Correspondence

A Nonuniform Sampled Coherent Pulsed Doppler Ultrasonic Velocimeter with Increased Velocity Range

Gessé Eduardo Calvo Nogueira, Ademar Ferreira, *Member, IEEE*, and José Tort Vidal

***Abstract*—Coherent pulsed Doppler ultrasonic velocimeters cannot measure large blood velocities in deep vessels. To overcome this limitation, a nonuniform sampling method is proposed. The method is based on adding a delayed sampling sequence interlaced to the conventional one. The time interval between two consecutive samples can be continuously adjusted to avoid undesirable sample volumes. Experimental results are shown, confirming theoretical expectations that the maximum measurable velocity and the maximum measurable velocity width are doubled.**

I. Introduction

Coherent pulsed Doppler ultrasonic velocimeters (PDUV) have been used as an auxiliary noninvasive diagnostic device for vessel diseases. The system measures the blood velocity distribution of a selected range along the beam axis. The blood velocity information is derived from the frequency of the Doppler-shifted signal of the echoes. The spatial discrimination is accomplished by periodically sampling the blood velocity at a pulse repetition frequency PRF. As a consequence, the maximum accessible distance of the PDUV is $d_{\max} = c/2\mathrm{PRF}$, where c is the ultrasound propagation velocity. Because the Doppler-shifted signal is a periodically sampled signal, for a desired distance, sensitivity, and resolution, it is well-known that the maximum measurable frequency of the Doppler-shifted signal is $f_{d\max} = \mathrm{PRF}/2$ (see e.g., [1]). Thus, the measurable frequency range is $(0, \mathrm{PRF}/2)$ and the corresponding measurable velocity range is $(0, V_{\max})$. For PDUVs using quadrature detection, the Doppler-shifted signal spectrum corresponding to the velocity distribution in the sample volume of the PDUV must be limited in the frequency range $(-\mathrm{PRF}/2, \mathrm{PRF}/2)$, where positive frequencies represent direct flow and negative frequencies represent reverse flow. In this case, the overall bandwidth of the Doppler signal spectrum must be less than PRF. This measurable frequency (or velocity) range restricts the use of the PDUV for blood flow parameters estimations at relatively superficial arteries, and there is a great interest in techniques to overcome this limitation [1], [2].

For a selected distance $d < d_{\max}$, the echoes from moving structures located at the distances $(d_{\max} + d), (2d_{\max} + d)\ldots$ are superimposed to those from the desired distance. This physical phenomenon, known as range ambiguity [3], can be used to increase the maximum measurable velocity. By doubling the pulse repetition frequency (DPRF technique), the maximum measurable velocity is doubled, but the maximum measurable distance is reduced to $d'_{\max} = d_{\max}/2$. It is possible to locate the sample volume at a distance $d_2 > d'_{\max}$, by adjusting the PDUV to measure the virtual distance $d_1 = d_2 - d'_{\max}$, if there are no moving structures at d_1, which is inside the range $(0, d'_{\max})$. This procedure is used in practice to increase the maximum measurable velocity of the conventional PDUV [2], [3]. However, with this procedure, the PDUV operator has no flexibility to avoid undesired sample volumes at d_1, except by changing the pulse repetition frequency or repositioning the transducer, when this is feasible. Furthermore, for each new pulse repetition frequency selected PRF$'$, it is necessary to readjust d_1, because $d_1 = d_2 - c/2\mathrm{PRF}'$.

In directional PDUVs, based on additional physiologic information and if the Doppler signal bandwidth is strictly limited to less than PRF, frequency tracking techniques allow the Doppler signal spectrum to occupy a movable frequency range $(f_x, f_x + \mathrm{PRF})$, where f_x can be any frequency, positive or negative (see e.g., [2]). Many other techniques using coherent ultrasonic emission, as well as using pseudo-random and random emission, were suggested to increase the maximum measurable velocity, each one presenting advantages and drawbacks [2]. From these techniques, the only one commercially available seems to be a technique using two ultrasonic frequencies, which was tested in a PDUV by Nitzpon *et al.* [1]. Despite the fact that $V_{\max}$ can be considerably increased by using this technique, the achieved measurable Doppler signal bandwidth, however, is less than PRF, i.e., equal to the measurable bandwidth obtained with frequency tracking. As mentioned by Nitzpon *et al.* [1], this bandwidth can be insufficient to analyze broadband turbulent flow.

