Digital high speed holography for the measurement of the 3C-2D velocity field in confined flows

Virginia R. Palero¹, Julia Lobera¹, Pilar Arroyo¹

¹: Department of Applied Physics, I3A-University of Zaragoza, Zaragoza, Spain, palero@unizar.es

Abstract In the last years, several techniques have been developed and improved in such a way that it is possible to measure the three velocity components in a flow. Some techniques as Stereoscopic PIV or holographic PIV are well known. Others, like tomographic PIV started to develop more recently. Stereoscopic Particle Image Velocimetry (SPIV) or Tomographic PIV need a complex set-up, involving flow visualization with several cameras, and present important restrictions when applied to confined liquid flows. Others like Holographic PIV cannot be applied in flows with a high particle density. Here a new technique, called High Speed Digital Image Plane Holography has been developed for the measurement of the three velocity components in a complex geometry brain aneurysm model using a configuration similar to regular PIV. This was done by using a two cavity high speed laser, one double frame high speed camera and normal visualization. The two in-plane velocity components were obtained from the holograms reconstructed intensity distribution using a PIV analysis. The out-of-plane component was obtained from the reconstructed phase differences using an interferometric method. Due to the laser short coherence length, a portable and compact system has been built in order to be able to the measure larger fluid areas.

1. Introduction

In these days there is an increasing interest in the study of cardiovascular diseases, as they are one of the main causes of death in the developed countries. The formation of brain aneurysms, also called cerebral or intracranial aneurysms, is one of the most important. A brain aneurysm is an abnormal dilatation of one of the arteries in the brain (Stehbens, 1972). In many cases the aneurysms are discovered when they rupture, as often patients experience no symptoms, causing bleeding into the brain. A haemorrhage from a ruptured brain aneurysm can lead to a hemorrhagic stroke, brain damage and death.

The reasons under the genesis, growth and rupture of brain aneurysms are not well known yet. This is why medical specialists in different fields (neurologist, neuroradiologist, neurosurgeons,….) are involved in the diagnosis and treatment of patients with brain aneurysms, giving an idea of the complexity of the illness. Today, with the improvement in the neuroradiology techniques, unruptured brain aneurysms are detected more often and the preventive surgery is increasingly considered as a therapeutic option. This situation justifies the interest in making experimental measurements of the blood flow inside the aneurysms, as it is known that the local hemodynamics is one of the key factors in the aneurysm formation and evolution.

In the last years, several techniques have been developed and improved in such a way that it is possible to measure the three velocity components in a flow. Some techniques as Stereoscopic PIV or holographic PIV are well known. Others, like tomographic PIV started to develop only recently.

SPIV (Arroyo and Greated, 1991; Willert 1997) measures the three velocity components in a fluid plane combining the projection of the velocity field over two directions (2D-3C technique). SPIV requires a large optical access that can eventually prevent its application in confined flows (Palero et al. 2000) Besides, its application in liquid flows is also difficult as the change in the refractive
index introduces strong distortions in the images (Prasad and Adrian, 1993; Prasad and Jensen, 1995).

Holographic PIV (Barnhart et al., 1994; Hinsch, 2002; Meng et al. 2004) is a full 3D-3C technique that allows the recording of a fluid volume. The light scattered by the object (tracers in a flow, drops, bubbles,…) interferes with a reference beam in a photosensitive device. Therefore, both the amplitude and the phase of the object wave are captured. However, the size of the reconstructed particle images along the optical axis is several times bigger than the real size, making difficult to determine the position of the particle, preventing the application of the technique in a system with a high particle concentration, and leading to a poor spatial resolution.

In Tomographic PIV (Elsinga et al., 2006) a fluid volume is illuminated and the scattered particle light is recorded simultaneously from several viewing directions. It requires several CCD cameras (typically 3 to 6) and a calibration procedure similar to the calibration done in SPIV. Although a 3D-3C measurement can be done, it is at the cost of using a much more complicated set-up and image processing. Besides, as in SPIV, the application in confined and in liquid flows is seriously limited.

