Efficient Design of Ultrasound True-Velocity Flow Mapping

Ultrasound imaging is one of the most prominent noninvasive imaging modalities currently in use. Its applications extend from plain soft-tissue imaging, as in abdominal sonography, to laparoscopy and blood flow/perfusion mapping. Blood-flow mapping is one of the most important applications of ultrasound. In this application, 2D maps of blood velocity values are generated and displayed atop B-mode ultrasound images to correlate with the underlying anatomy (Fig. 1). The velocity values are color-coded in red or blue to distinguish them from the usual gray-scale used for B-mode ultrasound images. The red-scale is used for velocity values directed toward the transducer while the blue-scale is used for those directed away from the transducer. These types of flow maps with this particular color coding are usually referred to as color Doppler images. These images are of great value in many diagnostic applications, such as echocardiography and perfusion imaging. Also, their potential in novel tumor-detection techniques, such as sonoelasticity, and for adding a new flow dimension to quantitative tissue characterization is currently under investigation [1-2].

In general, currently available techniques for generating flow maps can be classified into two main categories: frequency-domain techniques and time-domain techniques. Both classes have their relative advantages and disadvantages as they try to provide solutions to the common problems of flow mapping, such as limited velocity estimation accuracy and mapping resolution. Nevertheless, the conventional forms of both techniques share one major limitation in that they are able to map only one or two components of the 3D flow velocity in using 1D transducers. Given the random orientation of flow velocities at different locations throughout the image plane, it is obvious that the resulting flow maps consist of velocity values that cannot be directly compared since they do not have the same reference. This problem has profound effects on the diagnostic abilities of the techniques and certainly can prevent quantitative characterization of image regions. To be able to solve the problem of creating true-velocity maps, we have to go back and revise our classical velocity estimation models and propose more general models instead.

In this article, we describe the general theory of the classical Doppler technique for flow mapping and show that its main assumptions do not generally hold for ultrasound imaging. We then develop a generalized model for frequency-domain flow mapping in practical ultrasound imaging situations. Using this model, we show that it is possible to compute the true velocity from single-aperture configurations. We discuss improving the resolution and velocity estimation accuracies and propose a novel approach based on a generalization of the radar-ambiguity function model. We also consider the same problems for time-domain techniques. We propose a generalization of the correlation technique that takes into account the ultrasound field effect, and show that it is theoretically possible to obtain true-velocity flow maps from single-aperture configurations. Finally, we discuss the relative advantages and disadvantages of both frequency-domain and time-domain techniques.

Frequency-domain Flow Mapping

Let us consider the class of methods that derive flow maps directly from the power spectrum of the received pulse-echo-mode ultrasound signal. The traditional way of doing this is to use the Doppler equation, which has the form:

\[ f_0 = 2 \frac{v \cdot \cos \phi}{c} f_r \]

Here, \( f_0 \) is the frequency shift between the received signal and the transmitted signal, \( v \) is the true-flow velocity, \( c \) is the velocity of ultrasound in tissues, \( f_r \) is the transmitted
frequency, and $\phi$ is the angle between the direction of flow and the transmitter/receiver transducer axis. Given the transmitted frequency and assuming the velocity of ultrasound in tissue to be constant and equal to 1540 m/s, the true-flow velocity value can be estimated from the Doppler equation by measuring the frequency shift and the exact 3D geometry of the problem, as characterized by $\phi$. Unfortunately, the geometry of the problem is usually not available and quite difficult to assess. As a result, the above equation can be used to estimate the projected velocity vector, $v \cos \phi$, which represents the component of the true velocity in the direction of the transducer axis, or axial velocity (Fig. 2). It follows that computation of the true velocity using this method would theoretically require at least three non-coplanar transducers to be able to reconstruct the 3D magnitude from three independent projections.

The Doppler equation suffers from a number of major problems that can greatly limit its value for flow-mapping applications. It is derived under the assumption of monochromatic plane wave excitation. Hence, it may not be valid in describing practical ultrasound imaging situations. Although this condition can be approximately satisfied with a very small transducer and a very long excitation signal, the resultant system cannot be used in flow mapping because of its extremely poor resolution in both range and azimuth. As a result, the effects of finite transducer size and excitation length appear in the form of distortions in the computed power spectrum. These distortions are commonly known as geometric and transit-time broadening effects, respectively [2]. Another problem with the Doppler model is the complexity of the practical implementation of true-velocity estimation systems from multiple transducer configurations. Temporal and spatial registration, in addition to limited vessel accessibility, are examples of the problems encountered in this area. For these reasons, the currently available flow-mapping techniques based on the Doppler model provide an axial velocity map instead of the more convenient true-velocity map.

