ARRAY SIGNAL PROCESSING APPROACHES TO ULTRASOUND-BASED ARTERIAL PULSE WAVE VELOCITY ESTIMATION

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ABSTRACT

Assessment of the arterial pulse wave velocity (PWV) has long been an area of interest in physiology, and ultrasound has long been used to provide measurements for such assessments. Recently, new signal processing approaches for ultrasound data have emerged. However, these methods suffer from inaccuracies due to pulse wave reflections, which are always present and can strongly bias the PWV estimates away from the true velocity. Recently the authors [1] showed that pulse wave velocity estimation from several ultrasound measurements taken along a short uniform arterial segment is equivalent to the broadband directional of arrival problem with coherent multipath found in radar and sonar. This tutorial paper reviews the physiological and ultrasound-systems aspects of the PWV estimation problem, and examines its relationship to the direction-of-arrival estimation problem. The paper also demonstrates why nonlinear, high-resolution methods are needed and outlines the application of several such estimators to the problem.

1. INTRODUCTION

When the left ventricle of the human heart contracts, an impulse of pressure travels through the arterial system. The speed with which this pressure pulse travels is called the arterial pulse wave velocity (PWV). Roughly, it varies in the range of 3 m/s to as high as 15 m/s in the human body [2]. The mechanism of propagation of this wave is the elastic expansion of the walls of the arteries, not motion of the blood, although there is an impulse of blood velocity associated with the propagating pressure. This propagation mechanism can be approximately described by a linear wave equation parameterized by blood density and arterial wall elasticity. (This is an approximation because the arterial walls are neither purely elastic nor linearly elastic.) If we extract the PWV from this wave equation, we get the Moens-Korteweg equation:

$$v_p = \sqrt{\frac{E\,h}{2r\rho}} \qquad (1)$$

where E is the modulus of elasticity, h is the thickness of the artery wall, r is the radius of the arterial lumen and ρ is the blood density. The observable disturbance whose propagation is described by the wave equation is the expansion or *distension* of the artery under the influence of the pressure pulse.

Because pulse propagation is based on elastic expansion of the arterial walls, the PWV is indicative of the local elastic state of an arterial segment, which may be indicative of arterial disease. In addition, the PWV changes with the mean radius of the arteries, and so the PWV can be generally indicative of blood pressure, since increased pressures will result in greater mean arterial distension. Thus the PWV is potentially a physiological measurement of interest.

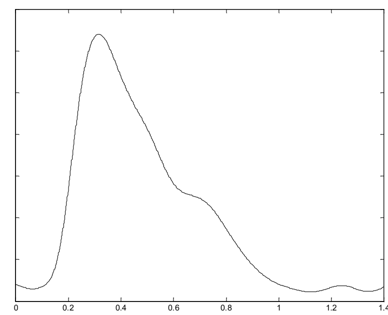

Figure 1. Pressure waveform from plethysmography data.

A typical arterial pressure waveform is plotted in Figure 1. This waveform was obtained from the femoral artery of a human subject using impedance plethysmography, which in this application measures change in total blood volume in a section of the leg. A single heart cycle is depicted, which lasts for roughly 1.4 seconds in this case. Such a waveform will have a fundamental wavelength of several meters for pulse wave velocities in the normal physiological range.

Pulse wave reflections arise at points in the arterial system where the arteries change their size, such as at bifurcations. The change in radius changes the characteristic impedance of the transmission medium, which results in a reflection traveling with the same speed as the forward wave. The arterial system is tree-structured, and so reflection sites are numerous. Therefore any measurement of the arterial pressure pulse, or the arterial distension that results from it, will be affected by reflections, and the effect can be great. The observed combination of forward and reflected waves will produce different wave shapes at distinct measurement sites. This change of shape means that techniques based on identification of a common feature of the disturbance at two sites and measurement the time delay between its appearance at those sites will be biased away from the true value of the delay of the forward-propagating disturbance. Such a biased measurement, formed under the influence of reflections, is called an *apparent* pulse wave velocity [2]. The true value of the PWV, which would be observed in a (hypothetical) long uniform arterial section, is called the characteristic PWV, and it is this quantity that is given by the Moens-Korteweg equation.

Milnor [2] describes several methods of measuring PWV, based on comparison of pressure, flow or distension measurements at two arterial sites. Two such techniques are analyzed in [1]. The standard PWV estimate of this kind is the "foot-to-foot" method, which is based on the time difference between the onset of systolic pressure. It acquires limited resistance to the effect of reflections through selection of this feature, but it is more biased than model-based approaches described here [1].

