cdot c}{2f_0 \cos \theta}$$

This equation demonstrates how measured frequency shift, converted by physical constants and system parameters, yields blood velocity. The formula shows that velocity measurement depends on precise knowledge of the transmitted frequency and crucially depends on accurate angle correction. The inverse relationship between velocity and the cosine factor creates a mathematical amplification of angle errors, such that small angular errors at higher angles produce large velocity errors.

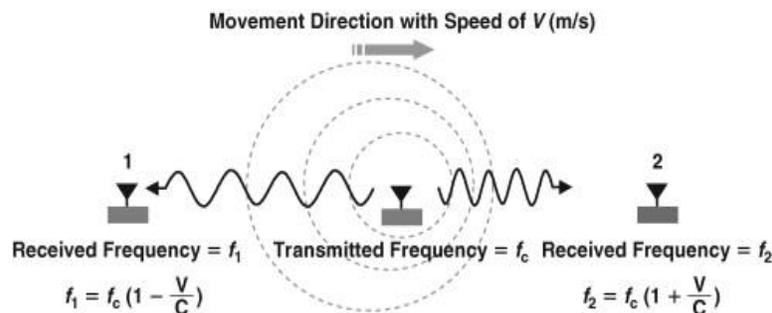
**Clinical Application Example:** Consider measurement of carotid artery stenosis. Suppose the system measures a frequency shift of 4500 Hz using a 5 MHz transducer. Assuming proper angle correction at approximately 0 degrees:

$$v = \frac{4500 \text{ Hz} \times 1540 \text{ m/s}}{2 \times 5,000,000 \text{ Hz} \times \cos(0^\circ)} = \frac{6,930,000}{10,000,000} = 0.693 \text{ m/s} = 69.3 \text{ cm/s}$$

This relatively modest velocity might increase significantly if measured at a 60-degree angle, where  $\cos(60^\circ) = 0.5$ :

$$v_{\text{at } 60^\circ} = 69.3/0.5 = 138.6 \text{ cm/s}$$

This doubling of velocity due to angle error alone demonstrates the critical importance of angle correction.



## Angle Dependency and Measurement Accuracy

The cosine function introduces angle dependency that represents perhaps the most important limitation in clinical Doppler measurements. To illustrate this dependency systematically:

Beam-Flow Angle	Cosine Value	Velocity Error if Ignored	Clinical Significance
0°	1.00	0%	Perfect measurement
15°	0.97	3%	Acceptable
30°	0.87	13%	Clinically significant
45°	0.71	29%	Major error
60°	0.50	50%	Severe underestimation
90°	0	No signal	Complete failure

Table 2: Quantitative relationship between beam-to-flow angle and velocity measurement accuracy

Clinical practice establishes that angle errors should remain below 60 degrees to maintain acceptable accuracy. Modern ultrasound systems incorporate angle correction algorithms that allow clinicians to input the estimated angle, enabling automatic mathematical compensation. However, angle estimation depends on operator skill and image interpretation, introducing an inherent source of measurement uncertainty.

## Bernoulli Equation: Pressure Gradient Estimation

Beyond velocity measurement, the Bernoulli equation enables estimation of pressure gradients across stenotic lesions or across narrowed cardiac valves. This equation expresses the relationship between fluid velocity and pressure:

$$\Delta P = 4 \times (v_{\text{peak}})^2$$

This simplified form of Bernoulli's theorem assumes laminar flow and negligible proximal flow velocity. The constants and exponent derive from fluid mechanics principles. The equation produces pressure in millimeters of mercury when peak velocity is measured in meters per second.

**Clinical Application Example:** A patient with aortic stenosis demonstrates peak velocity across the aortic valve of 4.2 m/s:

$$\Delta P = 4 \times (4.2)^2 = 4 \times 17.64 = 70.56 \text{ mmHg}$$

This pressure gradient exceeds the threshold for moderate aortic stenosis, informing clinical decisions regarding valve replacement timing. The equation's validity depends on meeting several

assumptions, including uniform flow geometry and predominance of convective energy changes over viscous losses, which may not hold in severely turbulent flow patterns.

## Volumetric Flow Rate Calculation

Beyond velocity measurement, clinicians often require quantification of total blood volume flowing through a vessel during a specified time period. This calculation incorporates velocity information with vessel cross-sectional area:

$$Q = V_{\text{mean}} \times A$$

where  $Q$  represents the volumetric flow rate in milliliters per second,  $V_{\text{mean}}$  represents the mean velocity throughout the cardiac cycle, and  $A$  represents the cross-sectional area of the vessel.