**Generalized Power-spectrum Model**

Consider now a more general model for the process of ultrasound flow mapping (Fig. 3). In this model, we do not assume particular forms for the ultrasound field and excitation signal. We only assume that Fresnel propagation conditions are satisfied [3]. We proceed to derive an analytical expression for the power spectrum of the received signal as a function of the imaging conditions, as modified by the 3D flow (Fig. 4). Given this expression, we can then consider the problem of estimating the true-velocity parameters by measuring the power spectrum. In deriving our model, we try to associate all assumptions with known physical quantities/phenomena in the actual process.

In laminar flow, blood, consisting of plasma and formed elements, moves in a particular direction with uniform velocity. Given that the plasma itself is a homogeneous fluid, we do not expect to receive any reflection/scattering phenomena from ultrasound waves propagating through this medium. On the other hand, the formed elements — mainly red blood cells (RBC) — will be a source of scattering. This scattering can generally be assumed to be uniform (Rayleigh) scattering [4]. It is usually of very weak intensity, and that is the reason why the inside of blood vessels appear dark in B-mode images. Nevertheless, this scattering phenomenon is the sole source of information for velocity estimation in flow mapping. In uniform flow, the RBCs will be distributed fairly uniformly in the volume inside the vessel, in a random but fixed fashion for any given small period of time. Even though the randomness assumption is not hard to justify, the assumption of a fixed pattern may need some explanation. A time-varying pattern inside a uniform flow must be due to space-varying local forces inside the moving fluid that can act on the solid blood elements to cause their relative motion and, hence, their change in the moving pattern. Forces of this nature are usually associated with flow acceleration, which is not the case in our situation. In other words, our model holds for brief time intervals during which the assumptions of constant velocity and zero acceleration are both valid. In this case, we may view...
the pattern of RBCs as a large but finite collection of parallel straight lines of ideal Rayleigh scatterers. The distributions of the scatterers along these lines are assumed to be independent and identically distributed with a Poisson impulses random distribution. Under the conditions of Fresnel propagation, the return power spectrum from a single line of moving scatterers takes the form [5]:

$$
\Phi_{\text{RR}}(\omega) = \frac{\lambda^* \sigma^2_j}{\nu_1 (1 + 2 \frac{v_z}{c})} \left| U(\frac{\omega}{\nu_1}) \right|^2 \left( \frac{\omega}{1 + 2 \frac{v_z}{c}} \right)^3
$$

where $\Phi_{\text{RR}}(\omega)$ is the power spectrum of the received signal; $\lambda^*$ is a function of the Poisson model parameter; $\sigma_j$ is the scattering cross-section of the effective scatterers; $c$ is the phase velocity of ultrasound in tissues; $\nu_1$ and $\nu_2$ are the transverse and axial velocities, respectively; $U(\cdot)$ is the Fourier transform of the effective transmit/receive field along the line of motion; and $S(\cdot)$ is the Fourier transform of the excitation signal. From this expression, we can see that the axial velocity appears as a scaling of the excitation signal, while the lateral velocity appears as a scaling of the beam pattern. Scaling means dilation or shrinking of the original function. The effect of velocity values on the magnitude of the power spectrum is of little use without a reference magnitude for comparison.

The above description suggests that these scaling effects can be measured by observing the power spectrum and choosing the velocity parameters by directly fitting this observed spectrum to the model. This is not possible, in general, in practical situations, since the received echo is the superposition of the return from many lines of scatterers experiencing different fields. It is easy to see that space shift variations in the power spectrum shape are quite complex and mathematically intractable in practical situations where the spatial extension of the volume of moving scatterers is not known. As a result, we must find common features among all flow lines, which are only functions of the velocity parameters, given some general description of the field at the depth of the flow.

