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DOPPLER ULTRASOUND TECHNIQUES IN MEDICINE

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Abstract:
Doppler ultrasound is a noninvasive technique which is widely used in medicine for the assessment of blood flow in intact vessels. The technique has improved much since Satomura first demonstrated the application of the Doppler effect to the measurement of blood flow velocity in 1959. However, rigorous analysis of their properties did not begin until the mid-1970s. Since then, Doppler assessment of blood flow has become routine in many diagnostic ultrasound exams. The analyses were motivated by a desire to extract specific physiologically relevant information with the devices. Measurements of interest included (1) flow rate, (2) velocity profiles, (3) coherent structures, (4) turbulent energy and turbulent spectra, (5) velocity gradients (shear rate), and (6) pressure drops. Inaccuracies in all of these measurements result from fundamental limitations in the Doppler ultrasound method itself. The instruments measure neither true flow nor point velocity. However, they provide a measure of the velocity distribution throughout the interrogated volume, and this unique aspect has suggested to researchers that the spectral content of the quadrature signals could be correlated to the severity of flow pathologies such as arterial stenosis and aneurism.

Ключевые слова:
Doppler ultrasound, Noninvasive technique, Blood flow, Spectral content, Velocity measurements

1. INTRODUCTION

The use of Doppler today ranges from assessing blood flow in the fetus and umbilical cord, to flow patterns through valves in the heart or monitoring of blood flow to the brain [1-6]. A variety of instruments is available, ranging from simple pocket-size versions that give an audio output, to more sophisticated imaging systems that integrate blood velocity measurements with images of anatomy. Even these high-performance machines are portable and yet still give diagnostic information in real time. This information may be used subjectively, to assess anatomy or the presence of flow, or to quantitatively assess physiological parameters.

The technical challenges of designing a machine that is capable of meeting the citeria of reasonable cost (when compared to other imaging modalities), mobility, and processing speed are high. This complexity is compounded by the fact that blood velocities in the human body can range from several cm/s in small vessels feeding tumors to over 10 m/s through a stenosed valve in the heart. As many systems are now designed to be used in different clinical applications, there must be sufficient user control so that this wide variety of flow states can be assessed. Ultrasound is also no longer simply a transcutaneous exam. Transvaginal, transesophageal probes, and intravascular exams are now fairly routine and intraoperative ultrasound is becoming more commonplace [7].

Doppler methods are unique among clinical techniques in ultrasound in that they have the potential to offer information related to the function of an organ rather than its morphology. However, they have in common with all ultrasound techniques that the information is derived from the interaction of a beam of sound with a volume of tissue and must therefore represent a combination of these two influences. Doppler systems are based on the principle that ultrasound, emitted by an ultrasonic transducer, is returned partially towards the transducer by the moving red blood cells, thereby inducing a shift in frequency proportional to the emitted frequency and the velocity along the ultrasound beam. Much of the interpretation of Doppler signals in clinical practice entails the extraction of information about the underlying blood flow from confounding factors related to the Doppler technique. This process has been made progressively more straight-forward with the refinement of instruments for the acquisition and analysis of Doppler signals. However, the mere fact that the data cannot be presented as a conventional image can often bewilder the sonographer who relies on an intuitive interpretation of an ultrasound study. An appreciation of the physical principles of the Doppler effect not only help extend such an intuition into blood flow studies, but is an essential prerequisite for the quantitative interpretation of Doppler signals [1-6]. In order to better understand how these multiple requirements are met, this study looks at how the Doppler effect is applied in commercial systems, its clinical uses, and reseach developments.

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2. THE DOPPLER EFFECT

When an observer is moving relative to a wave source, the measured frequency is different from the emitted frequency. If the source and observer are moving towards each other, the observed frequency is higher than the emitted frequency; if they are moving apart the observed frequency is lower. This simple but important assertion is known as the Doppler effect and was described by Christian Doppler in 1842 [1]. When a wave is reflected from a moving target, the frequency of the wave received differs from that which is transmitted. This difference in frequency is known as the Doppler shift and depends on, among other things, the speed at which the target is moving and whether the motion is to-