In this correspondence a new alternative method, similar to the DPRF technique, is presented in Section II, intended to increase the measurable frequency range to $(-\mathrm{PRF}, \mathrm{PRF})$, and the overall bandwidth to 2PRF. The implementation of the proposed method in a directional PDUV is discussed in Section III. Results from in vivo experiments are presented in Section IV.

II. Nonuniform Sampling: Basic Principles

The method proposed in this correspondence for extension of the maximum measurable velocity of PDUVs is based on the emission of another sequence of equispaced ultrasonic pulses, interlaced to the conventional emission sequence, as shown in Fig. 1. For the first sequence (the conventional one), pulses occur at the instants n/PRF, where $n = 0, \pm 1, \pm 2, \ldots$. The second sequence (the interlaced sequence) is delayed from the first one by k seconds, and its pulses occur at the instants $n/\mathrm{PRF} + k$. As will be shown next, the interlaced sequence produces additional independent samples, doubling the maximum measurable frequency range. The strategy to avoid undesirable sample volumes, caused by the interlaced samples, is to adjust

Manuscript received January 14, 1997; accepted September 23, 1998. This work was supported by the Programa de Apoio ao Desenvolvimento Científico e Tecnológico and the Conselho Nacional de Desenvolvimento Científico e Tecnológico - PADCT/CNPq.

G. E. C. Nogueira and J. T. Vidal are with the Instituto de Pesquisas Energéticas e Nucleares, São Paulo, Brasil (e-mail: genoguei@net.ipen.br).

A. Ferreira is with the Electronics Engineering Department, Universidade de São Paulo, Brasil.

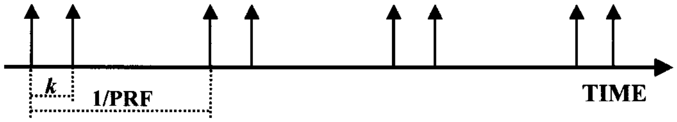

Fig. 1. Two sequences of equispaced ultrasonic pulses interlaced. In each sequence, pulses are 1/PRF seconds spaced apart, and the delay between the sequences is k.

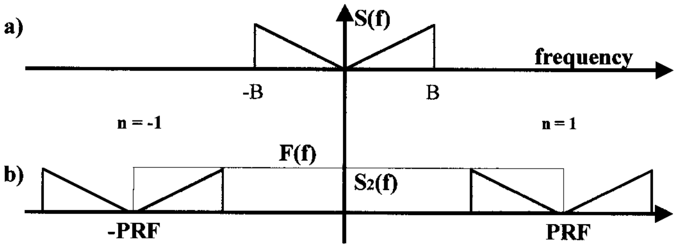

Fig. 2. (a) Spectrum of a low-pass signal; (b) spectrum of the low-pass signal of (a) second-order sampled: the replica at the origin is suppressed. $F(f)$ is an ideal filter, used to select the lower bands of the replicas at $n = \pm 1$.

the delay between sequences, differently from the DPRF technique which the delay is fixed. This topic is further discussed in this section.

A. Interpolation Considerations on Nonuniformly Sampled Doppler Signals

The emission pattern depicted in Fig. 1 produces nonuniformly time-spaced samples of the Doppler signal. This sampling pattern is known as second-order sampling because two uniform streams of samples are interlaced. The recovery of the original signal (to extract all relevant Doppler signal information) can be accomplished by using two interpolators [4]. These interpolators, however, are difficult to implement by analog filters. Instead, they often are approximated by finite impulse response filters, requiring many considerations concerning the filter length, computational efficiency, guard-bands, and reconstruction errors [4]. Furthermore, if it is necessary to adjust the delay of the interlaced sequence, as in our case, then the interpolators must be readjusted for each new delay. Probably for reasons like these, to our knowledge, nonuniform sampling and interpolating were never used for ultrasonic systems.

The PDUV presented in this correspondence uses a new nonuniform sampling method, recently proposed to simplify the recovery of the original signal from its nonuniform samples [5]. This method is discussed here only for the second-order case.

Consider a real signal $s(t)$, band-limited to the frequency range $|f| < B$ Hz, as depicted in Fig. 2(a). Here $s(t)$ represents the Doppler signal after coherent detection, i.e., at a low-pass position. We consider $s(t)$ a real signal instead of a complex one, as it is normally done when quadrature detection is used. This choice is only to simplify the following analysis, which can be easily extended to complex signals.