A convenient option for studying confined liquid flows would be to use a technique that will provide the three velocity components simultaneously but using a PIV set-up. That is, one double frame camera, two pulsed lasers and normal visualization. In this work we will show the advantages of using a high speed version of the Digital Image Plane Holography (DIPH) technique (Palero et al., 2010). In its regular version DIPH has been used for velocity measurements (Lobera et al., 2003; Arroyo et al. 2006), as well as for particle sizing (Palero et al. 2005; Palero et al. 2007). DIPH is a 2D-3C technique that can be extended to be a quasi 3D-3C technique, with the recording of close parallel planes. In its basic configuration, a fluid plane is imaged onto a digital camera and the velocity is obtained from the analysis of the mathematically reconstructed complex amplitude distribution of holograms recorded consecutively. The in-plane components are obtained from the intensity distribution that produces an image similar to regular PIV images that can be analysed alike. The out-of-plane component is extracted from the phase distribution, using an interferometric analysis. However, the time intervals suitable for PIV are 50-100 times larger than the time intervals needed for interferometry. With a conventional PIV system, working in ranges of 10-100 Hz, three holograms are needed, that can be obtained using three pulsed laser, and two cameras, since 10 ms is too long a time for PIV in most applications.

In this paper a High Speed system capable of recording rates over 1000 Hz is proposed for simplifying the implementation of DIPH as a 3C velocimetry technique and thus, broadening its applicability. In particular, DIPH has been used for the measurement of the three dimensional flow inside a complex geometry brain aneurysm model, as time intervals of about 1 ms are suitable for PIV in a wide range of circulatory flows, which makes possible to implement DIPH with only two laser cavities and one camera. However, the high speed laser coherence length is about 10 mm in air, what limits its application to even smaller fluid areas. In order to increase the area under study, a compact and portable system has been designed for artificially increasing the laser coherence length up to several times its original value.

2. Fundamental of the technique

In Digital Image Plane Holography (DIPH), a fluid plane is illuminated with a laser sheet. A lens is used to form its image in a camera sensor (as in PIV) where it interferes with a reference wave, making a so-called image hologram. The use of a collimated reference wave allows an easy
reconstruction of the object wave (Burke et al. 2002). But the isolation of the real image from the virtual image and the dc term is not fully possible for the small angles between the reference and the object waves allowed by the sensor spatial resolution. A divergent reference beam, whose source is located at the same distance from the sensor as the imaging lens aperture, has been used in several works (Lobera et al., 2004, Arroyo et al. 2006). This holographic set-up produces a lensless Fourier transform hologram of the lens aperture and allows an easy isolation of the real image term from the virtual image and dc terms. If \( o = oe^{j\phi_0} \) is the light scattered by the object and \( r = re^{j\phi_0} \) is the reference beam, the hologram intensity can be written as:

\[
I = o^2 + r^2 + e^{j(\phi_0 - \phi_1)} + e^{-j(\phi_0 - \phi_1)} \quad (1)
\]

In the spatial domain these four terms are mixed, but they can be separated in the frequency domain using the appropriated reference beam. Figure 1 shows the Fourier transform of one of these holograms. The heptagons are the real and virtual images of the lens aperture (the third and fourth terms in the previous equation), which are completely separated. The dc term contains \( o^2 \) and \( r^2 \). \( r^2 \) is approximately a delta function and generally appears as a bright spot in the Fourier plane center. Also in the center \( o^2 \) spreads and overlaps slightly with the real and virtual images. The bright triangles near the center come from the interference of the light reflected in the glass that protects the camera sensor.

![Figure 1: Fourier Transform of a DIPH hologram](image)

In the reconstruction process (which has been minutely explained in several works (Lobera et al. 2004)) the real image of the lens aperture is selected, while the rest is blocked. Then, the complex amplitude distribution is propagated to the object plane, were the intensity and phase distributions are reconstructed. The reconstructed intensity distribution is a photography, a PIV image, so the in-plane velocity components can be measured by correlating pairs of them, separated a time interval TPIV. The out of plane component can be extracted from subtracting two phase distributions, separated a time interval \( \Delta T \), using an interferometric analysis. The exact ratio, TPIV/\( \Delta T \), depends on the relative values of the three velocity components and on the magnification; this ratio being smaller for higher magnifications. Figure (2) illustrates the laser pulse configuration for measuring simultaneously the three velocity components.

The pulse sequence shown in figure 2 can be obtained using two high-speed lasers and one high-speed double frame camera. In the pulses nA and nB a double exposure hologram is recorded. The phase difference map (shown at the bottom of the image) is obtained from them. The images in the upper part of the figure are the reconstructed intensity distributions and the 2D velocity vector map is obtained from them. Both lasers should be working at the same frequency, typically larger than 1000 Hz, so the time interval between two consecutive pulses can be adjusted to be suitable for PIV
(TPIV). $\Delta T$ can be adjusted to be suitable for the interferometric analysis.

![Figure 2: Pulse sequence for recording double pulsed holograms suitable for PIV and interferometric analysis.](image)

The PIV analysis for the measurement of the in-plane velocity components is well known and can be found in many works (Raffel et al., 1998). The interferometric analysis for getting the out-of-plane component is also known (Vest, 1979), however a brief reminder is presented in the following.