The cross-sectional area calculation utilizes the measured vessel diameter:

$$A = \pi r^2 = \pi \left(\frac{d}{2}\right)^2$$

where  $d$  represents the vessel diameter measured in centimeters.

A critical limitation of this calculation emerges from the quadratic relationship between diameter and area. A ten percent error in diameter measurement produces a twenty percent error in cross-sectional area, which directly translates to a twenty percent error in calculated flow volume. This mathematical relationship explains why diameter measurements in Doppler ultrasound require meticulous attention to accuracy. Small errors in diameter estimation produce disproportionately large errors in flow calculations, potentially leading to misinterpretation of hemodynamic significance.

## Resistivity Indices: Assessment of Vascular Resistance

Beyond absolute velocity measurements, ultrasound systems calculate derived indices that characterize vascular properties and blood flow patterns. The Resistive Index quantifies the relationship between peak systolic velocity and end-diastolic velocity:

$$RI = \frac{PSV - EDV}{PSV}$$

where PSV represents peak systolic velocity and EDV represents end-diastolic velocity. This ratio produces a dimensionless number between 0 and 1 that reflects peripheral vascular resistance. Values below 0.5 typically indicate low resistance, values between 0.5 and 0.7 suggest moderate resistance, and values exceeding 0.7 often indicate elevated peripheral resistance potentially associated with vascular disease or organ dysfunction.

The Pulsatility Index similarly characterizes flow patterns:

$$PI = \frac{PSV - EDV}{V_{\text{mean}}}$$

This index compares the amplitude of velocity variation to the mean velocity, providing information about flow pulsatility and waveform morphology.

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## 3. Doppler Ultrasound Modalities

### 3.1 Continuous-Wave Doppler

Continuous-wave Doppler represents the oldest and most straightforward Doppler modality. The technique employs two separate piezoelectric elements within a single probe: one element continuously transmits ultrasonic waves while another element continuously receives echoes. Because transmission and reception occur simultaneously without interruption, the system can detect frequency shifts produced by targets moving at extremely high velocities without experiencing velocity aliasing.

The absence of velocity aliasing represents the primary advantage of continuous-wave Doppler, allowing accurate measurement of very high velocities that would cause aliasing artifacts in pulsed-wave systems. This capability makes continuous-wave Doppler invaluable for cardiac applications requiring measurement of high-velocity jets across stenotic valves or regurgitant orifices. Peak velocities exceeding 6 meters per second can be measured reliably using continuous-wave Doppler without concern about signal aliasing.

However, continuous-wave Doppler sacrifices anatomical localization to achieve this velocity capability. Because the transmitted beam continuously occupies the ultrasound field and echoes return from all depths simultaneously, the system cannot determine the precise depth from which the measured velocity originates. When multiple vessels or flow patterns exist along the ultrasound beam path, the received signal represents a superposition of Doppler shifts from all depths, potentially complicating clinical interpretation. This limitation restricts continuous-wave Doppler use to scenarios where the anatomical location of the target flow is obvious or where the primary clinical goal involves measuring maximum velocity regardless of anatomical localization.



Figure 5: Portable continuous-wave Doppler device demonstrating compact design suitable for bedside evaluation



Figure 6: Clinical portable Doppler ultrasound unit used for rapid vascular assessment in various settings

### 3.2 Pulsed-Wave Doppler

Pulsed-wave Doppler implements a fundamentally different approach that trades velocity range for anatomical localization. The system uses the same transducer element for both transmission and reception, but operates them in an alternating fashion. The transducer transmits short-duration ultrasonic pulses, then listens for echoes returning from tissues. The time interval between pulse transmission and echo reception directly indicates the depth from which the echo originated, enabling precise depth localization through a principle called range gating.

By gating the receiver to accept echoes arriving only during a specific time window, pulsed-wave Doppler can measure blood flow at a particular depth while rejecting signals from other depths. This depth selectivity enables clinicians to interrogate specific vessels or cardiac chambers while excluding nearby structures, greatly enhancing diagnostic specificity. A clinician can position the sample volume within a specific region of interest and measure the Doppler characteristics of flow at that exact location.

The fundamental limitation of pulsed-wave Doppler emerges from the Nyquist sampling theorem, which establishes a maximum velocity that can be accurately measured without aliasing. This maximum velocity, called the Nyquist limit, equals half the pulse repetition frequency. When blood velocity exceeds the Nyquist limit, the system cannot distinguish between true velocity and false lower velocities aliased from higher frequencies. This produces a characteristic artifact in which the spectral display appears to wrap around, with velocities appearing to suddenly reverse direction—an obviously impossible physical situation that alerts the clinician to aliasing.