More recent work in this area has focused on obtaining the PWV in a single, unbifurcated arterial segment of uniform lumen radius. Such segments are limited in length to roughly 5 centimeters. It is to this problem that ultrasound can be applied with greatest effect, since it is able to produce nearly simultaneous, independent distension measurements from arterial sites spaced only millimeters away from each other.

The use of ultrasound measurements at multiple sites along the artery to estimate pulse wave velocity was introduced by Meinders, et. al. [4]. Because of the placement of the measurement sites with respect to the artery, we call this data set the *long-axis* measurement (see Figure 2). The measurement is typically made using a single linear array of ultrasound transducer elements, with the different measurement sites defined by different translations of the active aperture. In the Meinders approach, ultrasound is used to obtain distension waveforms at sixteen sites along a two centimeter long arterial segment, and spatial and temporal derivatives are obtained and divided to produce a spatial velocity estimate for the disturbance:

$$\hat{v}_p = \frac{dx}{dt} = \frac{\partial y / \partial t}{\partial y / \partial x} \qquad (2)$$

where y is the vessel diameter and x is the distance in the direction of wave propagation. This method makes no provision for pulse wave reflections and is highly biased by them [1].

The application of ultrasound to medical imaging is well-documented [3]. There are two basic ways in which the pulse wave disturbance can be sensed using ultrasound: direct distension measurement or distention rate measurement based on Doppler shift induced by the moving arterial walls. The technique of [4] uses arterial diameter as the basic measurement; a useful refinement to the method was introduced in [5] and uses tissue Doppler to overcome axial resolution limitations of the diameter measurement.

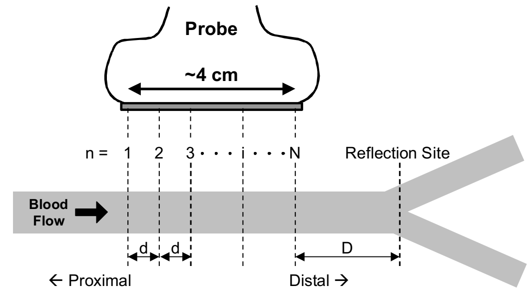

Figure 2. Long-axis ultrasound data collection.

The time required to acquire a single one-dimensional ultrasound image is determined by the round-trip acoustic propagation time. The speed of propagation of ultrasound in tissue varies around 1.54 mm/microsec. If a superficial artery is chosen for imaging, it may be required to image to a depth of 2-3 cm, requiring roughly 40 microseconds per line. If arterial area measurements are being taken, multiple lines will be needed. If Doppler measurements are being used, several insonifications are required to compute a single Doppler shift. In either case, time must be allowed between insonifications to let reverberations fade away. Also, data cannot be obtained simultaneously from more than one site. All of this affects the maximum sample rate at which distension or distension rate data may be acquired. In our experimental work, we use data sampled at approximately one Ksample/sec at six measurement sites.

In Section 2 of this paper, we describe an interpretation of the PWV estimation problem as one of direction-of-arrival estimation, which allows the application of a wide range of array signal processing techniques to the problem. In Section 3 we give an overview of the algorithms available for this problem and a simulation example. In Section 4 we summarize.

2. Long-axis PWV Estimation as a Broadband Array Signal Processing Problem

Figure 2 shows a schematic representation of the set-up for ultrasound data collection. At each of N measurement sites, corresponding to ultrasound A-lines separated by a distance d from one another, a Doppler signal due to arterial wall motion is recorded. From this data, we wish to estimate the pulse wave velocity.

Suppose that the most distal site is a distance D upstream from a reflection site with reflection coefficient Γ (see Figure 2). The total disturbance at the n^{th} site may be considered to be the sum of a forward and a reverse component. (We discount the possible effects of multiple reflections.) If we express all the components of the observed disturbances as delayed versions of the forward disturbance observed at the first site, denoted by z(t), then the disturbance at the N sites is given by a vector

$$\mathbf{y}(t)=\begin{bmatrix} z(t) \\ \vdots \\ z\left(t-\frac{(n-1)d}{v_p}\right) \\ \vdots \\ z\left(t-\frac{(N-1)d}{v_p}\right) \end{bmatrix}+\Gamma\begin{bmatrix} z\left(t-\frac{\Delta}{v_p}\right) \\ \vdots \\ z\left(t-\frac{-(n-1)d}{v_p}-\frac{\Delta}{v_p}\right) \\ \vdots \\ z\left(t-\frac{-(N-1)d}{v_p}-\frac{\Delta}{v_p}\right) \end{bmatrix} \quad (3)$$

where $\Delta = 2(N-1)d + 2D$, is the round-trip distance from the first measurement site to the reflection site and where v_p is the pulse wave velocity. Given a set of observations of the $\mathbf{y}(t)$ vector, we wish to estimate the three unknown parameters, D, Γ and v_p. (The quantity d is known, and D may also be known in some cases.)

