Recent Developments in Ultrasound Imaging Velocimetry: toward Clinical Application

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An overview is given of recent progress in the field of ultrasound imaging velocimetry (UIV, also known as echo-Particle Image Velocimetry). UIV is capable of providing instantaneous velocity fields in flows that are not accessible by conventional (i.e. optical) measurement techniques. In particular, it holds great promise for non-invasive measurement of blood flow patterns and the related wall shear stress distribution. These parameters can assist in e.g. evaluating atherosclerosis risks. While some promising proof-of-principle experiments have been performed recently, quantitative clinical applicability will require significant improvement of the technique, e.g. the accuracy of near-wall velocity results and the dynamic range. The UIV technique is compared to a range of alternatives, such as MRI-based velocimetry. Finally, translation of the technique to applications other than blood flow (e.g. multiphase flows) will be briefly discussed.

Keywords: ultrasound imaging velocimetry, PIV, blood flow measurement

1 INTRODUCTION

There is a strong drive to improve the *in vivo* measurement of blood flow patterns. Better knowledge of hemodynamic conditions will facilitate fundamental studies into the role of hemodynamics in cardiovascular development and in pathologies (e.g. during the development of aneurysms or atherosclerosis). Apart from being useful for fundamental studies, access to such hemodynamic information would be an important diagnostic tool. For obvious reasons, non-invasive measurement techniques with sufficient spatial resolution are desirable. This resolution criterion also stems from the fact that often wall shear stresses need to be derived from the velocity fields. The use state-of-the-art optical flow measurement techniques, such as (microscopic) particle image velocimetry (PIV), are limited due to the opaque nature of blood and the limited optical access. Only in a limited number of cases those techniques can be applied, for instance in the embryonic chicken – a common model system for human cardiovascular development [1]. Direct application of these optical techniques is unfortunately not possible in human patients or healthy volunteers.

In recent years, ultrasound imaging velocimetry (also known as speckle velocimetry or “echo-PIV” [2-5]) has been introduced as an alternative technique that can provide the desired blood flow patterns non-invasively [6-10]. As it is largely based on existing echo-/sonography hardware and protocols, it is an accessible and relatively cheap method. The method uses cross-correlation algorithms to track features in subsequent frames of an image sequence, similar to optical PIV. The main difference is the methodology to obtain images (using an ultrasound transducer rather than a camera). Note that in contrast to *Doppler* velocimetry, it provides instantaneous velocity fields of *two* velocity components. The related, additional benefit is that it does not rely on exact knowledge of the insonification angle; furthermore, the latter does not need to be optimized with respect to the mean flow direction, as two velocity components are available (i.e. not just the one along the direction of sound).

2 EXPERIMENTAL TECHNIQUES

2.1 Typical set-up and data acquisition method

To describe the basic techniques and procedures, a simple *in vitro* experiment is described that aims at measuring the velocity field of a steady laminar flow in a tube. More details are given by Poelma et al. (2010) [5].

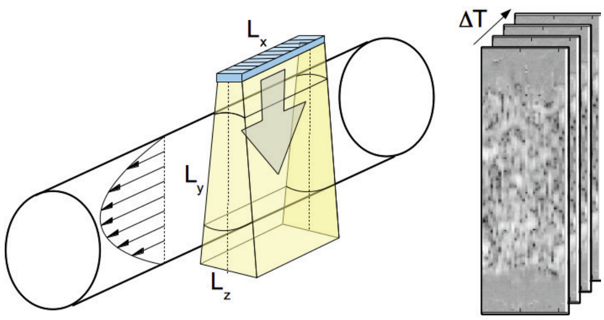

Figure 1: schematic representation of an UIV experiment (*left*) overview of geometry and orientation of the transducer (*right*) a sequence of B-mode images. Image adapted from Poelma et al., (2012) [10].