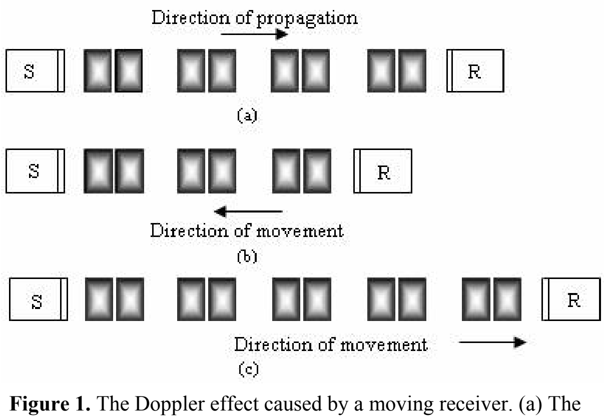

Figure 1. The Doppler effect caused by a moving receiver. (a) The source (S) and receiver (R) are stationary and the frequency received is equal to that transmitted. (b) The receiver is moving toward the source, thus encountering a greater number of compressions per second and detecting a higher frequency. (c) The receiver is receding from the source: fewer compressions per second reach it and consequently a lower frequency is detected

ward or away from the receiver. The Doppler effect occurs whenever there is relative motion between the source and the receiver of sound. Consider the case in which the source is stationary and the receiver is moving toward the source. Sound waves, comprising a series of compressions, travel toward the receiver at a steady speed determined by the medium. The frequency received is simply the number of these compressions detected per second by the receiver. In the example given in Figure 1(a), in which both the source and receiver are stationary, this is obviously equal to the frequency that is transmitted. Figure 1(b) shows that if the receiver moves toward the source, it will detect more compressions per second and so register a higher frequency. Conversely, if the receiver moves away from the source (Figure 1(c)), fewer compressions reach the transducer per second and a lower frequency is detected. A precisely analogous effect occurs if the source moves away from a stationary receiver.

In the case of ultrasound being scattered from moving red blood cells (Figure 2), two successive Doppler shifts are involved. First, the sound from the stationary transmitting transducer is received by the moving red blood cells. Second, the cells act as a moving source as they reradiate the ultrasound back toward the transducer, which is now a stationary receiver. To a first approximation, these two Doppler shifts are equal and simply add to each other. This account for the factor 2 appearing in the Doppler equation,

$f_D=2fv\cos\theta/c$. (1)

This equation relates the Doppler shift frequency f_D (measured in Hz) to the velocity of the moving blood v (in m/s), the frequency of the ultrasound f (in Hz), the velocity of sound c in the medium (in m/s), and the cosine angle θ between the direction of motion and the axis of the ultrasound beam. This angle θ enters the equation because it is seldom that a target, such as blood within a vessel, is moving directly toward or away from the transducer. More generally, it will be moving in a direction at some angle θ to the line between it and the transducer. The Doppler effect is a consequence only of motion along this line. It is therefore necessary to calculate the component of the velocity v along the direction of the ultrasound beam: this is given by $v\cos\theta$. In the extreme case in which the motion is aligned precisely with the beam, the angle θ is equal to 0, and cos0 is equal to 1, so that the component of velocity responsible for the Doppler shift is simply v. Conversely, if the motion is perpendicular to the beam, θ is equal to 90° and cos90° is equal to 0, so that there is no component of velocity along the beam and hence no Doppler shift. In physical terms, it is easy to see that the target is neither approaching nor receding from the transducer in this case.

The case of a single scatterer moving at a velocity, v, and at an angle, θ, with respect to the continuous wave

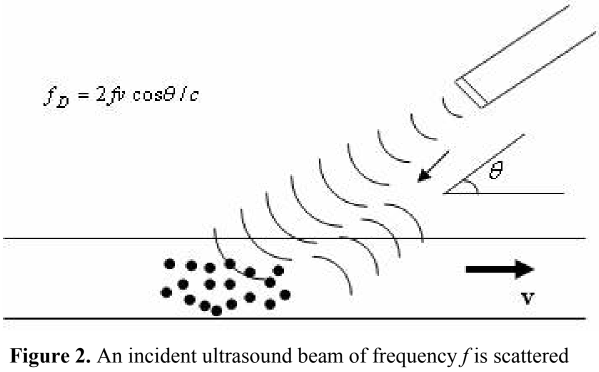

Figure 2. An incident ultrasound beam of frequency f is scattered by moving red blood cells. As a result of the Doppler effect, the received backscattered echo has a frequency that is higher by f_D. v, velocity; f, frequency of interrogating sound beam; θ, angle between sound beam and flow axis