It is clear from Figure 1 that z(t) is a broadband waveform, and equation (3) has a form similar to that of a broadband direction-of-arrival (DOA) estimation problem for two far-field sources impinging on a uniformly spaced line array. In the standard broadband direction-of-arrival scenario, the propagation speed is constant and the delay between the components observed at different array elements varies due to the direction of arrival. In the PWV estimation problem, the direction of arrival is fixed at $\alpha=0$, for the forward wave and $\alpha=\pi$, for the reflection, and the phase increment varies due to variation in the speed of propagation. Mathematically, this is a very similar problem, and the same mathematical ideas and algorithms can be used to solve it.

It is common to treat the broadband DOA problem by taking a Fourier transform of the observations and treating each separate frequency component as arising from a narrowband scenario [6], and we will adopt this approach. If we take the DFT of a segment of each of the observations, assuming that we have samples at times $t=0,T,\ldots,mT\ldots,(M-1)T$, we can write a vector of k^{th} DFT coefficients as [1]

$$\mathbf{Y}(k)=Z(k)\left[\boldsymbol{\varepsilon}\left(\frac{2\pi dk}{TMv_p}\right)+\Gamma\, e^{\frac{j2\pi\Delta}{TMv_p}}\,\boldsymbol{\varepsilon}\left(\frac{-2\pi dk}{TMv_p}\right)\right] \quad (4)$$

where we define {Z(k)} to be the DFT of {z(mT)}, and $\boldsymbol{\varepsilon}(\phi)$ is a narrowband phase response vector

$$\boldsymbol{\varepsilon}(\phi)=\begin{bmatrix}1 & \cdots & e^{j\phi(n-1)} & \cdots & e^{j\phi(N-1)}\end{bmatrix}^T \quad (5)$$

The change in phase between successive components of $\boldsymbol{\varepsilon}$ is referred to as the *phase increment* of $\boldsymbol{\varepsilon}$. The two phase increments given in (4) are negatives of one another, and are determined by the unknown pulse wave velocity. Referring to Figure 1, we note that most of the energy in the waveform resides in the lower frequency components. Thus, when performing PWV estimation on observation vectors from the higher indexed DFT bins, SNR may not be sufficient to produce good results.

The presence of the reflection term in (4) makes it similar to the symmetric multipath case [7]. Multipath occurs when two planewaves originate from the same signal source, which implies they will have the same phase relationship to each other in any observation. Symmetric multipath occurs when the two planewaves impinge on the array from symmetric angles. (The Z(k) factor in (4) does not affect the phase relationship between the two vectors, even though Z(k) can change in amplitude and phase between heartbeats due to changes in waveform shape and sampling phase.)

The coherent relationship between the two $\boldsymbol{\varepsilon}(\phi)$ vectors in (4) precludes the use of certain simple estimation procedures, such as the standard MUSIC algorithm without pre-processing. However, there is a great deal of signal processing literature dealing with the DOA estimation problem in this form, and we are now in a position to apply it to arterial PWV estimation using long-axis ultrasound.

3. ALGORITHMS

The simplest approach to DOA estimation is ordinary linear beamforming, corresponding to the maximization of the discrete-time Fourier transform (DTFT) periodogram of the observation. This requires a one-dimensional search over the PWV domain to maximize the quantity

$$I(v_p)=\left|\boldsymbol{\varepsilon}^H\left(\frac{2\pi dk}{TMv_p}\right)\mathbf{Y}(k)\right|^2 \quad (6)$$

where the superscript H represents Hermitian transposition. The problem with this procedure for the present application is that the aperture is only a small fraction of the equivalent wavelength for any DFT index that contains an appreciable fraction of the signal energy. If we define the propagation speed divided by the duration of a heart cycle as the fundamental wavelength, then a typical pressure pulse will be at least three meters long. A typical 4 cm. aperture is much too small to locate the PWV even in the absence of reflections, since the entire range of physiologically reasonable PWV's is entirely within the mainlobe of the "array pattern", for small k.