A transducer, here a linear phased array with 128 elements, is aligned with the tube axis (see Figure 1). As the working fluid by itself (water) has very little scattering efficiency, the flow must be seeded with a material with good acoustic scattering properties. “Good acoustic scattering” can be realized by using a dispersed material that has a large difference in acoustic impedance compared to the surround fluid. Here, a small amount of SonoVue contrast medium (SF_6 microbubbles) is added to the water. These scatterers are imaged in subsequent frames by the transducer, which emits at 7 MHz. Data is recorded as RF signals using an Ultrasonix RP500 system. The data can be converted (offline) to so-called B-mode images by means of envelope detection and log compression. The latter step is mainly used for visual inspection, but is not required (and often actually detrimental) for flow analysis. Image acquisition rates are typically in the tens to hundreds frames per second, see later.

2.2 Data processing

The sequence of RF data is processed using local cross-correlation, as in conventional PIV [11]: the total image is divided into smaller regions, so-called interrogation areas. The size of these interrogation areas ultimately determines the spatial resolution of the measurement. However, smaller interrogation areas also lead to a lower signal-to-noise ratio. Therefore, for each experiment (flow and imaging conditions, tracer concentration) the optimal size must be determined. Typically, areas of 16×16 to 64×64 pixels are used, with 50% overlap between neighboring areas. Cross-correlation of a pair of interrogation areas results in a “correlation plane” with a distinct peak at the location of the optimal shift of the particle image patterns, provided that there is a sufficient amount of particles that can be matched (see section 3). The cross-correlation is shown schematically in Figure 2. The red and blue objects indicate tracers at frame 1 and 2, respectively.

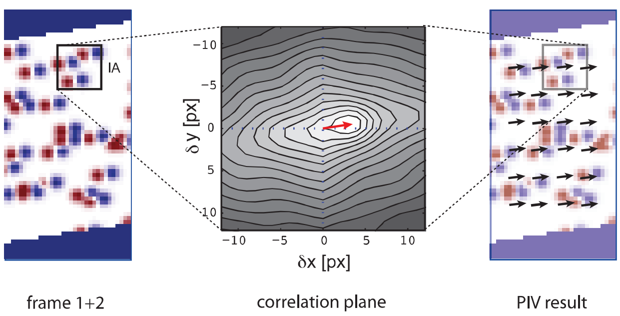

Figure 2: local cross-correlation between subsequent image frames provides the velocity field based on the mean local displacement of tracer particles (e.g. contrast bubbles).

The cross-correlation of interrogation areas is repeated for the entire image, so that a complete displacement field is generated for each image pair. The displacements can be converted to velocities by multiplication with a scaling coefficient (pixel/meter) and division by the time between subsequent frames. Note that the scaling coefficient is usually different for the x and y direction: the former is determined by the transducer pitch, while the latter is related to the wavelength of the ultrasound. The time between two subsequent frames is largely determined by the imaging acquisition rate of the ultrasound system. However, it should be noted that images are not recorded as snapshots, but are usually constructed line-by-line (sweep or `rolling shutter’). This means that – in the present case – any horizontal displacement will require a correction of the time between frames (ΔT); see e.g. Poelma et al. (2011) or Beulen et al. (2010) [4, 5] for a detailed discussion.

2.3 Typical results

A typical UIV result consists of one or more instantaneous two-dimensional velocity fields, describing the flow in a slice extending from the transducer (see also Figure 1). This transducer also determines the width L_x of the field-of-view (assuming that all transducer elements are used). The depth L_y is an acquisition setting. Note that larger values of L_y lead to lower frame rates (due to the finite velocity of sound, typically around 1500 m/s). The quality of the image decreases with increasing y coordinate due to attenuation and divergence of the ultrasound beams. The thickness of the measurement slice (L_z) is typically of the order of 1.5 mm and determined by the hardware design and beam focusing settings [10].

An example of a typical velocity field and a reference B-mode image are shown in Figure 3. Here, the flow in a curved, non-transparent tube has been measured; note the strong asymmetry in the flow profile - increasing with the flow from left to right - due to secondary flow patterns. The total field-of-view was 4.9×2.5 cm^2, with an in-plane spatial resolution of 0.5 mm. Errors in the averaged flow velocities were estimated to be below 1% of the maximum velocity.

2.4 (Phase-)Averaging strategies

For steady or periodic flows it is possible to significantly enhance the signal-to-noise ratio by averaging the results. In particular, averaging the intermediate *correlation planes*, rather than the vector fields, is known to greatly improve the results [11, 12]. Averaging is often essential due to the relatively low signal-to-noise ratio of UIV images, as compared to conventional PIV images. For steady flows, the averaging is straightforward. However, one must be careful with the interpretation of averaged PIV results from transient flows [13].