Two main approaches have been proposed to solve the problem of spatial variation. The first is to assume that the flow volume is small compared to the field at the depth of flow, and therefore the field at all lines is basically the same. This approach can be quite restricting in many cases. For example, if the flow is in the focal plane of a thin lens in the usual transducer-against-lens configuration, or is in the far field [6-7], the field has a quadratic phase term. This quadratic phase term requires the flow to be in the center of the field, in addition to being extremely localized away from the areas of the field where the phase variations cannot be overlooked. Moreover, the extension of the flow in the axial direction should be very limited to be able to assert range invariance. Therefore, we propose a more flexible approach to ensure range and lateral-flow-shift invariance [8]. In this approach, we design the field such that its angular spectrum is of compact support. This condition does not imply that the field will be the same for all lines anywhere in the image plane. However, it ensures that some basic properties of the measured power spectrum, such as its bandwidth and frequency shift, will be space-invariant. Given that these properties are measurable and can be directly related to the velocity parameters, the process of velocity estimation can be transformed into a space-shift-invariant problem (Fig. 5).

Another important observation can be made from the generalized model. If the excitation field is chosen to be circularly symmetric, it is clear to see that the x-axis of our model can be arbitrarily selected to be the direction of the projection of the true velocity onto the transverse plane. As a result, the estimated axial and lateral velocity components are basically the only two components of the true-velocity vector. Therefore, the true-velocity magnitude can be obtained once these two components are computed. Consequently, it becomes possible, in principle, to generate true-velocity maps from single coplanar aperture configurations.

**Improvement of Frequency-domain-based Flow Mapping**

As in all measurement procedures, the parameter estimates deviate from their true values. The estimation errors can be essentially divided into two major sources: systematic (determini-
nistic) errors, due to the measurement procedure itself, and random errors, due to contamination of the measurables by noise. Random errors can be controlled by taking more measurements and averaging them. On the other hand, deterministic errors are not as easy to eliminate, in general, because they are unique for each problem. These errors can be minimized by optimizing the measurement procedure for the problem at hand, given its associated practical constraints. In the case of flow mapping, the problem is to measure and map true-velocity values in the 2D image plane. Our measurable quantities here are the axial and lateral velocities and their spatial coordinates. We therefore must construct a model for our measurement procedure and consider the optimization of this procedure for the practical imaging time and aperture-size constraints of ultrasound imaging.

**Generalized Ambiguity Function Model**

Given the similarities that exist between medical ultrasound imaging and radar and sonar imaging, many of the mathematical tools used to study problems in the latter have been adapted to ultrasound imaging. The most important of these tools is the range-Doppler shift-ambiguity function. This function captures the response of a given imaging system as described by the excitation signal. In particular, it defines the resolution in the range direction and the accuracy of the velocity estimation from the Doppler shift (Fig. 6). It is not difficult to see that because of the Fourier uncertainty principle, there will be a trade-off between the improvement in these two directions. As the excitation signal gets longer, the axial resolution becomes worse. On the other hand, a longer signal provides more accurate estimates of the Doppler shift and, hence, the axial velocity. Therefore, it is clear that an ideal ambiguity function of a δ-function form is impossible to achieve. The mathematical description of this problem provides a convenient way of comparing different possible excitation design solutions. Moreover, several successive improvement schemes have been proposed to achieve desirable accuracy levels simultaneously in both the axial resolution and the Doppler shift.

Although this approach is quite sufficient to describe radar and sonar imaging, it does not provide a complete description of medical ultrasound imaging. The ambiguity function assumes that the Doppler equation model is valid. As we have seen, this model does not generally hold in ultrasound imaging. This means that the lateral resolution and lateral velocity accuracy are left out in the present improvement schemes. Therefore, it is necessary to develop a generalized version of the radar-ambiguity function to devise improvement schemes that take into account the resolution in the 2D image plane, in addition to both components of the true velocity described in our generalized model. In [9], we developed the generalized ambiguity function (GAF) description of the true-velocity flow mapping. We showed that this function is of the form:

\[
GAF(x, z; v_x, v_y) = \int \left[ a(x + v_x t) \cdot u(x + v_y t) \right] \left[ s \left( \frac{1 + 2v_x}{c} \right) - \frac{2v_x}{c} \right] \left( \frac{1 + 2v_y}{c} - \frac{2v_y}{c} \right) dt
\]