For example, Figure 3 depicts a simulation of the use of the DTFT periodogram approach. The simulation uses a single noise-free recorded pressure waveform like that in Figure 1. The simulated PWV was 4 m/s, and a reflection was simulated with Γ = 0.3 and D = 2 cm. Eight measurements spaced over a 5 cm aperture were simulated, and the equivalent narrowband wavelength for the DFT index used was 0.95 meters. The forward and reflected PWV's are marked by arrows in Fig. 3, and the height of the arrows encodes the relative amplitude of the two disturbance components. The response is plotted against inverse PWV, which corresponds to angle of arrival in the DOA problem and the result for the 5 cm aperture is plotted in Figure 3 as a solid line. Note that the maximum value of this solid line occurs at 6.2 m/s. By way of contrast, a 2-meter-long, 16-element aperture was simulated using the same arterial parameters and is plotted in Figure 3 as a dashed line. This aperture allows accurate estimation of the PWV, but clearly such a data gathering arrangement is out of the question due to its size. This suggests the need for estimators that can exceed the Fourier resolution limit for the small phisical aperture.

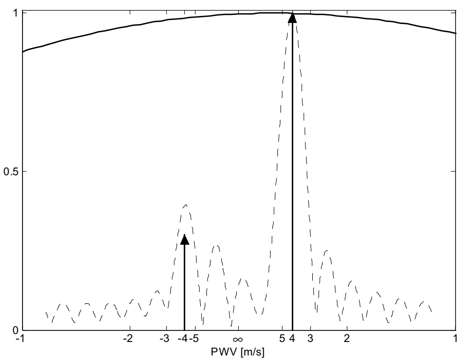

Figure 3. PWV estimation using linear Fourier method.

The Fourier resolution limit can be exceeded by nonlinear, parametric methods for high-resolution DOA estimation, most of which can be formulated as operating on the covariance matrix of the observation. In the present application, a sample covariance matrix is formed by averaging over multiple heart cycles the outer products of the data vectors from the k^{th} DFT bin.

$$\hat{\mathbf{R}}(\mathrm{k}) = \frac{1}{\mathrm{N}} \sum_{\mathrm{n}=1}^{\mathrm{N}} \mathbf{Y}_{\mathrm{n}}(\mathrm{k})\, \mathbf{Y}_{\mathrm{n}}^{\mathrm{H}}(\mathrm{k}) \qquad (7)$$

Due to the coherence of the two response vectors in (4), $\hat{\mathbf{R}}(\mathrm{k})$ will have a signal subspace dimension of one.

One way to overcome this rank deficiency is to employ methods such as least-squares [1] or maximum likelihood [7]. Note that the M-dimensional search generally required by these approaches is actually 1-D in this application. These methods have the drawback that they tend to require high SNR.

Another workable approach is eigenanalysis of the averaged covariance matrix obtained from spatial smoothing [8]. The MUSIC algorithm is easy to apply in this case because the search is one-dimensional. ESPRIT [8] is also a possibility because of the regular spacing of the measurement sites. Other methods related to ESPRIT, such as subspace fitting [8], can also be used.

In general, all of these methods will result in some number of distinct PWV estimates for different DFT bins. *A posteriori* combination of these estimates can present a problem, especially for low SNR observations. Thus the approximate coherent combination approach of Wang and Kaveh [9] may be especially well-suited to the present application.

4. Discussion

In this paper we have presented a new signal processing framework for the problem of estimating the arterial pulse wave velocity from a long-axis ultrasound data set. The attempt to apply model-based signal processing to overcome the limitations of existing methods in this application has revealed an underlying similarity to broadband direction-of-arrival estimation problems using uniform line arrays. The similarity between these two problems is more than a mere analogy. If fact, in ultrasound PWV estimation we are sensing a propagating disturbance which is constrained to travel in the "endfire" direction with respect to the array, and whose propagation velocity varies.

The use of derived measurements at every "sensor" is the primary difference between the present application and the DOA estimation using a line array. The use of such measurements can result in problems of consistency. For example, if the alignment between the wall of the artery and the ultrasound lines varies between measurement sites, then the Doppler signature of the disturbance will be different for different sites. The question of which of the high-resolution, nonlinear procedures is best suited to work using ultrasound data is currently the primary focus of our research.

5. REFERENCES

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