Despite this limitation, pulsed-wave Doppler remains the workhorse modality for most vascular and cardiac examinations. The combination of depth localization with acceptable velocity ranges for most clinical scenarios makes pulsed-wave Doppler extraordinarily useful for detailed hemodynamic assessment. Skilled operators manage aliasing through multiple technical adjustments including increasing pulse repetition frequency, shifting the spectral baseline, or switching to continuous-wave Doppler when necessary.

### 3.3 Color Doppler Imaging

Color Doppler revolutionized ultrasound practice by providing real-time visualization of blood flow superimposed upon grayscale anatomical images. The technique processes multiple pulse repetitions along each scan line, computing mean Doppler frequency shifts for each image pixel and assigning colors according to a standardized convention. By convention, red colors indicate blood flow toward the transducer while blue colors indicate flow away from the transducer. The saturation or brightness of the color indicates velocity magnitude, with brighter colors typically representing higher velocities.

Color Doppler provides extraordinary clinical value through rapid visualization of normal and abnormal flow patterns. Normal vessels appear as coherent color signals indicating laminar flow in one direction. Stenotic lesions often produce color turbulence—areas of mixed color indicating complex, multidirectional flow. Vascular occlusions appear as complete absence of color signals. Cardiac regurgitation appears as abnormal color jets violating normal anatomical boundaries. This real-time color feedback allows clinicians to rapidly screen large anatomical areas, identify abnormalities, and guide placement of sample volumes for subsequent spectral analysis.

The primary limitations of color Doppler relate to angle dependency and aliasing. Quantitative velocity measurements derived from color information are angle-dependent, similar to spectral Doppler. Additionally, color Doppler exhibits frame rate limitations—the temporal resolution decreases when examining large anatomical areas or when using smaller color box dimensions. Excessive color gain produces overgaining artifacts in which color appears to bleed beyond true vessel boundaries.

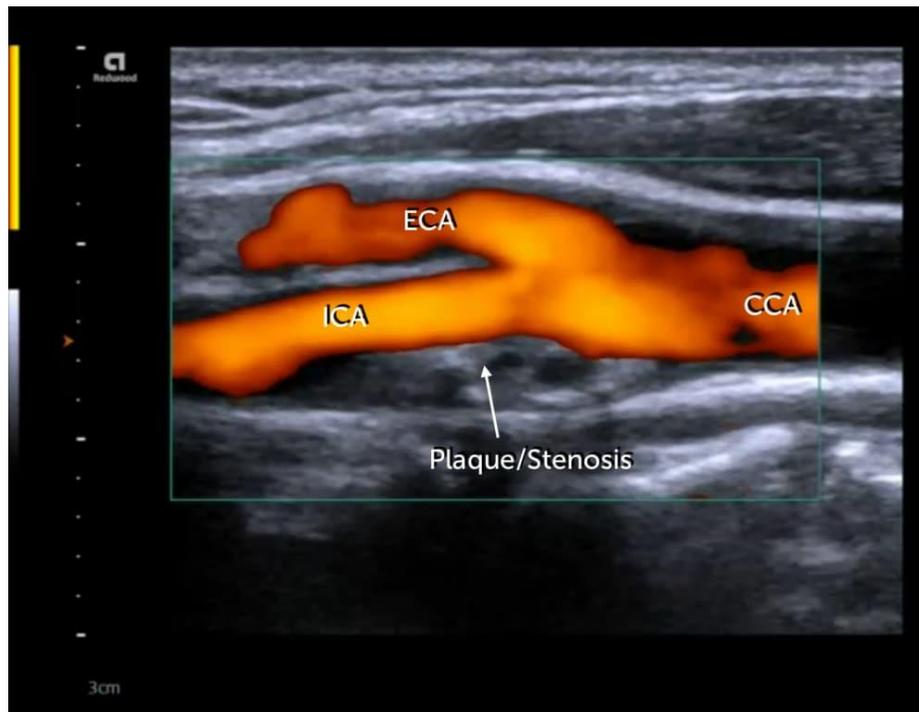


Figure 7: Color Doppler image demonstrating stenotic lesion in carotid artery with characteristic turbulent flow patterns