where \( x \) and \( z \) are the location parameters in the transverse and axial directions, respectively; \( v_x \) and \( v_y \) are the transverse and axial velocities, respectively; \( (x^*, z^*, v_x^*, v_y^*) \) are their actual values; \( a() \) is the effective transmit-receive field function; \( s() \) is the excitation profile; and \( c \) is the ultrasound phase velocity in tissue. As can be seen, the function yields the radar-ambiguity function in the special case when the excitation field is a plane wave. In general, the resolution and velocity accuracy in the axial and lateral directions are coupled. Any successive improvement scheme should be based on combinations of excitation signals.
pattern has moved a distance $vT$. Given that the received signal maps the axial component of this displacement, it is again possible to estimate the axial velocity by detecting the peak of the correlation between the received signals from these two consecutive excitations by using the equation [11-12]:

$$
\tau_{peak} = \frac{2 \cdot \Delta T \cdot \cos \theta}{c}
$$

Here, $\tau_{peak}$ is the time shift of the correlation peak, $\Delta T$ is the distance traveled by the scattering pattern in the pulse-repetition period $T$, $v$ is the axial velocity, $c$ is the speed of sound in tissue, and $\theta$ is the angle of inclination of the flow direction onto the axis of the imaging transducer. As can be seen, this technique is similar to the Doppler equation model in that it provides only one component of the true velocity. Nevertheless, the two techniques differ in two respects: the complexity of computation and the relationship between velocity estimation and mapping resolution. The time-domain technique has a much heavier computational complexity because of the need to compute the correlation function from long RF sample streams. On the other hand, the trade-off between the accuracies of velocity estimation and mapping resolution does not exist in time-domain technique theory — that is, very short excitations can be used to give good resolutions in both dimensions of flow mapping. With the advent of custom-made hardware to handle the large number of computations required in a timely manner by the correlation, time-domain techniques are becoming more appealing as the new standard for flow mapping. However, note that the time-domain methods require at least two successive excitations. Therefore, the total illumination time is double that of the frequency-domain methods, which can theoretically obtain the velocity estimate with one excitation.

The major limitation of time-domain techniques is the difficulty of obtaining true-velocity maps from practical configurations. The theory that supports these techniques assumes that the excitation field is basically a plane wave. To understand the effect of other fields that are used in practice, we develop a generalized model for the flow-mapping process using time-domain techniques.

### Generalized Correlation Model

If we start with the same assumptions as in the generalized power-spectrum model, we can show that the correlation function between two consecutive excitations with a pulse-repetition period $T$ is given by [13]:

$$
R_s(t, t + \tau) = h^2 \cdot G_s(t) \cdot G_s^*(t + \tau) + \lambda \cdot G_f(t, \tau).
$$

Here:

$$
G_s(t) = \int_{-\infty}^{\infty} u(r \sin \theta + v \cdot t) \cdot \frac{1}{c} \left[ t(1 + \frac{v}{c}) - 2 \cdot \frac{rcos \theta}{c} \right] dr
$$

and
\[ G_2(t) = \frac{1}{\pi} u(r \sin \theta + v_1 t) \cdot \left( \frac{r + \Delta r \sin \theta + v_1 (t + \tau)}{\left(1 + \frac{2 \Delta r}{c} \cdot \frac{c}{r \cos \theta}ight) \cdot \left(1 + \frac{2 \Delta r}{c} \cdot \frac{c}{r + \Delta r \cos \theta}ight) \cdot \frac{c}{r \cos \theta}} \right) d\tau. \]

Notice that \( G_2(\cdot) \) is the effective component representing the excitation/field correlation function, while \( G_1(\cdot) \) is less important and can be set to vanish by design in many cases [13]. As long as the above expression is rather complex, especially because of the possible coupling between its excitation waveform component and the ultrasound field component. Therefore, we consider important special cases of this expression to illustrate the idea behind the conventional correlation technique. We also suggest a new method of obtaining the true velocity from a single transducer using this technique.

