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●*Original Contribution*

CHANGE IN AMPLITUDE DISTRIBUTIONS OF DOPPLER SPECTROGRAMS RECORDED BELOW THE AORTIC VALVE IN PATIENTS WITH A VALVULAR AORTIC STENOSIS

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Abstract—Amplitude distributions of Doppler spectrograms were characterized in a group of 22 patients having no aortic pressure gradient and another group of 26 patients having a stenotic aortic valve. Specifically, for each patient, the ratios of the mean amplitude in three normalized frequency bands (low, middle and high) to the mean amplitude of the Doppler spectrogram computed in selected portions of the systolic period were considered. Pulsed-wave Doppler spectrograms were recorded by positioning the sample volume in the left ventricular outflow tract, approximately 1 cm below the aortic valve. Statistically significant differences were found between the middle (p = 0.041) and high (p = 0.028) frequency bands of Doppler signals recorded from the two groups of patients. The differences observed are believed to be attributed to blood flow eddies generated below the stenotic aortic heart valve and to changes in blood flow orientation.

Key Words: **Doppler ultrasound spectrum, Amplitude distributions, Attenuation of backscattered ultrasonic signals, Blood flow eddies, Left ventricular outflow tract, Pulsed-wave transmission, Aortic valve stenosis.**

INTRODUCTION

Using continuous-wave (CW) and pulsed-wave (PW) Doppler echocardiography, patients with aortic valve stenosis can be identified by detection of high flow velocities across the valve in systole. An inherent limitation of the PW technique is its limited transmitted repetition frequency caused by the fact that time is needed to allow each burst of the ultrasonic signal to travel from the surface of the body to the blood flow under investigation. This limited pulse-repetition frequency (PRF) of the ultrasonic beam reduces the frequency bandwidth of the Doppler spectrum and thus maximal blood flow velocity that can be detected. In cardiac applications, this limitation has led most investigators to use CW technique to estimate maximal blood flow velocity of the aortic jet (Hatle and Angelsen 1985; Zoghbi 1988). Using the value of maximal blood flow velocity, aortic pressure gradient can be estimated with the modified Bernoulli equation. Neglecting viscous and inertial effects in the original form of the Bernoulli equation (see Caro et al. 1978), the pressure drop through the valve (dP) is given by:

$$dP \approx p(V_2^2 - V_1^2)/2, \qquad (1)$$

where p is the density of blood ($\approx$1060 kg/m^3), V_2 the velocity in the aortic jet (m/s), and V_1 the velocity in the left ventricular outflow tract (m/s). Because V_2 is significantly higher than V_1 in severe aortic stenosis, V_1 is generally neglected in practical situations. The modified Bernoulli equation used in clinical evaluations of patients with aortic valve stenosis is then reduced to $dP \approx 4\ V_2^2$, where dP is the pressure drop in mmHg (1 mmHg = 133.3 kg/m $\cdot$ s^2).

Very interesting results were reported in the literature concerning the use of the modified Bernoulli equation. Correlations generally greater than 80%

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were measured between pressure gradients estimated with cardiac catheterization and those evaluated with the modified Bernoulli equation (Stamm and Martin 1983; Berger et al. 1984; Currie et al. 1985; Panidis et al. 1986; Fan et al. 1988). The most interesting results (correlation of 93%) were reported by Currie et al. (1985) on 100 patients and were obtained by simultaneously measuring pressure gradients by catheterization and Doppler techniques. Despite these good clinical results, inherent limitations of the Bernoulli equation must be considered.

The most difficult problem is certainly the localization of the aortic jet. Because conventional ultrasonic systems do not generally simultaneously combine CW Doppler analysis with B-mode imaging, it can be very time consuming to measure the maximal velocity of the aortic jet without any visual guidance. Another problem that should be considered is the orientation of the aortic jet in severe aortic valve stenosis. The jet can be off-axis and oriented directly toward the wall of the aorta. This particularity certainly increases difficulties associated with the detection of maximal blood flow velocities. Sampling blood flow from multiple sites is often necessary and detection of the maximal velocity of the jet is not guaranteed. In addition, even when the optimal site is selected, an underestimation of the gradient is often obtained in some patients because morphology of the thorax and spatial orientation of the heart do not allow reduction of the angle between the ultrasound beam and the direction of the aortic blood flow (Berger et al. 1984). Recently, color-guided Doppler echocardiography has been proposed to circumvent problems associated with the detection of the jet. This technique reduces the time needed to perform the clinical examination but does not allow detection of the jet in some patients (Fan et al. 1988). In the study of Fan et al. (1988), the aortic stenotic jet was not detected with the color Doppler technique in 39 of the 100 patients analyzed. In those patients, the correlation between the pressure gradient estimated with cardiac catheterization and Bernoulli equation was only 70%. In the study of Stamm and Martin (1983), only 81% of the 35 patients had an adequate CW Doppler examination (the color Doppler technique was not available in this study).

Because of these difficulties associated with the CW Doppler technique and because the accuracy of the Bernoulli equation is reduced in patients with concomitant aortic regurgitation (Panidis et al. 1986; Danielsen et al. 1988) and atrial fibrillation (Panidis et al. 1986), complementary techniques allowing the detection of aortic valve stenosis would be interesting. Recently, Penn and Dumesnil (1988) proposed an alternative to the estimation of the aortic pressure gradient by the Bernoulli equation. They estimated the pressure gradient by extrapolating to their point of intersection the maximal accelerating and decelerating velocities of the PW Doppler spectrograms recorded in the left ventricular outflow tract. In the present study, an alternative approach based also on the characterization of the blood flow located below the aortic valve is proposed. The objectives of our study are to demonstrate that:

(a) diagnostic information concerning the status of the aortic valve can be found in amplitude distributions of PW Doppler spectrograms, and
(b) diagnostic information can be found from blood flow located in the left ventricular outflow tract.

THEORY

Although, based on our knowledge, no study reported diagnostic information extracted from amplitude distributions of Doppler spectrograms, several factors influencing the amplitude of backscattered ultrasonic signals are well known. Factors that may modify the amplitude of cardiac Doppler signals are summarized in Fig. 1 and discussed briefly in this section.