Second-order sampling of $s(t)$, according to the method suggested in [5], [6], is defined as:

$$s_2(t) = s(t)\left[a_1 \sum_{n=-\infty}^{\infty} \delta(t - n/\mathrm{PRF}) + a_2 \sum_{n=-\infty}^{\infty} \delta(t - n/\mathrm{PRF} - k)\right] \quad (1)$$

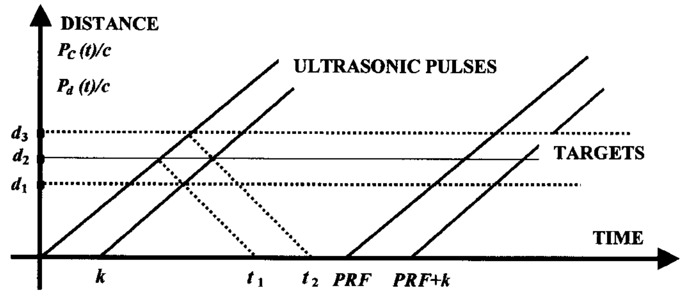

Fig. 3. Trajectory of the front ends of two equispaced sequences of ultrasonic pulses: the sequences are interlaced. Only the pulses at $t = 0, t = 1/\mathrm{PRF}$ and $t = k, t = k + 1/\mathrm{PRF}$, of $p_c(t)$ and $p_d(t)$, respectively representing a conventional and a delayed sequence, is depicted. The trajectories of three isolated (slow moving) targets, at distances d_1, d_2, and d_3 also are depicted. All trajectories are normalized by the sound propagation velocity c. The echoes at t_1 and at t_2 are samples from the target at d_2.

where a_1 and a_2 are scalar and $\delta(t)$ is the Dirac delta function. To simplify the following analysis, we restrict the delay between sequences to the interval $0 < k < 1/\mathrm{PRF}$. The spectrum of the sampled signal (1) is:

$$S_2(f) = \mathrm{PRF} \sum_{n=-\infty}^{\infty} [a_1 + a_2 \exp(-j2\pi nk\mathrm{PRF})] S(f - n\mathrm{PRF}) \quad (2)$$

where $j = \sqrt{-1}$ and $S(f)$ designates the Fourier transform of $s(t)$. In general the replicas of $S(f)$ in (2) are PRF spaced apart, and the signal bandwidth should be $B < \mathrm{PRF}/2$ to avoid spectral overlapping. Notice that, in (2) for $a_1 = 1$ and $a_2 = -1$, the replica at $n = 0$ is suppressed so that there is a gap at the origin of the spectrum (2). This gap can be occupied by the lower bands of the adjacent replicas, located at $n = \pm 1$, without aliasing in the frequency range $[-\mathrm{PRF}, \mathrm{PRF}]$, as depicted in Fig. 2(b). Notice that this gap does not depend on the delay k. Also, that this gap is produced entirely by the sampling operation (the unique restriction is $a_1 = 1$ and $a_2 = -1$), and not as a consequence of the interpolation process, as is normally done by known nonuniform interpolation methods.

Applying an ideal filter to the sampled signal (1) (with $a_1 = 1$ and $a_2 = -1$) with passband $|f| \leq \mathrm{PRF}$, the resulting signal spectrum is:

$$S_2^F(f) = S^-(f - \mathrm{PRF})[1 - \exp(-j2\pi k\mathrm{PRF})] + S^+(f + \mathrm{PRF})[1 - \exp(j2\pi k\mathrm{PRF})] \quad (3)$$

where $S(f) = S^+(f) + S^-(f)$, with superscripts indicating positive and negative parts (or bands) of the original (real) signal spectrum. Thus, the remaining spectrum is composed by phased, shifted, and spectrally inverted versions of bands of the original signal spectrum: it contains all relevant information about $S(f)$. That is, if $\overline{f}$ is the mean frequency of $s(t)$ and $\overline{f_2}$ is the mean frequency of $S_2^F(f)$ of (3), then, it is easy to prove that [6]:

$$\overline{f} = \mathrm{PRF} - \overline{f_2}. \quad (4)$$

Thus, the mean frequency and analogously other signal frequency parameters, such as spectral width, can be directly derived from the second-order sampled and low-pass filtered signal.