For periodic flows, the image sequence can be sorted before averaging by making use of an external trigger signal or by “self-gating”. In the latter method, a rough estimate of the flow field is used to determine the phase of the image frames; data with a similar phase is subsequently averaged [12]. For data that is sufficiently oversampled (i.e. the frame rate is relatively high compared to the highest frequencies of the flow), a sliding averaging approach can be utilized. In this method, a vector field is not obtained from a single image pair, but from a (small) series of frames (e.g. averaging frames 1-5, 3-7, 5-9, etc.). This technique can obviously be applied to any transient flow, not just periodic flows. Due to the high imaging rate requirements (discussed in the section 3), many flows studied by UIV are oversampled and thus suitable for this sliding averaging processing.

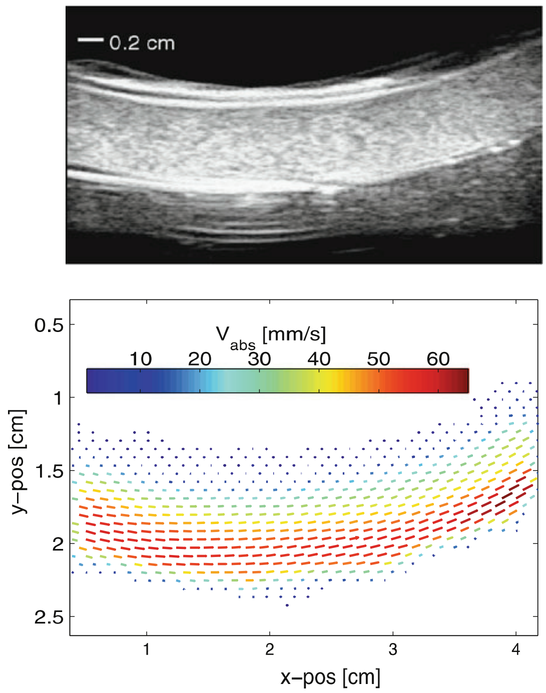

Figure 3: A typical velocity field resulting obtained with UIV; top image shows one B-mode image; the bottom figure is the corresponding mean flow pattern. Image adapted from Poelma et al. (2011) [5].

3 DISCUSSION

3.1 Resolution, dynamic range

The resolution that can be achieved by a UIV measurement depend on a number of factors: the *image* resolution in the lateral (“*x*”) direction is mostly dictated by the transducer design, i.e. the pitch between elements. However, the way a vertical image line is obtained (e.g. by firing multiple elements, beam forming) can lead to overlapping “cones” of neighboring imaging lines. This in turn can introduce artifacts when one tries to achieve sub-pixel accuracy in the displacement [10]. For each velocity vector, a number of horizontal pixels is used (the width of an interrogation area), typically 8-64. This number, multiplied by the transducer pitch, serves as a good estimate of the horizontal measurement resolution.

In the axial (“*y*”) direction, the image resolution is determined by the frequency of the ultrasound. Higher frequencies result in a better resolution. Depending on the frequency, axial image resolutions range from 1 mm (low frequency, 2-5 Mhz) to as fine as 30 micron (high frequency, 30-50 Mhz). However, higher resolutions come at the price of higher attenuation (and thus signal degradation). In practice, one thus has to compromise between measurement depth and resolution. Typical ultrasound frequencies in the range of 2-50 Mhz are used, with corresponding penetration depths from tens of centimeters to a few millimeters. As stated earlier, the axial resolution is usually significantly better than the lateral resolution. Values of the spatial resolution of an UIV *measurement* (not *image*) as small as 0.1 and 0.4 mm respectively have been reported. This was achieved for a time-averaged flow, which permitted small interrogation areas [14].