Ultrasound waves propagating through the human body are attenuated with increasing distance and frequency of the transmitted wave. Typically, the power intensity of ultrasound decreases along z direction according to the equation:

$$I(z) = I(0)e^{-2\alpha(f)z}, \tag{2}$$

where $I(0)$ is the intensity at $z = 0$, $\alpha(f)$ the frequency-dependent attenuation coefficient in dB/cm, and f the frequency of the transmitted wave usually ranging between 1–10 MHz. The dominant process involved in the attenuation of ultrasound in biological tissues is absorption as demonstrated in liver by Parker (1983). As reported by Johnston et al. (1979), the absorption phenomenon is intimately related to the protein content of tissues and fluids.

Other factors may influence the amplitude of the backscattered signal from blood. The myocardial ultrasonic backscatter was shown to be related to the contractional state of the heart. A cycle-dependent backscattering signal was found with elevated amplitude at end-diastole and significantly lower amplitude at end-systole (Madaras et al. 1983; Fitzgerald et al. 1986). Recent studies also showed the dependence of ultrasonic backscatter on the angle between the ultra-

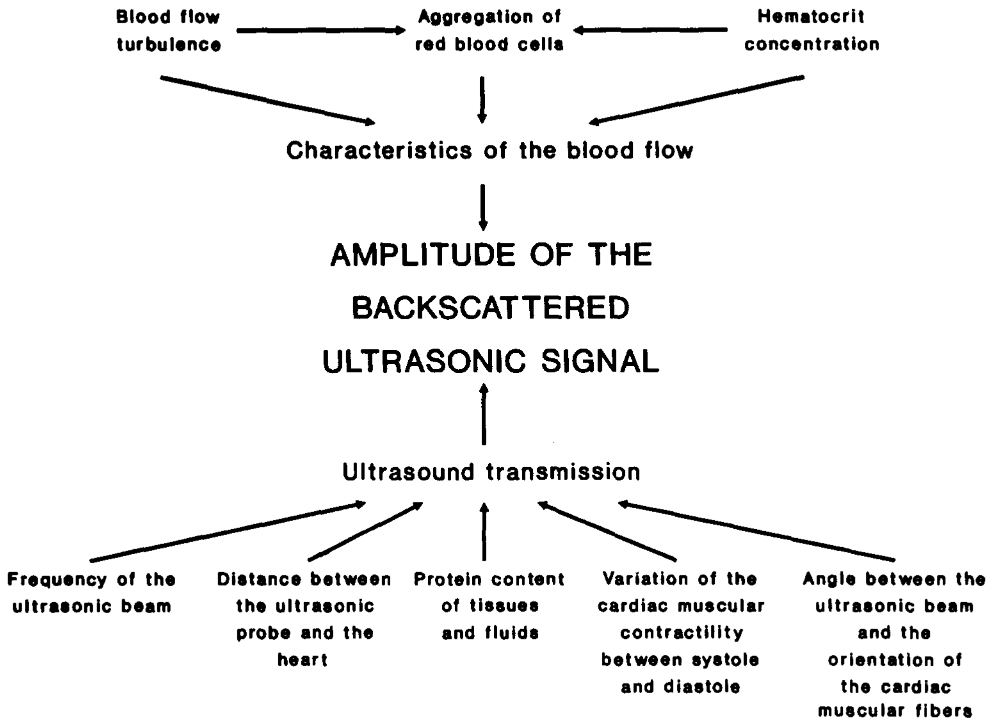

Fig. 1. Diagram showing the interaction of factors affecting the amplitude of cardiac backscattered ultrasonic signals.

sonic beam and the orientation of cardiac muscular fibers (Mottley and Miller 1988; Madaras et al. 1988).

Ultrasonic backscattering from blood

It is now generally accepted that erythrocytes or red blood cells are the major source of ultrasonic scattering from blood (Atkinson and Woodcock 1982; Hatle and Angelsen 1985). Doppler blood flow velocity can be estimated because of acoustic impedance mismatch between red blood cells and the surrounding plasma. The amplitude of the reflected scattering signal increases with the mismatch in acoustic impedance (Atkinson and Woodcock 1982). In a theoretical model of ultrasound scattering from blood, Angelsen (1980) attributed the scattering to local fluctuations in cell concentration. According to his theoretical model, the amplitude of the ultrasonic signal should increase in turbulent blood flow because fluctuations in cell concentration increase. This was verified in an experimental model by Shung et al. (1984) where ultrasonic backscatter from erythrocyte suspensions increased appreciably in the presence of blood flow disturbances. Recently, a unifying theory based on conventional particle and continuum models was proposed by Mo (1990). The backscattered power from blood is believed to be proportional to the variance of local hematocrits within elemental acoustic voxels.

Other factors modifying the amplitude of the backscattered ultrasonic signal from blood were reported in the literature. Experimental work carried out by Shung et al. (1976) on stationary suspensions of erythrocytes in plasma showed that the backscattering coefficient increases with the hematocrit up to a maximum and then decreases as the hematocrit further increases. Other experimental studies confirmed this relation between the amplitude of the backscattered signal and hematocrit concentration in flowing whole blood (Borders et al. 1978; Yuan and Shung 1988a).

Another well-described factor that modifies the amplitude of the backscattered signal is aggregation of red blood cells and the formation of rouleaux. Studies conducted in the field of hematology revealed that erythrocytes suspended in plasma form few if any aggregates in normal blood (Miale 1982). However, in abnormal blood, rouleaux formation may take place and depends on the protein composition of plasma, particularly with regard to fibrinogen and globulin (Hint and Arfors 1971; Miale 1982). In flowing whole blood, red cell aggregation depends also on hematocrit and flow conditions. At low shear rate, erythrocytes form aggregates that disintegrate when the shear rate increases. Interesting observations on the breakup and deformation of linear chains of liquid, metal and polystyrene spheres in relation to red blood cell aggregation were reported by Goldsmith and Mason (1968).