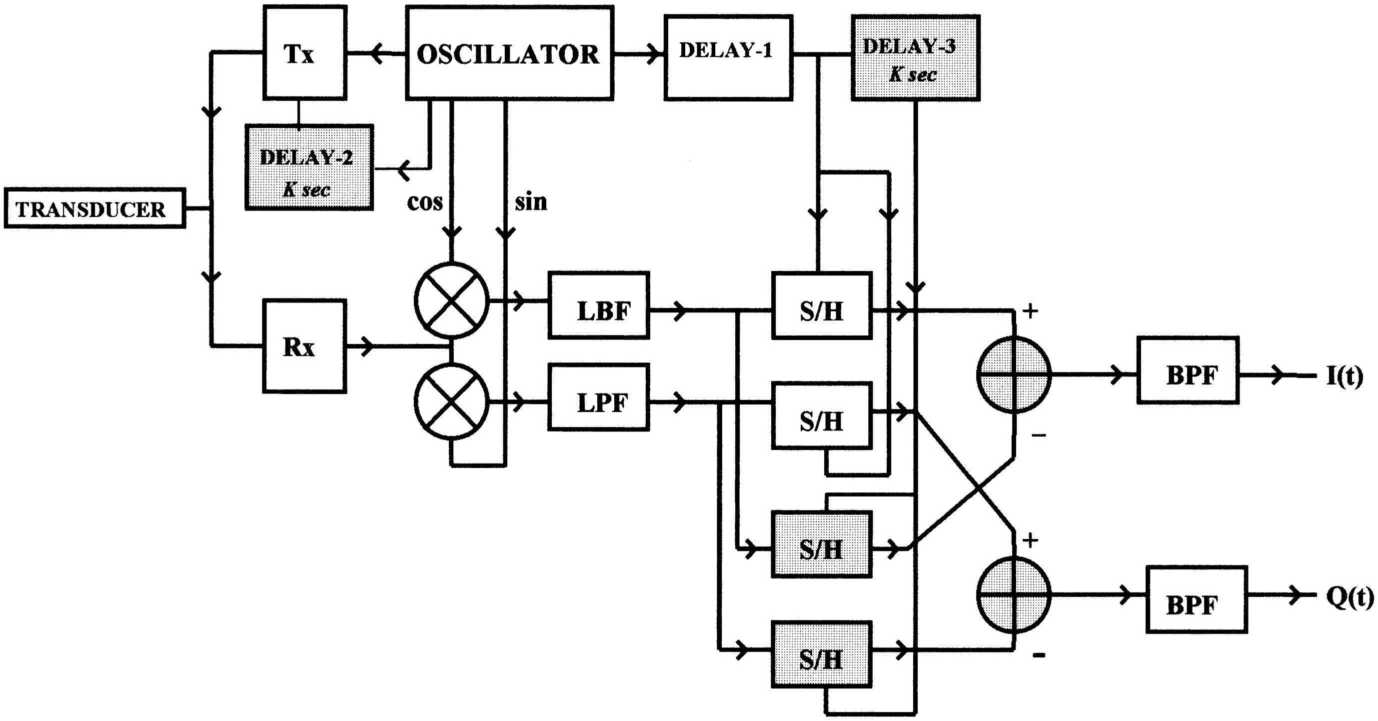

Fig. 4. Block diagram of a nonuniformly sampled directional PDUV. Tx stands for the transmitter amplifier, Rx is the receiver amplifier, LPF is a low-pass filter, BPF is a bandpass filter, and S/H are sample-and-hold devices. $I(t)$ and $Q(t)$ are in-phase and quadrature Doppler signals, respectively.