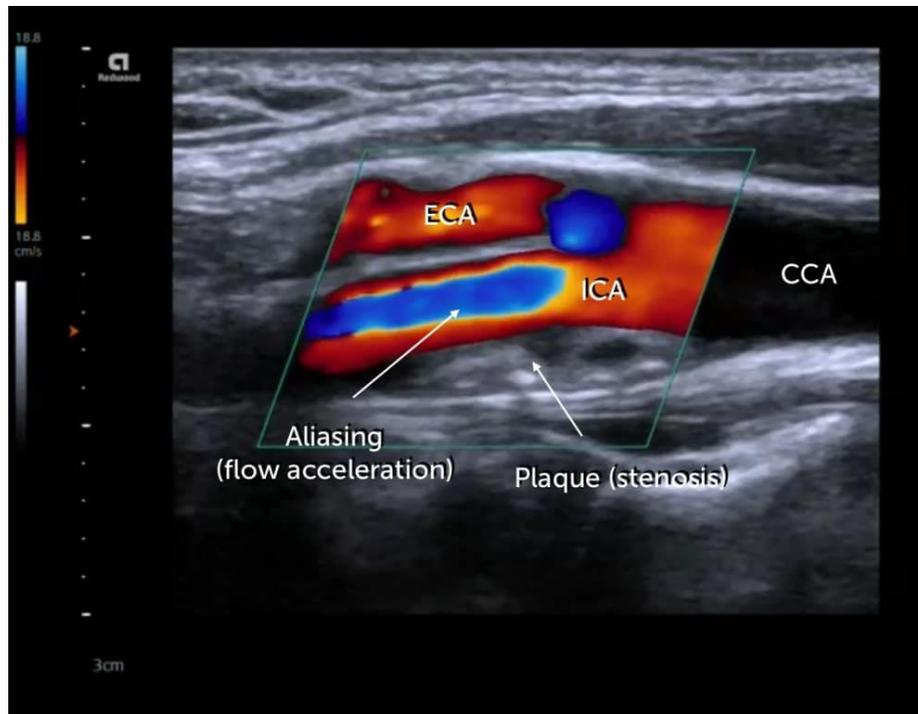


Figure 8: Advanced color Doppler showing detailed carotid artery assessment with stenosis evaluation

### 3.4 Power Doppler

Power Doppler represents an alternative display format that emphasizes the amplitude or energy of Doppler signals rather than frequency shift. The technique integrates the total power within the received Doppler signal, creating an image where pixel brightness indicates the total backscattered energy from moving blood. Because power Doppler does not extract directional information, it cannot indicate whether blood is moving toward or away from the transducer. However, this simplified processing offers distinct advantages in specific clinical scenarios.

Power Doppler demonstrates superior sensitivity to low-velocity blood flow and small-vessel perfusion compared to color Doppler. The enhanced sensitivity reflects the mathematical properties of the power-based algorithm, which emphasizes signal amplitude rather than frequency coherence. This increased sensitivity makes power Doppler particularly valuable for imaging small vessels in tissues, organ perfusion patterns, and vascularity of masses. Additionally, power Doppler exhibits reduced angle dependency compared to color or spectral Doppler, because angle affects frequency shift magnitude less dramatically when examining total integrated power across a broad frequency spectrum.

The absence of directional information limits power Doppler's utility for comprehensive hemodynamic assessment but proves irrelevant for perfusion imaging where detecting flow presence matters more than determining direction. Power Doppler's susceptibility to motion artifacts represents another limitation—patient movement or transducer motion can produce spurious power signals unrelated to true blood flow, potentially creating diagnostic confusion.

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## 4. Instrument Components and System Architecture

### 4.1 The Transducer: Foundation of Ultrasound Acquisition

The transducer represents the fundamental component of any ultrasound system, serving as the essential interface between the patient and the ultrasound machine. This device performs the critical dual function of converting electrical signals into mechanical vibrations that propagate into tissue as ultrasound waves and converting returning echoes back into electrical signals for processing. Modern transducers utilize piezoelectric materials, typically lead zirconate titanate crystals, that generate mechanical vibrations when subjected to electrical potential and conversely generate electrical signals when mechanically deformed by returning ultrasound echoes.



Figure 9: Various transducer types demonstrating different array configurations used for specific clinical applications



Figure 10: Detailed guide to ultrasound transducer types showing linear, curved, and phased array designs

The architecture of transducer arrays creates the image geometry and determines the field of view. Linear arrays produce rectangular images well-suited for imaging superficial structures and small parts such as the thyroid gland and tendons. Curved or convex arrays produce sector-shaped images with wider field of view at depth, making them ideal for abdominal and obstetric imaging. Phased arrays maintain small physical footprints while producing large ultrasound sectors, enabling imaging through small acoustic windows such as intercostal spaces in cardiac imaging.