A stochastic model of the continuous-wave Doppler ultrasound signal backscattered from undisturbed laminar blood flow was developed by Mo and Cobbold (1986) and was used to study the influence of rouleaux formation on the Doppler spectrum. In their study, they considered that small aggregations could be found in laminar blood flow. Results showed that red blood cell aggregation does not influence the shape of the spectrum but affects its magnitude. Recently, in an animal experiment where red blood cell aggregation was controlled by fibrinogen addition, the theoretical results of Mo and Cobbold (1986) were confirmed by Yuan and Shung (1988b) in flowing whole blood. The influence of the shear rates on the amplitude of the backscattered signal was also studied in relation to red blood cell aggregation (Yuan and Shung 1988a).

MATERIALS AND METHODS

Patient selection

A group of 48 patients referred to the Hôtel-Dieu de Montréal Hospital for a catheterization examination was included in the present study. Twenty-two patients had no aortic pressure gradient and 26 had aortic valve stenosis. The patient population having no aortic pressure gradient was composed of patients referred for pure aortic regurgitation or coronary artery disease.

Measurements of transvalvular maximal and mean pressure gradients were based on retrograde left-heart catheterization. Both pull-back and simultaneous recording methods were used to estimate aortic pressure gradients (Grossman 1986). When measuring maximal transvalvular gradients, the maximum difference was used. Supravalvular aortography was done in all patients to assess the severity of possible aortic regurgitation on a four-degree scale (minimal, mild, moderate and severe). Visualization of the regurgitant jet by cineangiography from a left anterior oblique projection served to estimate the severity of valvular insufficiency according to the criteria proposed by Grossman (1986). Cardiac output measured by the dye-dilution technique was also evaluated for all patients with suspected aortic valve disease.

Doppler data acquisition

A Doppler investigation was performed generally within one day of the catheterization examination. During this investigation, the following signals were recorded on a four-channel tape recorder for a period of approximately 30 s: the in-phase and quadrature Doppler signals, the phonocardiogram (PCG) recorded from the second right intercostal space (aortic area) and the electrocardiogram (ECG). A TASCAM 22-4 tape recorder having a dynamic range of 40 dB was modified to include two frequency modulated channels. Both Doppler signals were recorded in direct mode while the PCG and ECG signals were recorded in frequency modulation mode. At a speed of 7.5 in/s, the frequency response of the direct channels was uniform (−3 dB) between 50–20000 Hz and that of the frequency modulated channels was uniform (−3 dB) between 0–2500 Hz. The phonocardiogram, used in a previous analysis (Cloutier et al. 1990a), was not considered in the present study.

A duplex ultrasound scanner system (Advanced Technological Laboratory, Ultramark 8) using a pulsed-wave probe operating at 3 MHz served to record Doppler blood flow signals. The apical echocardiographic view was used to obtain PW signals recorded at the midpoint of the left ventricular outflow tract, approximately 1 cm below the aortic valve. During clinical Doppler examinations, care was taken to reduce the angle between the aortic root and the Doppler ultrasonic beam. Moreover, the "wall filter" was set at 200 or 400 Hz and the length of the sample volume was chosen at 1.5 mm for all analyses, as specified by the manufacturer. The high-pulse repetition frequency (HPRF) mode was selected in two patients because frequency aliasing occurred. The HPRF mode was not used if additional sample volumes were in a blood flow region.

Usually, frequency aliasing that appeared as a saturation of the maximal frequency envelope at PRF/2 value occurred in diastole and on some specific spectra of the spectrogram. Aliasing in diastole was attributed to aortic regurgitation. We also observed aliasing at the end of the systolic period when the aortic valve closed. Visual inspection of the spectrograms (Cloutier et al. 1990a) showed that only few patients had aliasing at peak systole. In these patients, aliasing was seen at the tail of a few spectra and only on a few frequency samples. We are aware that these inherent artifacts slightly modify the amplitude of the spectrograms but they should certainly not modify results on the amplitude distributions. Figure 2 is an example of the envelope of a Doppler spectrogram presenting frequency aliasing at peak systole.

Spectral analysis

During playback, the ECG and PW Doppler signals were digitized for 15 s with 12-bit resolution at

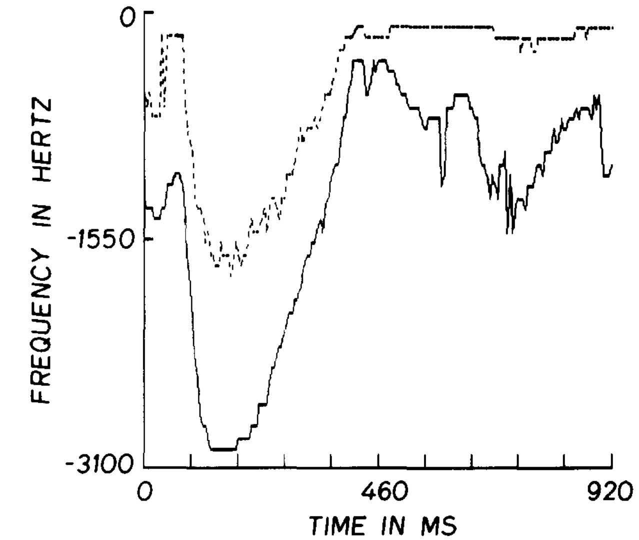

Fig. 2. Example of the envelope of a Doppler spectrogram presenting frequency aliasing at peak systole (note the clipping of the envelope). The dotted line represents the minimal while the full line corresponds to the maximal frequency contours of the envelope obtained before performing filtering.

sampling rates of 0.2 and 20 KHz, respectively. Prior to digitization, the Doppler signals were low-pass filtered at 9 KHz with two eight-order filters (−48 dB/octave). Before computing Doppler spectrograms, a QRS detection algorithm based on a correlation technique was used to synchronize the analysis of the Doppler signals to the cardiac cycle. The mean RR interval of each patient was then computed over the digitized period of 15 s. When analyzing Doppler signals of each cardiac cycle of a patient, only those having a duration within ±10% of the mean RR interval were kept for spectral analysis. This criterion was used in only a few patients because the variability in heart rate was generally within 10%.