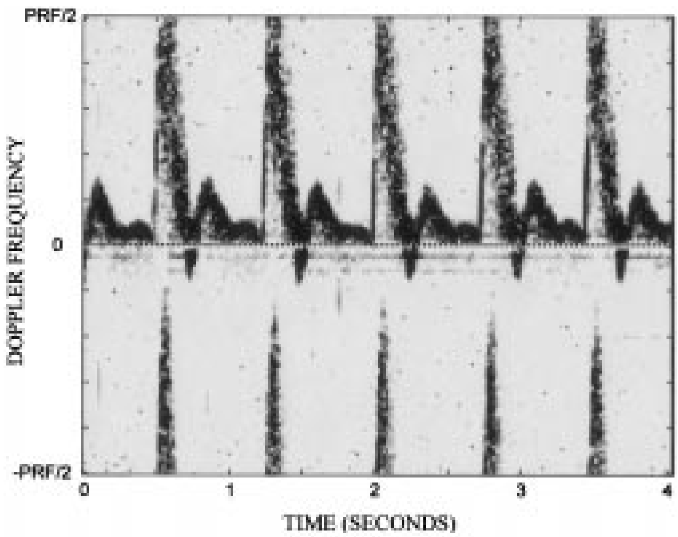

Fig. 5. Spectrogram from a brachial artery of a healthy person, obtained with a conventional PDUV. Positive frequencies represent direct flow and negative frequencies represent reverse flow. The spectral components from direct flow occupy the rage $(-\text{PRF}, 0)$, reserved for reverse flow, clearly indicating subsampled Doppler signals.

The filtered second-order sampled signal is attenuated or magnified by a factor that depends on the delay k. This fact can be verified by analyzing the absolute value of the positive part of the spectrum (3), i.e., $\text{abs}[1 - \exp(-j2\pi k\text{PRF})]$. As a consequence, when the delay is $1/2\text{PRF}$ the amplitude of the analog signal obtained by inverse Fourier transforming (3) is doubled when compared to the original signal. However, if the delay is zero or $1/\text{PRF}$, the amplitude of the resulting signal is zero. But, the limits of the delay interval $(0, 1/\text{PRF})$ are not of interest because the delayed pulse sequence must be positioned so that the corresponding echoes stay out of the nondelayed sample volume. The intrinsic PDUV noise power also is affected by nonuniform sampling, but this analysis is out of the scope of this correspondence. Here it is sufficient to know that it is possible to adjust the delay over a wide range before the signal-to-noise ratio becomes low (for details, see [6]).

B. Physical Considerations on Nonuniform Sampling

In situations that contain both fixed and moving structures in the measurable distance interval $(0, d_{\max})$, as frequently occurs in the human body [3], it is possible to locate an interlaced sampling sequence, by adjusting the delay k, to avoid undesirable sample volumes. A graphical analysis is used here to explain this method. Fig. 3 shows the trajectories of the front ends of the two sequences of ultrasonic pulses. Only two pulses of each sequence, at $t = 0, t = 1/\text{PRF}$ and at $t = k, t = 1/\text{PRF}+k$, of $p_c(t)$ and $p_d(t)$, respectively representing the conventional and the delayed sequence, is depicted. Fig. 3 also shows the trajectories of three isolated (slow moving) targets, at distances $d_1(t)$, $d_2(t)$, and $d_3(t)$. All trajectories are normalized by the sound propagation velocity c. Thus, in Fig. 3, the trajectory of each ultrasonic pulse is plotted at 45 degrees from the time axis. For each emitted pulse, its front end intercepts a target, reverses its propagation direction, and returns to the emitter. The distances of the targets are assumed to have only small variations over various pulses ($V \ll c$), so that their plots are approximately parallel to the time axis, i.e., $d_1(t) \simeq d_1$, $d_2(t) \simeq d_2$ and $d_3(t) \simeq d_3$. Suppose now that we are interested in measuring the velocity of the target located at d_2. The first pulse at $t = 0$ intercepts this target located at the distance d_2, is reflected, and returns to the emitter at time t_1. The second pulse, delayed by k seconds, is transmitted, intercepts the target at d_2, and returns to the emitter at t_2. Thus we have two samples of the trajectory of $d_2(t)$ at t_1 and t_2. Notice, however, that the second pulse also intercepts another target at a distance d_1, as shown in Fig. 3, and this echo also returns to the emitter at t_1. Thus, the distance d_1 is an ambiguous distance for the PDUV, when measuring a moving target at distance d_2. For the target located at the distance d_3, the same considerations apply. Thus, the interlaced sampling introduces additional range ambiguity inside the range $(0, d_{\max})$. These considerations also apply for the DPRF technique. In fact, for $k = 1/2\text{PRF}$, the interlaced sampling is equal to the DPRF method, which range ambiguities also occur inside the range $(0, d_{\max})$. For the DPRF method, however, as mentioned in Section I, to avoid undesirable sample volumes it is necessary to change the pulse repetition frequency and, consequently, to readjust the selected virtual distance. For interlaced sampling, to avoid undesirable targets, one can simply change the delay k. That is, as we showed in Section II,A, the delay k between the sequences can be continuously adjustable in the interval $(0, 1/\text{PRF})$. Thus, interlaced sampling presents the following advantage compared to the DPRF method: the interlaced sequence can be continuously adjusted to avoid undesirable sample volumes, without changing the pulse repetition rate of the PDUV, and consequently, without changing the distance axis. To adjust the delay k, we can inspect the pulse-echo response, similarly to what is normally done when the DPRF method is used [3]. As an alternative procedure to adjust the delay k for systems at which pulse-echo response is not available, one can simply hear the Doppler-audio signals when the interlaced sequence is switched on. If undesirable moving structures are detected when the interlaced sequence is switched on, it is necessary to change only the delay.