We consider first an ideal plane-wave excitation that has a very brief time duration. Let us approximate this excitation signal by a \( \delta \)-function. In this case, the correlation function is strongly peaked at a single location as a result of its impulse plus constant form (Fig. 8a). The location of this peak or impulse is given by:

\[ \tau_{\text{peak}} = \frac{2 \cdot \Delta r \cdot \cos \theta}{1 + \frac{2 \Delta r}{c} \cdot \frac{c}{r \cos \theta}}. \]

This expression is the same as the one suggested in the literature. Given the usual difficulty of obtaining a value for \( \theta \), this expression has been used to compute the axial velocity as:

\[ v_z = \frac{\Delta r \cdot \cos \theta}{T} = \frac{\tau_{\text{peak}} \cdot c}{2 \cdot T}. \]

For most practical cases, the excitation must have a finite time width. This finite time width smears out the single strong peak of the correlation function corresponding to the ideal case (Fig. 8b). This means that the correlation will still have a peak at the same location as in the ideal case, but this peak is not as well defined. Moreover, the strength of this peak gets weaker as the excitation becomes longer. As a result, we would theoretically expect larger peak-detection errors from longer excitations than from brief ones.

Another important special case is when the excitation is a narrow-band signal with a finite time width. This is the usual case in conventional ultrasound imaging. In this case, the correlation function will be narrow-band, with an envelope that has exactly the same form as the smeared-out function in the previous case (Fig. 8c). This brings about the possibility of missing the peak of the correlation function by up to one full excitation period if the peak detection is performed on the correlation function directly. Therefore, we would suggest demodulation of the correlation function to try to eliminate this problem.

Let us now consider the general case when both the excitation signal and the field are general functions. In all of the above special cases, the assumption was that the field was a plane wave. The consequence of this assumption is to be able to say that for any pulse repetition period \( T \), the correlation function will have the same form and peak magnitude, but with different shift. This is not true in practice, where strict conditions must be imposed upon \( T \). To understand this, consider the 1D case of a rectangular field. It can be easily seen that after a certain amount of time, all the scatterers that experienced the original excitation will have moved outside this rectangular field. As a result, the correlation between any signals obtained at later times will be completely uncorrelated with the original signal. Therefore, for accurate peak detection, short pulse-repetition periods are desirable to minimize this decorrelation effect.

In spite of the problems associated with this decorrelation effect, we can indeed use it to our benefit by identifying its sources. It can be seen that this effect will depend on both the beam profile at the level of flow, as well as the flow velocity in the lateral direction, which moves the scatterers outside the center of the field. Given the ultrasound field and assuming that we are using circularly symmetric apertures, we can actually derive a formula for the lateral velocity component of the flow in terms of the rate of decorrelation. In particular, we have [13]:

\[ v_z = \frac{\alpha}{W_T}. \]

Here, \( W_T \) is a suitable bandwidth measure the period between the first excitation and the time when the return signal from an excitation is nearly uncorrelated with the first signal return, while \( \alpha \) is a fudge factor that can be determined once the ultrasound field is known and is circularly symmetric. The parameter \( \alpha \) also depends on the specific bandwidth measure in use. This method is called decorrelation tracking, since it is based on observing several correlation peaks from consecutive excitations (Fig. 9).

The lateral velocity is computed from these peaks. Clearly, a minimum of two correlation peaks (three excitations) is required to obtain a lateral-velocity estimate. Nevertheless, more correlation peaks should be used to improve the estimation accuracy. It should be noted that the method of computing the axial velocity remains the same, since in practice, the determination of the peak location is predominantly a function of the axial velocity. Also, as discussed above, it should be kept in mind that the true velocity can be computed from the axial and lateral components based on the symmetry of the field. Hence, using the correlation tech-
nique, it is possible to obtain true-velocity maps from a single aperture.

Conclusions

It is possible to reconstruct true-velocity flow maps using ultrasound imaging from a single coplanar aperture by either frequency-domain or time-domain techniques. The advantages of frequency-domain techniques are the relatively lighter load of computations, which allows real-time processing, plus the theoretical possibility of single-shot true-velocity mapping. Nevertheless, these techniques require more excitations from different apertures to achieve good flow-mapping accuracy, as shown by the generalized ambiguity function model. The advantage of the time-domain technique is its relatively higher flow-mapping accuracy with very short excitations. On the other hand, these techniques are computationally extensive and require several excitations to be able to estimate the true velocity. The decision as to which technique to use should take into account their relative advantages/disadvantages in a particular situation.

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9. Illustration of the concept of the decorrelation tracking technique.

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