A Hanning window of 10 ms was applied to the Doppler signals and a complex fast Fourier transformation (FFT) was used to compute a 256-sample power spectrum. The Hanning window was then slid over the entire cardiac cycle and an FFT computed at each increment of 5 ms to produce a power spectrogram of the Doppler signals. An ensemble average of the power spectrograms computed on 5 cardiac cycles was done to reduce variability of the spectral representation. As seen in Cloutier et al. (1990b), averaging over 5 cycles considerably reduced the amplitude variability in systole. The temporal reference of the QRS complex of the ECG served to align the spectrograms. The averaging process was limited to the interval corresponding to the duration of the shortest heart cycle of the patient. Figure 3 presents an example of a PW-averaged Doppler spectrogram.

Spectral envelope determination

In the present study, the evaluation of amplitude distributions of Doppler spectrograms has been limited to blood flow signals. For this reason, the use of a

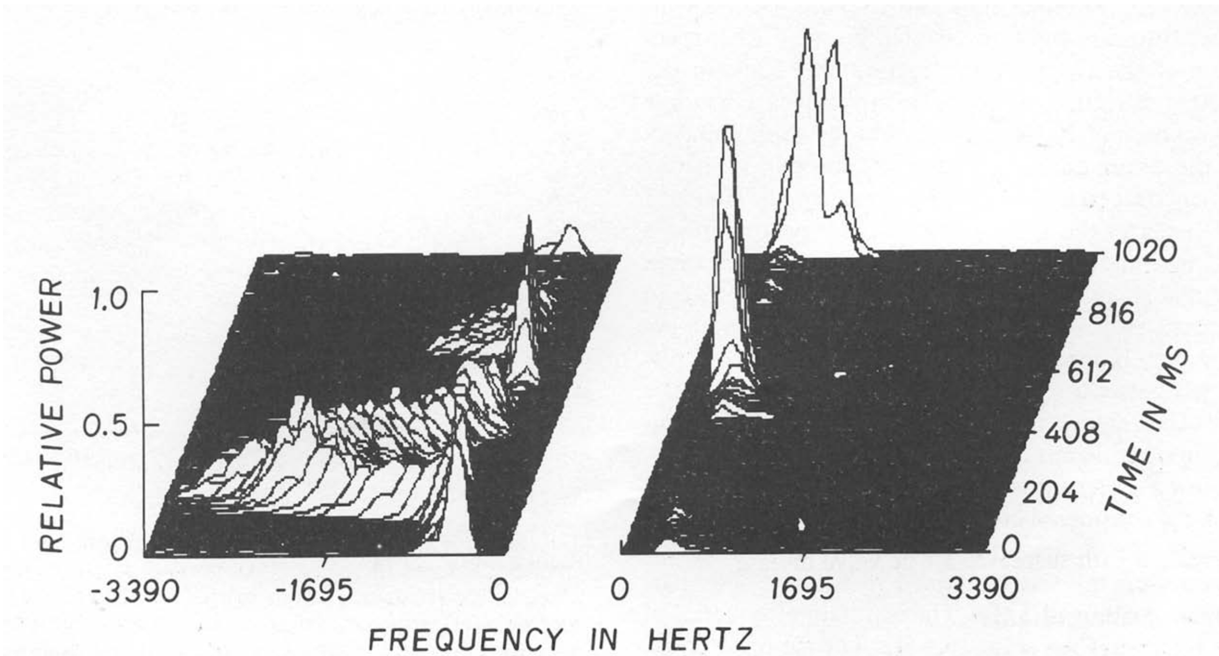

Fig. 3. Example of a pulsed-wave averaged Doppler spectrogram recorded in the left ventricular outflow tract, approximately 1 cm below the aortic valve.

reliable algorithm capable of separating blood flow signal from the background noise of the Doppler spectrogram was essential. In 1988, Mo et al. evaluated different algorithms to estimate the maximal frequency contour of a simulated carotid Doppler spectrogram. Recently, the modified threshold crossing method and the hybrid method proposed by Mo et al. (1988) were improved to evaluate both minimal and maximal frequency contours (Cloutier et al. 1990a). These algorithms were also modified to adapt to the level of the background noise of the spectrogram. Basically, the amplitude of the background noise was computed from samples localized at ±PRF/2 and ±(PRF/2 − 1 sample). In the present study, the modified threshold crossing method was used. Threshold levels equivalent to 8, 9.5 and 11 times the estimated mean amplitude of the background noise were chosen. These thresholds were selected because they correspond to the range used by Mo et al. (1988). A complete description of the modified threshold crossing algorithm can be found in Cloutier et al. (1990a).

Third order low-pass filtering of minimal and maximal frequency contours was performed to reduce variability of the envelope. A recursive finite impulse response (FIR) filter with constant coefficients and linear-phase characteristics was used to perform this task (Lynn 1971, 1972). The selection of the extinction frequency of the filter was based on the following procedure. Two waveforms having a profile that can be associated with systolic contours of Doppler spectrograms were simulated (Cloutier et al. 1990a). These are the Blackman and the sine–cosine windows. Because the time overlap between each spectrum of the Doppler spectrogram was 5 ms in the present study, the number of samples of each window was chosen at 70, which corresponds approximately to the mean duration of the systolic period of our patient data base (350 ms). For instance, the standard reference for the mean left ventricular ejection time is 294 ms in normal man and 306 ms in normal woman for a heart rate of 70 beats/min (Harrison 1974). In patients with a stenotic aortic valve, the systolic period may become prolonged (Harrison 1974). A discrete Fourier transformation was then applied to the simulated signals and the spectral harmonic containing approximately 90% of the energy of each spectrum was determined. The spectral harmonic at 0 Hz was not considered in the computation of the spectral energy. For both windows, more than 90% of the spectral energy was contained in the first four harmonics (within 14.3 Hz). The extinction frequency of the filter was then set at 15.4 Hz. The recursive equation of the digital filter is described by:

$$y(n) = y(n-3) - 3y(n-2) + 3y(n-1) - x(n-3k-3) + 3x(n-k-2) - 3x(n+k-1) + x(n+3k), \quad (3)$$

where $y(n)$ is the output signal of the filter and $x(n)$ the input signal. In this equation, the extinction frequency Fe = the sampling frequency divided by $2k + 1$, where $k = 6$ and the sampling frequency = 200 Hz (1/5 ms).