III. Implementation

We constructed a directional 5 MHz nonuniformly sampled PDUV to verify the practicability of the proposed nonuniform sampling method. A simplified block diagram of the constructed system is shown in Fig. 4, where gray blocks indicate the necessary modifications, compared to a conventional PDUV. The oscillator provides the fundamental transmission frequency and the pulse repetition frequency PRF. The combination of the pulse repetition frequency with the fundamental emission frequency produces an equispaced burst sequence. A first delay (delay-1 in Fig. 4), synchronized by the burst sequence, produces the sampling delay, proportional to the desired range. A second delay (delay-2 in Fig. 4), synchronized by the burst generator, produces another sequence of equispaced bursts, delayed by k seconds. Adding these two sequences produces the interlaced transmission sequence. The delay k can be adjusted to avoid undesirable sample volumes, as previously explained. Synchronized by the first delay, a third delay (delay-3 in Fig. 4) produces the interlaced sampling delay, proportional to the desired range. The subtraction of the two selected sequences (by the sample-and-hold devices) causes the suppression of the replica of the original signal spectrum at $f = 0$. The $I(t)$ and $Q(t)$ signals in Fig. 4 are “in phase” and “in quadrature” signals, normally used to derive the direct and reverse flow.

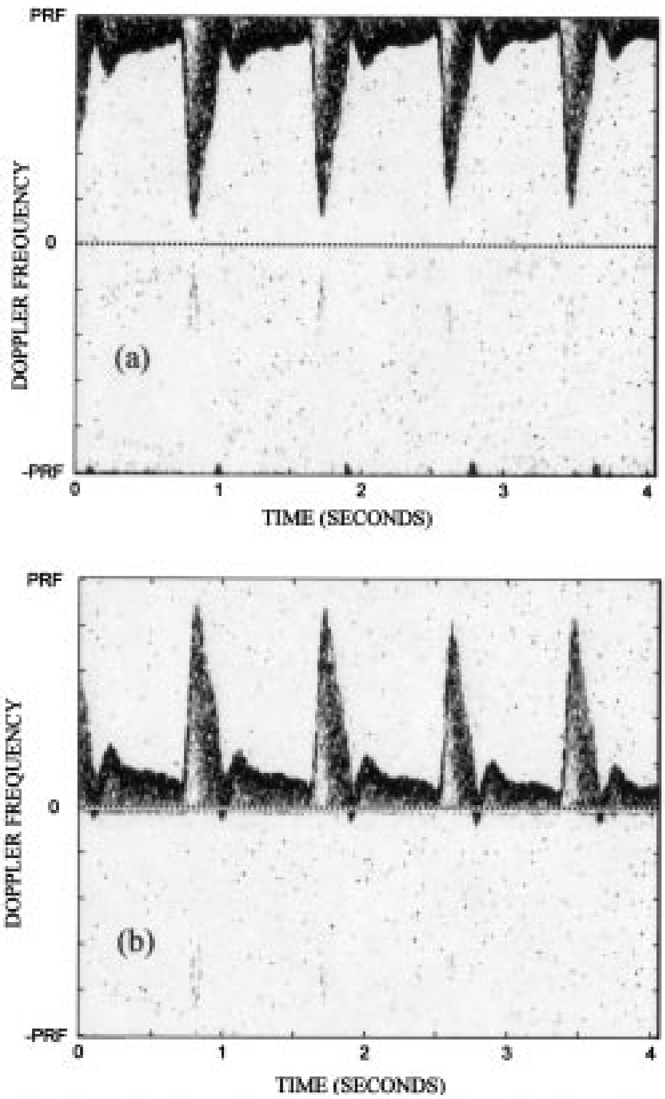

Fig. 6. Spectrogram from a brachial artery of a healthy person, obtained from a nonuniformly sampled PDUV in (a), where spectral components from replicas at the origin of the spectrum are suppressed, increasing the measurable frequency range to $(-\text{PRF}, \text{PRF})$. The spectrogram (a) can be inverted and sifted, as shown in (b).