Apart from resolution, the dynamic range of the UIV technique needs to be discussed. This range describes the minimum and maximum velocity that can be measured. The location of the displacement peak in the correlation plane (Figure 2) can in practice only be determined with a finite accuracy. This means that there is a lower limit for the velocities that can be measured. For conventional PIV, this lower limit (i.e. the random error in the displacement) is usually around 0.1 pixel for properly optimized experiments; it is questionable if the same can be achieved for UIV due to the noisy character of the data. Larger displacements will mean the relative contribution from the inaccuracy in the peak location will decrease. However, for the correlation algorithm, it is essential that features can be tracked: a certain fraction of a group of particles must remain within the interrogation area used for the correlation. For large displacements, particles will move out of the interrogation area (or even out of the total field of view)[15]. This means that the displacement must be optimized to balance between the two errors (peak location, loss of particles). In practice, a displacement of 4-8 pixels is common in PIV. This directly dictates the required time between subsequent frames – assuming that the velocity to be measured is fixed. For cardiovascular flows, which can be as fast as 1 m/s, this means that imaging rates of 100-1000 fps are required [10]. This is beyond the capabilities of many ultrasound machines. However, frame rates can be increased by using a smaller subset of the transducer elements. Alternatively, dedicated hardware can be used, which is designed for velocimetry, rather than conventional imaging [16].

3.2 Comparison with other techniques

Ultrasound is a relatively cheap and accessible measurement technique that can provide velocity fields in applications without optical access. In recent years, other techniques have been introduced that have similar capabilities. In particular, techniques based on magnetic resonance imaging have shown to be very powerful [17]. The resolution that can be obtained is superior: for instance, Van Ooij et al. report a resolution of 0.2x0.2x0.33 mm^3 in a pc-MRI experiment using a 3T system with a 7 cm coil [18]. Furthermore, these measurements provide volumetric/3D data, something that has currently not been demonstrated using ultrasound (unless the transducer is translated [5]). Such MRI measurements require a long measurement time, however. A further drawback of MRI is the complexity of the hardware, which translates into high costs (acquisition and operating) and personnel training requirements. Finally, the strong magnetic field restricts the use of materials that can be used in/near the facility, so that experiments need to be designed carefully.

Another recent technique for flow measurement in non-transparent flow utilizes x-rays. These measurements are based on the attenuation of rays from one or more radiation sources [19]. As the attenuation represents the projection along the trajectory (i.e. the average along a line through the flow domain), some form of tomographic reconstruction is required to obtain a 3D spatial field. This means that multiple projections are necessary to reconstruct the average flow field, e.g. by either rotating the source or the actual flow geometry. If the flow is rotationally-symmetric, the flow field can be reconstructed from a single view [20]. While very promising for particular applications, the safety restrictions and hardware complexity make these techniques much less accessible than ultrasound-based techniques.

3.2 Translation to other application areas

The majority of research in UIV is performed with application to cardiovascular flow in mind. This is understandable, as the hardware that is used is already present in clinical settings. Nevertheless, it is interesting to note that one of the first flow measurements using UIV was in the field of sediment transport [21]. In this study, flow fields were studied in densely-laden liquid flows that were not accessible to optical techniques. Surprisingly, there has been little follow up with similar applications. UIV seems a perfect fit for measurement of multiphase flows. The presence of two phases (with likely a difference in acoustic impedance, e.g. oil droplets or fibers in water) seems naturally suited for UIV. Certain conditions may need to be avoided, however: for instance, very large air bubbles will lead to strong reflections and shadowing effects. Relatively little is known how to optimize these ultrasound images for velocimetry – i.e. what kind of post-processing improves flow velocity measurement, rather than imaging? Some techniques can be borrowed (e.g. harmonic imaging), while others may need to be avoided (general “image enhancement” filtering).

A drawback that remains is the limit of maximum velocities that can be measured. In particular, this limits the application in many industrial flow applications. New hardware [16] and advanced imaging methods are presently being developed to remedy this.

4. CONCLUSIONS

Ultrasound Imaging Velocimetry can provide instantaneous velocity fields in flows without the need for optical access. This makes it possible to measure, non-invasively, in opaque flows. Compared to alterative techniques (based on MRI or x-rays), it is relatively simple, safe and cheap and can have a comparable spatial resolution. While the majority of development work so far has focused on cardiovascular flow, the technique seems very suitable to applications in other application areas. It is very likely that successful studies in e.g. process technology, food industry and other areas will be reported in the literature soon.

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