Evaluation of amplitude distributions

The evaluation of amplitude distributions of the mean power Doppler spectrogram of each patient was limited to the frequency samples included in the spectral envelope. Samples located in the envelope but below the wall filter frequency were ignored because these samples were attenuated by the high-pass function of the filter. The amplitude distributions were evaluated in windows of 30, 60 and 120 ms centered on the position of the maximal detectable frequency of the systolic Doppler spectrogram. Three windows were tested because we did not know how this parameter could affect possible changes in amplitude distributions. These amplitude distributions were also evaluated by subdividing the Doppler spectrogram in three equal portions as shown in Fig. 4. The three frequency bands, referred to in the present paper as the normalized low, middle and high frequency bands, were subdivided by taking into account the value of the maximal frequency of each spectrogram. Different frequency bands were thus selected for dif-

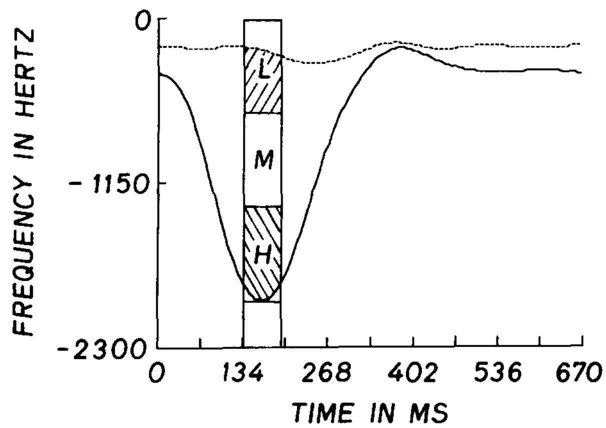

Fig. 4. Minimal (dotted line) and maximal (full line) frequency contours of the negative portion of an averaged Doppler spectrogram. The normalized low (L), middle (M) and high (H) frequency bands used to compute the amplitude distributions are presented for a temporal window of 60 ms.

ferent patients. The reason for using normalized frequency bands was to obtain amplitude distributions independent of the value of the maximal blood flow velocity. The amplitude profile of the mean spectrogram of all patients can then be compared even if the spread of the frequency distributions differed from patient-to-patient.

The amplitude distributions (ADS) were computed from:

$$\mathrm{ADS} = 100 \times \frac{1/N1 \sum_{i=T1}^{T2} \sum_{k=\mathrm{Fa}}^{\mathrm{Fb}} P_i(k)}{1/N\mathrm{tot} \sum_{i=T1}^{T2} \sum_{k=F\mathrm{wall}}^{F\mathrm{max}} P_i(k)}. \quad (4)$$

In this equation, parameters $N1$, Ntot, Fwall and Fmax correspond respectively to the number of frequency samples included in the frequency band of the selected portion of the systolic period, the total number of frequency samples included in the same portion of the systolic period, the frequency of the wall filter and the maximal detectable frequency of the systolic Doppler spectrogram. The frequency samples of the power Doppler spectrogram included into the frequency envelope are represented by $P_i(k)$. The amplitude of the three frequency bands was evaluated by giving consecutively the values of Fwall and Fmax/3, Fmax/3 and 2Fmax/3, and 2Fmax/3 and Fmax to parameters Fa and Fb, respectively. Considering Tmax being the temporal position of the maximal frequency of the systolic Doppler spectrogram of each patient, the three systolic windows tested were obtained by consecutively giving values of Tmax − 15 and Tmax + 15 to parameters $T1$ and $T2$ for the window of 30 ms, values of Tmax − 30 and Tmax + 30 for the window of 60 ms, and finally values of Tmax − 60 and Tmax + 60 for the window of 120 ms.

RESULTS

Among the 48 patients submitted to hemodynamic investigations, 22 patients had no aortic pressure gradient and 26 had a maximal transvalvular pressure gradient greater than 5 mmHg. The average maximal and mean pressure gradients were 85 ± 36 mmHg (mean ± standard deviation, range of 7–150 mmHg) and 59 ± 28 mmHg (range of 4–120 mmHg), respectively. In the group of patients with no aortic pressure gradient, 4 had minimal-to-mild and 3 had moderate-to-severe aortic regurgitation and 13 had coronary artery disease. The mean age of patients included in this group was 56 ± 12 years (range of 32–77 years). In the group of patients with aortic pressure gradient, 15 had minimal-to-mild and 10 had moderate-to-severe aortic insufficiency. In addition, 10 patients had coronary artery disease. The mean age of patients of that group was 65 ± 11 years (range of 35–81 years). No statistically significant difference was found between the cardiac output measured in both groups.

Amplitude distributions of Doppler spectrograms

Only the negative frequencies (blood flow moving toward the aorta) of the pulsed-wave Doppler spectrograms recorded from the apical window were used to compute amplitude distributions. As seen in Fig. 5, the maximal frequency of the minimal frequency contour of Doppler spectrograms was sometimes close to the maximal frequency of the envelope. If no frequency sample was found in a given frequency band; the value of 0% was assigned. In addition, if all the frequency samples were located in a single frequency band, the amplitude distribution in that band was set to 100%.