If two holograms are recorded within a time interval $\Delta T$, the intensity distribution on the sensor plane for each of them can be written as:

\[
I_A = o_A^2 + r^2 + 2o_A r \cos(\phi_{oA} - \phi_r) \\
I_B = o_B^2 + r^2 + 2o_B r \cos(\phi_{oB} - \phi_r)
\]

(2)

where the amplitude and phase of the reference are considered constant. As the object is a particle image field, the phase is a spatially random magnitude. However, the difference $\Delta \phi = \phi_{oB} - \phi_{oA}$ is not and it is related with the local displacement $L$ as:

\[
\Delta \phi = (\vec{k}_o - \vec{k}_r) \vec{L} = \vec{K} \cdot \vec{L} = \vec{K} \cdot \vec{V} \Delta T
\]

(3)

The vector $\vec{K}$ is the so-called sensitivity vector, which is the difference between the observation ($\vec{k}_o$) and the illumination ($\vec{k}_r$) vectors. The illumination vector is a vector in the direction of the propagation of the illumination beam, and is given by $\vec{k}_r = -\frac{2\pi}{\lambda} \hat{i}$ while the observation vector depends on the direction of visualization. Depending on the sensor size and the magnification, the $k_{ox}$ and $k_{oy}$ components can be neglected and the observation vector can be approximated by $\vec{k}_o = \frac{2\pi}{\lambda} \hat{k}$. Then, the phase difference can be expressed as:

\[
\Delta \phi = \frac{2\pi}{\lambda} (L_x + L_z)
\]

(4)

Therefore, the phase difference maps contain information about the in-plane ($L_x$) and the out-of-plane ($L_z$) displacement. Once $L_x$ is known, $L_z$ can be obtained directly.
3. Recording set-up

The recording set-up has been described extensively in Palero et al. (2010). However, a few details are given in the following. A High Speed two-cavity New Wave Pegasus laser ($\lambda = 527$ nm, energy per pulse = 10 mJ at 1000 Hz) was used as a light source. Series of holograms were recorded with a CMOS HighSpeed Star 5 camera (sensor size of 1024x1024 pixels, pixel size of 17 $\mu$m x 17 $\mu$m), at the maximum camera full size recording rate: 3000 images/s. Each laser was fired at half this rate (1500 pulses/s, TPIV=666.7 $\mu$s) with a time interval of $\Delta T= 5$ $\mu$s between the pulses from laser A and B. Let us note that each camera frame records only one pulse. A 105 mm focal length lens has been used to image the illuminated fluid plane into the sensor with a magnification close to unity ($M= 1.06$). The lens was working at its maximum allowed f# which was 32.

![Recording setup diagram](image)

Figure 3. a) Recording set-up. b) Picture of the portable system. c) Brain aneurysm model made of borosilicate glass.

The laser beam is divided in the reference and object beam (figure (3a)). The object beam was shaped into a laser sheet that illuminates a plane in the aneurysm model. The reference beam is split in four references with increasing optical paths using a compact and portable system. This part of the set-up is enclosed in a rectangle in figure (3a). This system is needed due to the short...
laser coherence length ($L_c \sim 10 \text{ mm in air}, Lc/n_i$ in a fluid with refractive index $n_i$) which limits the length of the region that can be recorded. The ‘apparent’ coherence length of the reference beam can be changed by overlapping several reference beams with controlled differences in their optical path lengths. Each beam will interfere with a different part of the object, whose length is the same as the laser coherence length. The details of this part of the experimental set-up are detailed in Palero et al (2010). Figure 3b shows a picture of this system. The arrows indicate the entrance and exits of the reference beam. The optical elements are mounted on individual mounts over a plate of 25cm x 20cm making it compact and completely portable so it can be used with any kind of laser, once $\Delta L$ is adjusted to the laser coherence $L_c$.

The brain aneurysm model is shown in figure (3c). The model was built life-size in borosilicate glass (refractive index $n = 1.4734$). The major dilatation is a balloon of approximately 15 mm in diameter which is placed where all the vessels join. However, only one input vessel (A) and two output vessels are directly connected to the balloon. Although the model size and shape are quite realistic, there are other characteristics that are not, as the flexibility. This model was very useful for testing the performance of the technique in objects with a complex geometry that might be difficult to study with other 3-C velocimetry techniques.