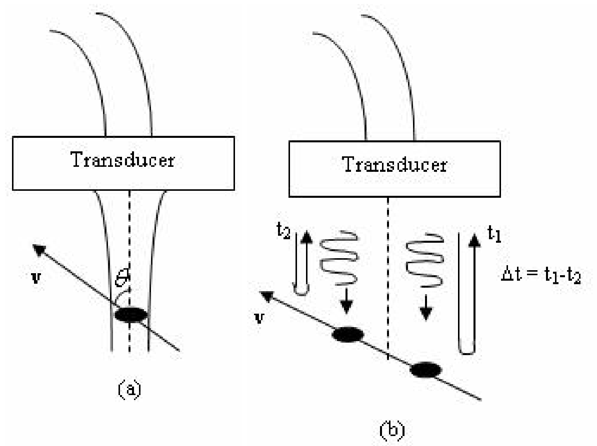

Figure 3. Measurement of axial velocity of a single scatterer: (a) Continuous wave insonation; (b) Pulsed wave insonation. Δt is the time difference between the return from successive pulses

insonation is shown in Figure 3(a). As the majority of Doppler systems work in a pulsed mode, it is also useful to derive the Doppler shift in the time domain. The velocity can be calculated from the time difference (Δt) between the signal return from successive pulses (Figure 3b) [7]. For a given pulse repetition interval (PRI), the axial velocity of a single moving scatterer is:

$$v\cos\theta = \frac{c\Delta t}{2PRI} \qquad (2)$$

Combining equations (1) and (2), and the fact that frequency can be expressed as a derivative of phase with respect to time, f_D can also be expressed as:

$$f_D = \frac{\Delta\varphi}{PRI} \qquad (3)$$

where $\Delta\varphi$ is the phase shift between the two consecutive received signals. The assumption of a single scatterer moving at a single velocity is obviously not valid in practice. This means that rather than a single frequency, a spectrum of frequency shifts are detected, which represent the motion within the area being sampled.

The most common use of Doppler is in the detection of moving blood. Red blood cells (RBC) or erythrocytes are assumed to be the primary source of scattering [1]. This is due to their size and number relative to the platelets and white blood cells and also to their impedance mismatch with the surrounding plasma. However, the scattering can still be assumed to be weak (i.e., no multiple reflections) and the backscatter from blood is on the order of 40 dB less than that from tissue. Individual red cells are around 6 mm in diameter and are described as semi-rigid biconcave discs. As typical diagnostic frequencies lie in a range 2 to 10 MHz (a wavelength of 0.3 mm at 5 MHz), individual cells cannot be resolved. At low shear rates, the erythrocytes join together to form rouleaux which are broken up as the shear rate is increased. The distribution of the red cells and rouleaux throughout the vessel also varies with flow condition and with the volume concentration of red cells (hematocrit). The effect of these hemodynamic variables on the Doppler signal has been observed experimentally and there are several theories as to the mechanism behind the scattering. Doppler angle and Doppler shift frequency are used in estimating velocity of blood flow and the size of potential errors induced by an uncertainty in the angle measurement should be considered. On the other hand, at angles less than 20° these errors are reduced to insignificance.

The other common use of Doppler is in the measurement the volume of blood flow. The two techniques used most widely have limited success due to their implicit assumptions, the one common to both being that the vessel under investigation is circular in cross-section. The first method uses a pulsed wave sample volume that is large enough to completely insonate the vessel. By calculating the intensity-weighted mean velocity through the whole vessel, v, and measuring the diameter of the vessel, d, from the B-mode image, the volume, Q, is calculated from:

$$Q = \pi \frac{d^2}{4} \bar{v} \tag{4}$$

In practice, the insonation of the vessel is rarely uniform, so the expected variance in this method is fairly large.

The second method makes the opposite assumption that the beam is so narrow that individual velocity estimates made across the vessel can be used to determine the flow profile or variation of velocity across the vessel. If the velocity flow is axi-symmetric and the annulus for each velocity estimate (diameter Δr) is sufficiently small, the volume can be calculated from:

$$Q = \sum_{i=1}^{N} \pi v_i r_i \Delta r \tag{5}$$

where v_i is the velocity estimate at each radial location, r_i. This method has been implemented in a flow imaging system with a color M-mode display. In this case, the velocity estimates are also used to estimate the extent of the vessel [1].