The absolute mean values of the maximal detectable frequencies, obtained by using a threshold level of 9.5 times the estimated amplitudes of the background noise, were computed. Values were 2148 ± 781 Hz for patients having no aortic pressure gradient and 2486 ± 1049 Hz for patients with a stenotic aortic valve. Considering ideally an angle of zero degrees between the ultrasonic beam and the direction of the blood flow, these frequencies correspond, respectively, to a velocity of 0.6 ± 0.2 m/s and 0.7 ± 0.3

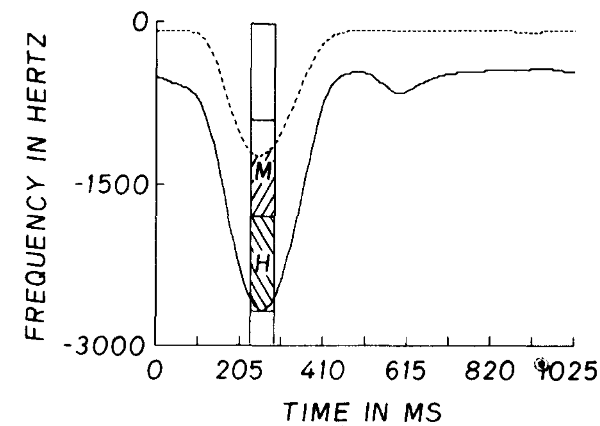

Fig. 5. Minimal (dotted line) and maximal (full line) frequency contours of an averaged Doppler spectrogram. The normalized middle (M) and high (H) frequency bands used to compute the amplitude distributions are presented for a temporal window of 60 ms. Note the absence of Doppler blood flow in the low frequency band.

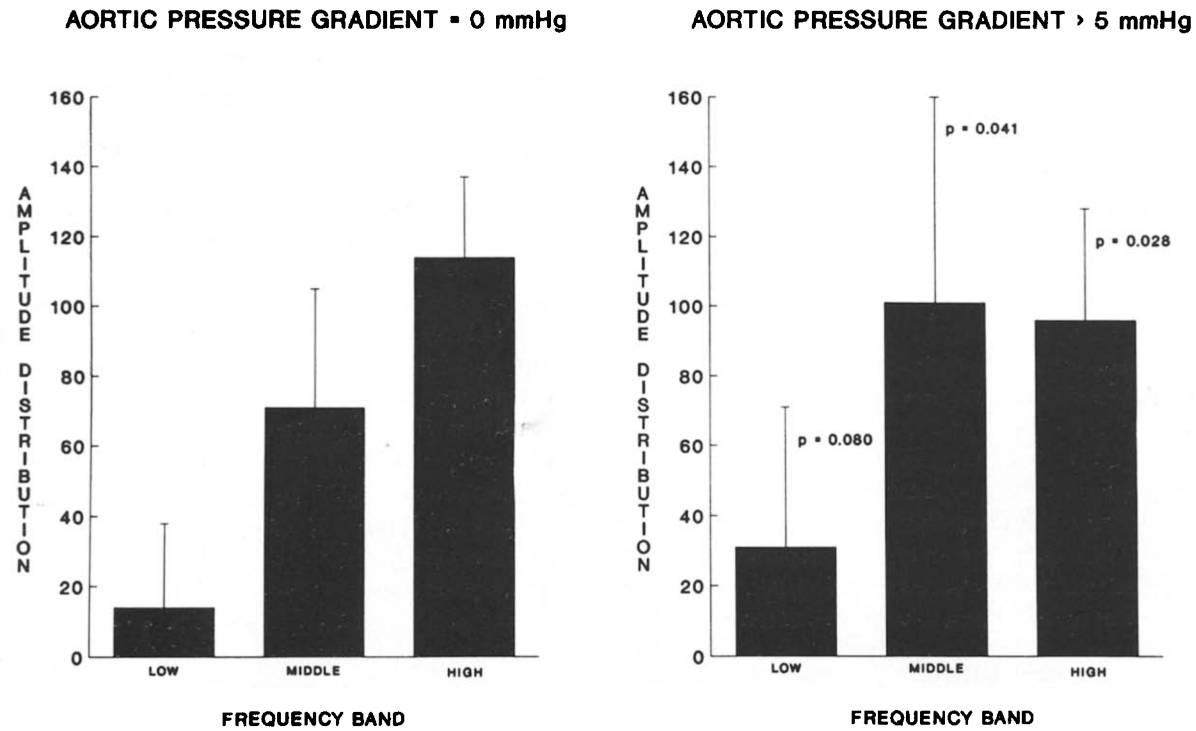

Fig. 6. Amplitude distributions (in %) of pulsed-wave Doppler spectrograms recorded in the left ventricular outflow tract, approximately 1 cm below the aortic valve. Left panel represents the amplitude distribution obtained from a group of patients with no aortic pressure gradient while right panel corresponds to the amplitude distribution of a group of patients with a valvular aortic stenosis.

m/s. According to the averaged maximal detectable frequencies obtained, the mean bandwidths of the normalized low, middle and high frequency bands were similar in both groups of patients. They were 716 Hz in patients with no pressure gradient and 828.7 Hz in patients with aortic valve stenosis. In patients with no pressure gradient, low, middle and high frequency bands corresponded to frequencies ranging between 0–716 Hz, 717–1432 Hz and 1433–2148 Hz, respectively. In patients with aortic valve stenosis, frequency ranges corresponding to these bands were 0–829 Hz, 830–1657 Hz and 1658–2486 Hz, respectively.

A three-way analysis of variance (ANOVA) with repeated measures on two factors (threshold levels and temporal windows) was used to study the value of amplitude distributions for discriminating between the two groups of patients. This statistical analysis also served to study the influence of the threshold level and temporal window on the amplitude distributions of Doppler spectrograms. The discriminating properties of the normalized low, middle and high frequency bands of the amplitude distributions appear to be unrelated to the choice of the threshold level and temporal window. Results of the statistical analysis also showed a significant difference between mean values of the middle (p = 0.041) and high (p = 0.028) frequency bands of the Doppler spectrograms measured on both groups of patients. Because the discriminating properties of amplitude distributions were unrelated to the selection of the threshold level and temporal window duration, we have chosen arbitrarily to describe the results for a threshold level of 9.5 times the estimated values of the background noise, and for a temporal window of 60 ms. Results are shown in Fig. 6.