The liquid that fills the model was chosen to match the borosilicate glass refractive index and to have similar characteristics than blood (density and viscosity). The best compromise is achieved with a mixture of benzyl alcohol ($n = 1.54$) and glycerol tri-acetate ($n=1.429$) in a ratio of 40/60. The refractive index of the mixture is exactly 1.4734 and the viscosity is 6.44 cp, slightly higher than the blood viscosity. The liquid was seeded with 4-7 $\mu$m latex particles. This liquid was pumped into the model with a diaphragm pump and a shock absorber was connected to the rubber pipes to remove the vibrations produced by the pump. The flow rate could be controlled changing the voltage that feeds the pump, and was set at 450 ml/min which is a typical value for these systems (Ford et al., 2008). The glass model was immersed in a rectangular glass cell filled with the same liquid than the aneurysm to avoid reflections in the external walls of the model.

The feasibility of the externally modified coherence length laser beam was tested by recording pairs of holograms with each individual reference beam while blocking the others. An aneurysm plane perpendicular to the input vessel A was selected for the tests. The reconstructed complex amplitudes are shown in figure 4. Columns (a) and (b) show the intensity and phase difference distributions respectively. We can divide the hologram in 128x128 pixel windows and calculate the Fourier transforms (figure 4, column (c)). These images give a clear idea of the aneurysm areas where each reference beam is interfering. All the interference areas should have the same length, that is, $Lc/n_i$, independently of the reference. However, it is impossible to divide the hologram in the exact windows that will cover that length. The optical path length difference from one reference beam to the next could be slightly increased in order to cover a longer region. However, if possible, it is preferable to have them overlapped since the actual length of each region can change a little from pulse to pulse and it will also depend on the hologram quality. If the overlapping is not perfect, the discontinuities in the phase map and in the intensity distribution can make impossible its analysis.

Figure 5 shows the intensity and phase distribution when the four reference beams are used together. These images present no discontinuity, showing the good performance of the system. It is also worth mentioning that the phase difference map shows no phase leap indicating that the possible variations in the optical path will affect all the reference beams equally, due to the short times involved in the hologram recording and the use of a compact system.
Figure 4: Reconstructed a) intensity distribution and b) phase map difference, for each individual reference beam. c) Fourier transform map, the presence of the heptagons indicates the areas where the object was interfering with the reference beam

The reconstructed intensity distributions (figure 4a) show not only the aneurysm area that was interfering with the corresponding reference beam as it was expected. In fact, in each one of these pictures there is information coming from the whole object. This can be explained considering that in these holograms only part of the reference beam is interfering with part of the object, so the intensity distribution in the hologram can be written as

\[ I = o^2 + r^2 + \gamma \left[ r o e^{i(\phi_x - \phi_y)} + r o e^{-i(\phi_x - \phi_y)} \right] \] (5)
Here, the interferential terms are multiplied by the coherence factor $\gamma$ that can be approximated as $\gamma = 1$, in the hologram area where the object and the reference beams are interfering and as $\gamma = 0$ in the rest of the hologram. This factor $\gamma$ is in fact a coherence factor that reduces the effective contribution of the light scattered by the object to the interferential terms as compared with the dc term intensity. Thus the part of this dc term that overlaps the aperture real image is strong enough so that some particle images and reflections outside the coherence zone are reconstructed (figure 4a). Although the phase maps have also some extra noise, in practice it not affects the correct analysis of the fringes, and consequently the velocity measurements.

![Figure 5: Reconstructed a) intensity distribution and b) phase map difference obtained with the four reference beams simultaneously. c) Fourier transform map; now there is interference all over the object.](image)

### 4. Processing of the phase difference maps

The phase difference maps shown in figures 4 and 5 are wrapped maps, i.e., each fringe represents a $2\pi$ phase change, but there is not any information about the fringe order. Therefore, these wrapped phase difference maps need some further processing to produce data ready to be used for velocity calculation. The essential part of this processing is to unwrap the phase difference map so that the true phase is known. The unwrapping approach of Qian et al. (2005) with a windowed Fourier transform filtering has been used in this paper with some changes to adapt the method for fluid velocimetry, as follows. It is very convenient to discriminate the speckle noise from the real phase difference measurements, those that come from the particle images. Thus we filter a complex amplitude distribution $a(x,y)$, where the exponential expression of the phase is weighted by the averaged amplitude of both exposures:

$$a(x,y) = \left( \frac{o_a(x,y) + o_b(x,y)}{2} \right) \exp[- j(\varphi_{ab}(x,y) - \varphi_{a1}(x,y))] \quad (6)$$

Qian’s method relies on a local analysis where the frequencies with the biggest contribution in a fixed spectral range are selected. Several parameters characterize the filtering:

- **The window size.** A Gaussian window has been chosen. It was found experimentally that a half width of 10 pixels was the optimum value for our holograms, as the window has to be bigger than the speckle size, but not so much as to loose the advantages of a local filtering.