IV. Experimental Results

We experimentally verified the above considerations on the nonuniformly sampled PDUV. In all experiments we used $\text{PRF} = 3.9$ kHz and the delay was adjusted to $k = 1/3\text{PRF}$. The Doppler signals $I(t)$ and $Q(t)$ (Fig. 4) were used to produce conventional spectrograms. The experimental results are presented by Figs. 5 and 6.

Fig. 5 shows a spectrogram from a brachial artery of a healthy subject, obtained without the interlaced sequence, i.e., the PDUV was operated as a conventional one. The spectral window is fixed to $(-\text{PRF}/2, \text{PRF}/2)$, which is the measurable frequency range for conventional directional PDUVs. As we can verify in Fig. 5, spectral components from direct flow are larger than the frequency range $(0, \text{PRF}/2)$, and their higher components are aliasing the frequency range $(-\text{PRF}/2, 0)$, reserved for reverse flow, clearly indicating that the Doppler signals were under-sampled.

Fig. 6(a) shows a spectrogram obtained in the same conditions as in the previous experiment, except that the interlaced sequence was switched on. In this case the spectral components from the replicas located at the origin are suppressed. The remaining ones are the components from direct flow, located in the frequency axis at $f = \mathrm{PRF}$ (spectrally inverted and shifted) and the spectral components from reverse flow, at $f = -\mathrm{PRF}$ (also spectrally inverted and shifted). The spectral components are shown inverted and shifted, as they are after suppression of the replicas at the origin, as explained in Section II. In practice it is easy to plot this spectrogram with the remaining spectral components at the origin, simply by inverting and shifting the FFT components after computation, as shown in Fig. 6(b). As we can verify in Figs. 6(a) or (b), spectral components of direct and reverse flows can occupy all the frequency range $(0, \mathrm{PRF})$, and $(-\mathrm{PRF}, 0)$, respectively, without aliasing, so that the measurable spectral width is extended to 2PRF.

V. Conclusions

A nonuniform sampling technique, intended to increase the velocity range of PDUVs, was implemented in a 5 MHz directional coherent pulsed Doppler velocimeter. The technique is similar to the known DPRF technique (doubling the pulse repetition rate of conventional PDUVs to increase the measurable velocity). The main advantage of the proposed technique, compared to the DPRF technique, is the flexibility to avoid undesired sample volumes. Experimental results from in vivo experiments, corroborating the theoretical expectation, show that the maximum measurable velocity is doubled. The cost of this is the reduction of the continuous measurable distance interval (which also occurs with the DPRF technique) and a tolerable degradation of the Doppler signal power. These results suggest a simple low cost technique, intended to double the maximum measurable velocity and velocity width of the PDUV.

References

[1] H. J. Nitzpon, J. C. Rajaonah, C. B. Burckhardt, B. Dousse, and J. J. Meister, "A new pulsed Wave Doppler ultrasound system to measure blood velocities beyond the Nyquist limit," *IEEE Trans. Ultrason., Ferroelect., Freq. Contr.*, vol. 42, pp. 265–279, Mar. 1995.

[2] D. H. Evans, "Techniques for color-flow imaging" in *Advances in Ultrasound Techniques and Instrumentation*, vol. 28, Churchill Livingstone, New York, NY, 1993, pp. 87–107.

[3] I. A. Jaward, M. L. Taylor, F. Pruitt-Hudson, and Y. H. Sohn, "Range ambiguity in pulsed Doppler ultrasound: The ambiguity clarified?" *J. Clin. Ultrasound*, vol. 13, pp. 475–479, Sep. 1985.

[4] A. J. Coulson, "A generalization of nonuniform bandpass sampling," *IEEE Trans. Signal Processing*, vol. 43, no. 3, pp. 694–704, 1995.

[5] G. E. C. Nogueira and A. Ferreira, "Higher-order sampling and recovering of low-pass signals," submitted for publication.

[6] G. E. C. Nogueira, "Extension of the measurable velocity range of the pulsed Doppler ultrasonic velocimeter (Extensão da faixa de velocidades mensuráveis do velocímetro Doppler Ultra-sônico pulsátil)," Ph.D. dissertation, Instituto de Pesquisas Energéticas e Nucleares, São Paulo, Brazil, 1995.