As seen in this figure, the amplitude distribution increases progressively between the normalized low, middle and high frequency bands in the group of patients with no aortic pressure gradient. The amplitude distribution also increases between the normalized low and middle frequency bands of patients with a stenotic aortic valve. However, the amplitude distribution decreases slightly between the middle and high frequency bands in this group of patients. The profile observed in these two bands is also relatively flat with mean amplitudes close to the normalized value of 100%, which means that the mean amplitude in the two frequency bands was approximately the same as the mean amplitude of the Doppler spectrogram computed within the window of 60 ms and within the spectral envelope. As seen in Fig. 6, the mean values of the amplitude distribution in each frequency band were, respectively, 14 ± 24%, 71 ± 34% and 114 ± 23% for patients with no pressure gradient. Mean values of 31 ± 40%, 101 ± 60% and 96 ± 32% were obtained for patients with a stenotic valve.

DISCUSSION

As discussed previously, the variation in the amplitude of the cardiac backscattered ultrasonic signals can be attributed to factors summarized in Fig. 1 and possibly to other phenomena not already elucidated. Among factors of Fig. 1, the influence of the distance, the protein content of the tissues and fluids and the angle between the ultrasonic beam and the orientation of cardiac muscular fibers do not explain variations observed in the amplitude distributions of Doppler spectrograms. For instance, because these factors attenuate all frequencies in a similar way for a given patient, using ratios of mean amplitudes in the computation of amplitude distributions nullifies these possible influences. The cycle-dependent attenuation attributed to variation of cardiac muscular contractility does not explain the results either because the temporal windows used to compute the amplitude distributions were centered on the systolic interval and were similar for both groups of patients. Concerning the aggregation of red blood cells, this phenomenon does not explain the differences observed in Fig. 6 because shear stresses in the left ventricle certainly thwarts blood cell aggregation. Finally, it is unlikely that a variation of hematocrit may be observed between both groups of patients. This factor was also rejected as an explanation of the results. From the various factors seen in Fig. 1, only the effects of the frequency-dependent attenuation (frequency of the ultrasonic beam in Fig. 1) and of blood flow turbulence could then be used to explain the differences observed in Fig. 6.

Effect of the frequency-dependent attenuation on the amplitude distributions

In theoretical studies, Newhouse et al. (1977) and Embree and O'Brien (1990) showed that the frequency-dependent attenuation of ultrasound and Rayleigh scattering from blood distorted the Doppler spectrum. Because attenuation in soft tissues increases almost linearly with frequency, higher Doppler frequencies are more attenuated than lower frequencies or negative frequencies of the spectrogram. On the other hand, Rayleigh scattering from blood implying that the backscattered power is proportional to f^4 tends to increase the relative contribution of the higher frequencies of the spectrogram.

The importance of spectral distortion associated with these phenomena is related to different parameters such as the ratio of the transmitted signal bandwidth to center frequency of the Doppler system (Newhouse et al. 1977; Embree and O'Brien 1990), the shape of the transmitted ultrasonic signal (Newhouse et al. 1977) and the characteristics of the transducer (Holland et al. 1984). For a ratio of transmitted signal bandwidth to center frequency much less than unity, spectral distortion is minimal as predicted by theoretical models (Newhouse et al. 1977; Embree and O'Brien 1990). However, for a ratio greater than one, the distortion may be significant and depends on the relative contribution of attenuation and Rayleigh scattering. Considering only the frequency-dependent attenuation phenomenon, an error of approximately 15% in the estimated mean value of the Doppler frequency shift was measured experimentally by Holland et al. (1984) for a tissue phantom thickness of 7 cm and a ratio of transmitted signal bandwidth to center frequency of 0.2. This significant error caused by spectral distortion may be surprising considering the small ratio of 0.2. However, although Holland et al. (1984) explained partially their results by the fact that the mimicking tissue used had a higher attenuation coefficient than soft biological tissue, the spectral distortion caused by the frequency-dependent attenuation should not be neglected, especially for important tissue thickness. Other experimental validations of the theoretical results of Newhouse et al. (1977) and Embree and O'Brien (1990) would certainly be helpful to confirm all possible effects on Doppler backscattering power.

In the present study, attenuation certainly dominated over Rayleigh scattering because all Doppler measurements were performed from the apical window at a distance approximately ~5–15 cm from the left ventricular outflow tract. Since maximal detectable frequencies and consequently frequency shifts of PW recordings were shown to be similar for both groups of patients, the difference observed in normalized middle and high frequency bands cannot be explained by the frequency-dependent attenuation of ultrasound.

Effect of blood flow turbulence on amplitude distributions

The influence of blood flow turbulence on the amplitude distributions should not be taken into account in the present study because the possibility of having blood flow turbulence in the left ventricular outflow tract is improbable. For instance, the low velocity observed below the aortic valve (<1 m/s) should not favor the development of blood flow turbulence (Reynolds number below its critical transition value). Based on this observation, another factor presented next is suggested to explain the results of Fig. 6.

Effect of blood flow eddies and flow orientation on the amplitude distributions

Based on the results reported in Fig. 6, it appears that the proportion of blood corpuscles that move at high velocity is higher than those moving at middle and low velocities in patients with no aortic pressure gradient. In patients with a stenotic valve, we postulate that irregular eddying motion is generated at peak systole when red blood cells approach the obstruction (see Fig. 7). At peak systole, a part of the laminar flow cannot enter the small orifice and is probably diverged along the outflow tract thus creating local eddies. In addition, modifications in the morphology of the aortic valve leaflets and left ventricular outflow tract in patients with a stenotic valve may provide local cavities or geometries leading to the production of additional eddies. Consequently, the distribution of the velocity within the sample volume is flattened when eddies are added to the laminar blood flow. This is observed in the right panel of Fig. 6 where the amplitude in the low and middle frequency bands is increased and that in the high frequency band is reduced.