3. INTERPRETATION OF THE DOPPLER SPECTRUM

The Doppler signal is conventionally interpreted by analyzing its spectral content. Diagnosis and disease monitoring are assessed by analysis of spectral shape and parameters. When evaluating minor stenosis problems, the spectral mean frequency and bandwidth are relevant parameters for diagnosis. The former is directly proportional to the blood flow velocities. The latter, due to several causes, namely blood velocity gradient and change on velocity directions within the considered sample volume, is related to blood flow disturbance [1-4,8].

Presence of flow: Determining whether or not flow is present is one of the simplest but perhaps most useful applications of Doppler. It may be used to exclude occlusion by thrombosis, for example, or to determine the point of occlusion of a vessel during a pressure measurement. Confirming the absence of flow is a little more difficult. It is necessary to be certain that the lack of Doppler signal is a consequence of lack of flow, rather than the acoustical or electrical parameters of the system.

Direction of flow: The directional resolution of the Doppler system is best when the beam-vessel angle is relatively small and the signal lies in the middle of the dynamic range of the receiver. If a mirror image of the signal is seen in the reverse flow sideband of the display, care should be taken with the adjustment of angle and receiver gain so as to obtain a trace showing an unambiguous direction.

Identification of characteristic flow: It is apparent that normal flow in various parts of the arterial circulation shows distinct characteristics according to its precise location. Those characteristics of waveform shape and spectral distribution are often consequences of hemodynamic factors unique to each vessel. These allow the identification of the origin of a flow signal from the spectral display alone and is particularly useful in circumstances where the image may be ambivalent.

Table 1.
Some indices used to describe the shape of the waveform

Pulsatility index (PI)	*PI=(S-D)/M*
Resistivity index (RI)	*RI=(S-D)/S*
S/D ratio	*S/D* rewritten as *S/D=1/(1-RI)*
Constant flow ratio (*CFR*)	*CFR=DT/A*
Height width index (*HWI*)	*HWI=PI(T/t$_5$)*

where the variables S, D, M, A, T, t_5 are defined in Figure 4.

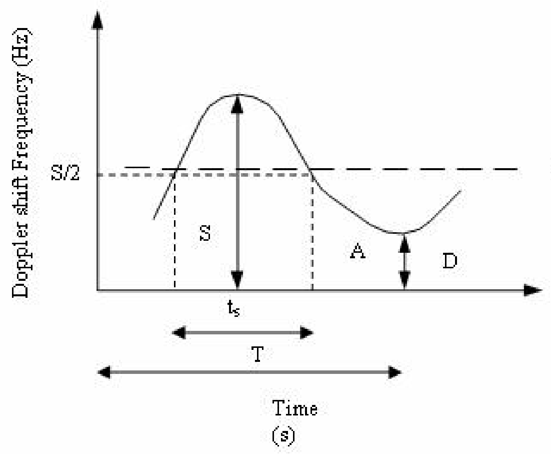

Figure 4. Diagram illustrating the variables involved in the definitions of pulsatility index, resistance index, *S/D* ratio, constant flow ratio and height width index. *S* is maximum systolic height , *D* is end diastolic height and *M* is mean height over the cardiac cycle, *T* the length of the cardiac cycle, *A* the area under curve, and t_5 the duration of the systolic peak (measured between half amplitude points)

Disturbed flow: The characteristics that distinguish disturbed and turbulent flow from laminar flow are reflected in the content of their respective Doppler signals. In a vessel with a flat velocity profile, the spectral display shows a narrow range of Doppler shift frequencies, especially in systole. This is the origin of the "window" below the spectral trace in systole – a Doppler sign of laminar flow with a blunt profile. Slower moving laminae are found toward the edge of the vessel, so the size of the window can be increased by employing a small sample volume situated toward the center of the vessel. Flow disturbance produces velocity vectors whose direction varies. This means that the components of velocity along the direction of the Doppler beam change with time. The combination of many such components results in a wide range of Doppler shifts, seen as a broadening of the spectral display and as a reduction in the size of the window. As the velocity increases, vortices form and the Doppler sample volume encounters rotating flow elements. Velocities, and hence Doppler shifts, are noticeably higher. The vortices contain simultaneous forward and reverse flow in a range of velocities. Suddenly, laminar flow gives way to rotating flow, and the Doppler spectrum shows simultaneous forward and reverse velocities together with a broad range of Doppler shifts. Laminar flow reappears in diastole as the Reynolds number is reduced. Then, increased velocity, spectral broadening, simultaneous forward and reverse flow, and fluctuations of flow velocity with time are Doppler features of disturbed flow.