Transducers operate at specific frequencies determined by their piezoelectric element thickness and material properties. Higher-frequency transducers ranging from 10 to 15 megahertz provide superior spatial resolution but limited tissue penetration, making them appropriate for superficial vessel imaging in the extremities and neck. Lower-frequency transducers operating between 2 and 5 megahertz penetrate deeper but produce lower resolution images, making them suitable for abdominal, cardiac, and obstetric applications. Modern broadband transducers offer flexibility by operating across multiple frequency bands, allowing clinicians to optimize frequency selection for specific clinical scenarios.

## 4.2 Signal Processing and Doppler Analysis

The signal processing unit receives raw electrical signals from the transducer and transforms them into clinically useful information through multiple signal processing stages. Initial amplification increases the magnitude of weak echoes to levels suitable for subsequent processing. Analog filters remove unwanted frequency components, with high-pass filters eliminating low-frequency wall motion artifacts and other noise while preserving blood flow signals in the desired frequency range.

Demodulation converts the high-frequency Doppler signal into a baseband signal containing frequency shift information that can be analyzed and displayed. Spectral analysis employs Fast Fourier Transform algorithms to decompose the received Doppler signal into frequency components, enabling generation of spectral displays showing velocity distribution over time. Color Doppler processing computes mean frequency shifts for each image pixel and assigns colors according to the standardized red-blue convention. Advanced processing features include wall filters that suppress low-frequency signals from tissue motion while preserving blood flow signals, automated gain compensation that maintains consistent signal levels across depth ranges, and angle correction algorithms that mathematically compensate for non-ideal beam-to-flow angles.



Figure 11: Spectral Doppler waveform display showing velocity measurements and cardiac cycle analysis

## 4.3 Display and Measurement Systems

Modern ultrasound display systems present multiple simultaneous viewing modes. The B-mode image provides grayscale anatomical visualization of tissue structures and vessel walls. Spectral Doppler displays present velocity information as time-velocity spectrograms, with vertical axis showing velocity, horizontal axis showing time, and intensity representing signal amplitude. Color Doppler overlays provide real-time visualization of flow direction and velocity variations across the anatomical image. Multiple simultaneous display modes allow comparison of bilateral structures or simultaneous visualization of different vascular beds.

Integrated measurement tools enable rapid calculation of hemodynamic parameters. The system automatically identifies spectral envelope boundaries, from which it derives peak systolic velocity, end-diastolic velocity, and mean velocity. Resistive and pulsatility indices are calculated from these velocity measurements. Vessel diameter measurements enable volumetric flow calculations. Time-velocity integral measurements over one or more cardiac cycles provide information about total flow volume per beat.

## 4.4 Software Integration and Workflow

Contemporary ultrasound systems integrate sophisticated software that manages signal processing, image rendering, measurement calculations, and clinical workflow. Customizable presets allow rapid optimization for different examination types including cardiac, vascular, obstetric, and superficial structure protocols. Data storage functionality captures still images, video loops, and quantitative measurements for documentation and archival purposes. Network connectivity enables integration with hospital imaging systems and telemedicine applications. Advanced systems increasingly incorporate artificial intelligence algorithms for automated vessel detection, stenosis grading, and measurement optimization.

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# 5. Blood Flow Measurement Techniques and Clinical Methodology

## 5.1 Detailed Analysis of Mathematical Formulations

The mathematical foundation of Doppler measurements enables quantitative assessment of hemodynamic parameters with precision previously impossible using clinical physical examination alone. These formulations translate physical principles into practical clinical tools while revealing inherent measurement limitations that require careful operator attention.

### Formula 1: Fundamental Velocity Equation

The fundamental equation relating measured Doppler frequency shift to actual blood velocity forms the basis of all quantitative Doppler assessment:

$$v = \frac{\Delta f \cdot c}{2f_0 \cos \theta}$$

This equation quantifies how the ultrasound system converts a measured frequency difference into velocity units. The numerator contains the measured frequency shift multiplied by the speed of sound, representing the fundamental product of the Doppler principle. The denominator contains twice the transmitted frequency and the cosine of the beam-to-flow angle, factors that emerge from

both the double-encounter principle and the geometric relationship between beam and flow directions.

The practical application of this formula requires accurate measurement or estimation of the beam-to-flow angle. Modern ultrasound systems incorporate angle correction tools allowing manual angle input, which the system uses to automatically correct velocity measurements. However, angle estimation remains operator-dependent and subject to error. Angles should ideally remain below 60 degrees to maintain measurement accuracy, although practical limitations sometimes necessitate accepting higher angles while recognizing the resulting velocity underestimation.

**Detailed Clinical Example:** Consider assessment of carotid artery stenosis. The ultrasound system measures a Doppler frequency shift of 6000 Hertz using a 7 MHz