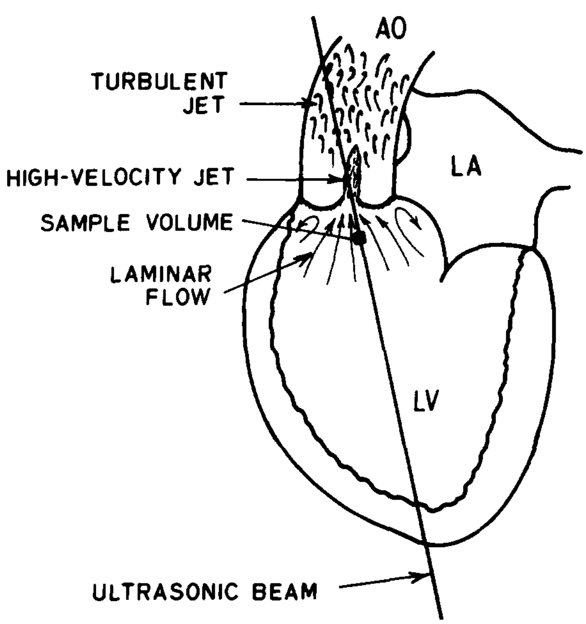

Fig. 7. Illustration of blood flow dynamics near the aortic valve of a patient with a valvular aortic stenosis. In this figure, AO represents the aorta, LA the left atrium, and LV the left ventricle. Note the location of the PW ultrasonic sample volume (black rectangle) approximately 1 cm below the aortic valve.

Another factor that may reduce the amplitude in the high frequency bands of patients with a stenotic aortic valve is changes in flow orientation below the valve. As seen in Fig. 7, the narrowing of the orifice produces curvatures of the stream line orientation. According to the Doppler equation (see Hatle and Angelsen 1985), increasing the angle between the flow and the ultrasonic beam direction results in a decrease of the Doppler frequency shift. An important point that must be considered to appreciate the importance of this observation on changes in amplitude distributions is the width of the Doppler ultrasonic beam. As seen in Fig. 7, the curvature of the flow orientation is minimal at the center of the orifice and increases when approaching the wall of the left ventricular outflow tract. A large beam width would then increase changes in amplitude distributions due to this phenomenon.

Maximal blood flow velocities in the left ventricular outflow tract

In the present study, mean values of maximal detectable frequencies for patients with no aortic pressure gradient and those with aortic valve stenosis were 2148 Hz (0.6 m/s) and 2486 Hz (0.7 m/s), respectively. In 117 healthy subjects aged between 1–65 years, maximal velocity recorded 1 cm below the aortic valve ranged between 0.5–1 m/s (Van Dam et al. 1987). Our results on blood flow velocity in patients with no aortic pressure gradient are thus in accordance with those of Van Dam et al. (1987). In patients with aortic valve stenosis and/or insufficiency, no study reports results on the maximal detectable velocities below the valve. However, in a study of Penn and Dumesnil (1988), 20 patients with aortic valve stenosis (of which 19 had concomitant aortic regurgitation) were analyzed to estimate maximal aortic pressure gradients from PW Doppler tracing recorded in the left ventricular outflow tract. In their study, they showed that the envelope of maximal detectable frequencies in systole had the appearance of a truncated rather than a triangular shaped peak as observed in normal subjects. As reported, the acceleration of blood at peak systole seems significantly reduced when the flow approaches the obstruction. This observation certainly explains why we have obtained a mean velocity of 0.7 m/s in patients with aortic valve stenosis. Moreover, this observation also supports our hypothesis concerning the production of blood flow eddies below the valve. Indeed, when the laminar blood flow approaches the obstruction at peak systole, the addition of eddies certainly decelerates blood flow velocity.

Variability of amplitude distributions

The standard deviations observed in Fig. 6 suggest that the variability of amplitude distributions is relatively important from patient-to-patient. Although statistically significant differences were obtained in normalized middle and high frequency bands, the overlap between both groups of patients is considerable.

As suggested by Mo and Cobbold (1986), the Doppler signal can be considered as a Gaussian random process since the amplitude of the backscattered ultrasonic signal is given by the sum of a large number of randomly distributed red blood cells in space. The stochastic properties of Doppler spectrograms were also considered in one-dimensional computer models of the scattering of ultrasound by blood (Routh et al. 1987; Gough et al. 1988; Routh et al. 1989). Routh et al. (1987) showed that the backscattered reflection coefficient evaluated from a computer simulation of ultrasound propagating into different organized chains of finite-sized slabs is inherently noisy. The standard deviation of the amplitude of the reflection coefficient was found to be close to 100%. This high variability in the amplitude of the simulated backscattered signals was confirmed in an improved version of the model (Routh et al. 1989). Consequently, the stochastic properties of the Doppler spectrograms should contribute to the variability of amplitude distributions shown in Fig. 6. Averaging Doppler spectrograms on more than 5 cardiac cycles would probably decrease this variability.

As seen in Fig. 6, the amplitude variability is higher for amplitude distributions evaluated in patients with a stenotic aortic valve compared to that measured in patients with no pressure gradient. This could be explained by the fact that maximal aortic pressure gradients ranged between 7–150 mmHg. The nonhomogeneity of the severity of aortic valve stenosis has also certainly increased the variability of amplitude distributions.

SUMMARY AND CONCLUSION

Amplitude distributions of PW Doppler spectrograms recorded in the left ventricular outflow tract were reported in the present study. Statistically significant differences were found between the normalized middle and high frequency bands of Doppler signals recorded in a group of patients with no aortic pressure gradient and another group of patients with a stenotic aortic valve. These differences were believed to be attributed to blood flow eddies generated in the left ventricular outflow tract.

Little is known concerning the diagnostic use of features extracted from the amplitude of Doppler spectrograms. Although significant overlaps were found in the middle and high frequency bands of PW signals recorded in both groups of patients, these results represent a first step toward the extraction of new diagnostic information from Doppler spectrograms. It is clear from our results that the proposed technique should be improved to reduce the variance of amplitude distributions. We believe that an improved technique could eventually be interesting to apply in patients where CW examination (Bernoulli equation) is difficult or impossible. The results of the present study may also motivate other studies in which the detection of blood flow eddies or disturbance may be relevant. The diagnosis of plaque deposit at carotid bifurcation and valvular insufficiencies are two examples of possible applications of the present study.

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