Waveform analysis: Doppler shift signal can be processed to achieve either a flow velocity waveform or a Doppler power spectrum. Clinically useful information can be extracted from these types of output. Abnormal Doppler waveforms can be recognized by analyzing waveforms, even if a particular physiological or pathological change gives rise to a variation in waveform shape. The Doppler shift signal that results from the demodulation process contains a wealth of information about blood flow occurring within the sample volume of the Doppler velocimeter. The most complete way to display this information is to perform a full spectral analysis and present the results in the form of a sonogram. A number of parameters related to the blood flow may be extracted from the sonogram and these are of high clinical value. Although such a display may be of great value for assessing the general quality of the signal and for making qualitative statements about disease state, it contains so much information that some feature extraction must take place before quantitative statements may be made. Feature extraction is a process of pattern recognition and consists of extracting and combining salient features of the pattern vector into a feature vector (for example resistivity index, pulsatility index). Then decision is given whether such a feature vector is obtained from a normal or abnormal artery. Various indices derived from the waveforms defined resistance index (RI), pulsatility index (PI) and these can yield information that correlates closely to the disease present. Specifically, maximum systolic height, end diastolic height, mean height of the waveform (Figure 4) are often used in the calculation of several indices which are given in Table 1. Both RI and PI are a reflection of the resistance to flow, downstream from the point of insonation. Great care must be taken with interpretation of RI and PI in the clinical setting as they are influenced by many factors including proximal stenosis, post stenosis and peripheral resistance. As it is not necessary to know the Doppler angle when calculating one of these indices, pulsatility can be assessed in vessels too small or tortuous to be imaged (e.g., the arcuate vessels of the kidney). For any measurement reflecting the shape or spectral content of the Doppler waveform, the index should be calculated for each of several cardiac cycles and an average value taken. Beat-to beat variation in the adult suggests that about five heartbeats is adequate for a measurement of pulsatility [1-5,8].

4. CONCLUSIONS

Doppler ultrasound is commonly used in cardiology, obstetrics, neonatology, ophthalmology, and the diagnosis of peripheral vascular stenosis. The objective of Doppler ultrasound in diagnosis is to obtain measurements of flow velocity and interpret them in terms of physiologically significant variables. In general, these variables are not measured directly by ultrasound, but must be derived from velocity measurements, supplemental measurements, and assumptions. The most fundamental quantity of interest is flow rate, because this indicates how well an organ or region is perfused by blood. Flow rate can be obtained from multiple measurements of velocity over the cross section of a vessel. These measurements are then integrated over space and averaged over time. It can also be measured by the uniform insonification method, in which spatially averaged velocity is obtained from a single measurement with a wide ultrasound beam and multiplied by the cross-sectional area. Velocity signal analysis is used in the assessment of peripheral arterial, cerebrovascular, and venous disease. Systolic blood pressures permit semi-quantification of the presence, location, and functional extent of peripheral arterial occlusive disease. Although measurements of blood velocity are routinely used in diagnosis, another useful measurement may be the volume of blood flow. The blood volume flow is a quantity of interest to clinicians because it allows judgements to be made about blood supply and tissue perfusion.

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СОВЕРШЕНСТВОВАНИЕ ОРГАНИЗАЦИИ МЕДИЦИНСКОЙ ПОМОЩИ ТРУДОСПОСОБНЫМ ЖИТЕЛЯМ РЕСПУБЛИКИ КОМИ

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Аннотация:

Нами были изучены результаты прохождения периодических медицинских осмотров работниками предприятий республики в 2003 -2007 гг. Установлено, что удельный вес профессиональной патологии у рабочих, выявленных при проведении периодических медицинских осмотров на территории РК, ежегодно растет. Медицинская и социальная реабилитация профессиональных больных проводятся не в полном объеме, имеются резервы для сохранения дальнейшей трудоспособности заболевших. Для совершенствования оказания профпатологической медицинской помощи необходимо расширять возможности республиканского центра профпатологии по лечению и реабилитации данного контингента, что будет способствовать снижению профзаболеваемости (за счет первичной профилактики), а также осуществлять мероприятия по усилению контроля за проведением периодических медицинских осмотров.

Ключевые слова:

профессиональные заболевания, периодические медицинские осмотры, Республика Коми