- **The local threshold for discarding the frequencies with lower contribution.** Although Quian et al. (2005) used a constant value, we used a percentage of the height of the maximum frequency peak that changes locally according to the object intensity. The optimum value changes with the holograms, but in general, we filtered out the frequency peaks smaller than 80% of the maximum.
- The frequency range. In general, the high frequencies are filtered out. As the spectrum includes the frequencies in the range \([-1/(2px):1/(2px)]\), where px is the pixel size, the range chosen is \([-1/(4px):1/(4px)]\).

Figure 6 shows the results of this filtering applied to the complex amplitude of a plane perpendicular to the aneurysm input vessel. Figure 6 (a), shows the intensity distribution, (b) the wrapped phase map, and (c) the unwrapped and filtered phase map (shown wrapped), from where the out-of-plane velocity component is obtained.

![Figure 6: a) Intensity distribution; b) Phase difference map; c) Filtered phase difference map.](image)

**5. Results**

The technique described above was applied for the measurement of the complete 3C velocity field in the plane which intensity and phase difference maps are shown in figure 6. This plane was located in the aneurysm balloon centre between the two output vessels, perpendicular to the input vessel. Therefore the Vz component is expected to be dominant, and the main contributor to the phase difference map.

The in-plane velocity components (Vx and Vy) were obtained correlating the averaged intensity distributions of one pair of holograms with the averaged intensity distribution of the next pair. That is, \(I_n = \sigma_{nx}^2 + \sigma_{ny}^2\) was correlated with \(I_{n+1} = \sigma_{(n+1)x}^2 + \sigma_{(n+1)y}^2\). The main advantage of using these intensity distributions is that the effective particle concentration increases without losing temporal resolution as \(\Delta T\) is much smaller than TPIV.

The image correlation was done with the software DaVis 7.2 from LaVision using a multi-pass processing. In the first step the images were analysed with a window size of 64x64 pixels and an overlapping of 50%. In the second step the window size was chosen to be 32x32 pixels also with a 50% of overlapping. The spurious vectors were eliminated using a median filter and the holes they leave were filled using a linear interpolation.

Once Vx is known, Vz can be calculated from the unwrapped phase difference maps, using equation (3). Considering that the sensor size is 17.4 mm and that the magnification was set close to unity, the error in neglecting the \(k_{ox}\) and \(k_{oy}\) components of the sensitivity vector is smaller than 0.2%. Thus equation (4) can be used for the measurement of Vz.

Figures 7 (a) and (b) show Vx and Vy iso-contours respectively. The Vz iso-contour is shown
overlapped with the in-plane velocity vector map (figure 7 (c)). It shows the presence of a velocity gradient in the area where the liquid enters the aneurysm. In the in-plane velocity vector map only one vector out of two has been plotted in each direction. The flow velocity is low in the balloon and increases up to 3 times in the output vessels.

In the right upper area of these figures there was impossible to obtain any measurement. This is due to the high in-plane displacements which introduce some decorrelation in the particle image interference.

Figure 7: Plane A; a) Vx, b) Vy, c) the in-plane velocity vector map is overlapped to Vz.

The accuracy of Vz can be extracted from the phase maps. Due to the illumination and recording geometry, the sensitivity vector forms an angle of 45° with the OX and OZ axis. Thus, in this case, each interference fringe represents a displacement in the order of $\lambda/\sqrt{2}$ in the sensitivity vector direction. The theoretical technique accuracy 0.01 of a fringe, therefore with a 0.527 µm wavelength, the minimum displacement of 5 nm can be measured. With our recording parameters (magnification, time interval, pixel size) this means that it is possible to measure out-of-plane velocity of 1 mm/s.

Conclusions

The three velocity components in a plane of a complex geometry brain aneurysm model have been simultaneously measured with Digital Image Plane Holography. A high speed system consisting of one camera and two lasers has been used in a configuration similar to PIV. This high speed system not only simplifies the optical set-up but also makes it suitable for its application in a wide range of circulatory flows.

As this high speed system was not originally intended for holography, the laser coherence length was short. A system has been built for artificially enlarging this coherence length up to 8 times the original. This optical system is completely portable and compact and can be used with any kind of laser. The feasibility of the system has been demonstrated.

Acknowledgments

This research was supported by the Spanish Ministry of Education and Science (FIS2006-06058), by the Aragon Regional Government (PI044/08) and by a Marie Curie European Reintegration Grant within the 7th European Community Framework Programme.
References


Prasad AK, Adrian R J (1993) Stereoscopic particle image velocimetry applied to liquid flows. Experiments in Fluids, 15: 49-60


