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Particle image velocimetry (PIV) evaluation of flow modification in aneurysm phantoms using asymmetric stents

Ciprian N. Ionita^b, Y. Hoi^e, H. Meng^e, and S. Rudin^{a,b,c,d}

^a Department of Physiology and Biophysics, Toshiba Stroke Research Center, University at Buffalo (SUNY), 3435 Main Street, Buffalo, NY 14214

^b Department of Physics, Toshiba Stroke Research Center, University at Buffalo (SUNY), 3435 Main Street, Buffalo, NY 14214

^c Department of Neurosurgery, Toshiba Stroke Research Center, University at Buffalo (SUNY), 3435 Main Street, Buffalo, NY 14214

^d Department of Radiology, Toshiba Stroke Research Center, University at Buffalo (SUNY), 3435 Main Street, Buffalo, NY 14214

^e Department of Mechanical & Aerospace Engineering, Toshiba Stroke Research Center, University at Buffalo (SUNY), 3435 Main Street, Buffalo, NY 14214

Abstract

Asymmetric stents are promising new devices for endovascular treatment of cerebrovascular aneurysms. For in vitro experiment a patch made of stainless steel mesh is directly attached onto a standard stent and deployed so that the patch is placed over the aneurysm orifice. Thus we modify substantially the flow into the aneurysm and decrease the shear stress on the aneurysm walls. We used mesh-patches having different permeabilities and evaluated the flow using Particle Image Velocimetry. PIV provides instantaneous velocity vector measurements in a cross-section of flow containing reflective micro-particles. A pulsed-laser light sheet illuminates the flow in the target area and images are acquired using a CCD camera. By registering the position of the particles in two successive images the fluid velocity vectors components are calculated. From the 2D velocity field a best polynomial fit is made to obtain a smooth function of each velocity with respect to the coordinates. Using the fit, we derived the values of quantities of interest in the plane of acquisition such as: tangent shear stress, vorticity and inflow. We used four meshes of different permeabilities. We found out that by using lower permeability meshes we create better conditions for the embolization of the aneurysm.

Keywords

asymmetric stent; particle image velocimetry (PIV); cerebral aneurysm; shear stress

1. Introduction

Intracranial aneurysms are deformations of the artery walls that occur mostly in regions subject to high mechanical factors like pressure and shear stress. These regions are located frequently in curved vessels or bifurcations (symmetric or asymmetric) [1], [2]. The most common methods for the treatment of intracranial aneurysms are clipping and coil embolization [3]. Clipping of the aneurysm is surgically invasive method. It consists of isolating the aneurysm dome by placing a metallic clip on the aneurysm neck before the removal of the aneurysm dome. However, most of the intracranial aneurysms occur in remote regions around the Circle of Willis, which makes clipping a very difficult and sometimes impossible task. Gugliemi in 1989 developed a new endovascular method [1], [3], which is much less invasive than clipping. Under x-ray guidance, the interventionalist, using microcatheters, packs the aneurysm dome with platinum coils in order to limit the flow and provide a scaffold for the thrombosis. This method is currently the most used method for the treatment of cerebral aneurysms. The risks associated with this method include perforation of the dome by the coils and protrusion of the coils into the main vessel. There is also a risk of aneurysm regrowth after coil embolization which is between 14% and 17 % for small aneurysm (diameter less than 5 mm), and 50% for large aneurysms [3].

Lately two new methods of treatment have been proposed and are currently under investigation. One employs fibrin gels for the treatment and the other uses stents. The first one consists of inserting fibrin gels in the aneurysm dome in order to disrupt flow and accelerate thrombosis [4]. The second method involves stenting of the vessel in the region of the aneurysm. By doing this the interventionalist intends to significantly modify the flow within the aneurysm, and create conditions for thrombosis. Conditions for thrombosis presume among other factors two principal conditions: one is to have activated platelets and the second is to have flow stasis in the aneurysm for clot formation. In [5] S. M. Slack et al. suggest that activation of platelets take place in areas with high shear rates. In the treatment with stents of the aneurysm in the area of the neck the wires of the stent produces sudden changes in blood flow velocity profiles which result in high shear rates thus stimulating activation of the platelets. The stent also reduces the inflow to 10% to 50% in comparison with the untreated case, depending on the type of stents used. Different groups have studied the usage of stents for treatment of intracranial aneurysm [6], [7], [8], [9], [14].

There is a direct correlation between the type of stent used and the flow modification in the aneurysm. B Lieber et al [7] studied the effects that parameters such as porosity of the stent or wire thickness have on the flow. This study shows that changes in parameters such as porosity or stent wire dimensions have a large impact on the flow. Building a stent that will totally change the flow stimulating thrombogenesis is not an easy task. There have been attempts to develop a neuro stent that will solve this problem. The difficulty is due to the location of such aneurysms; as we mentioned most of them occur in the Circle of Willis, such a stent has to fulfill two very important requirements. First, the stent has to be sufficiently porous to limit the probability of blockage of perforators, and secondly the stent has to be less porous at the region of the aneurysm neck in order to induce significant changes in the flow. In a study of probability of blockage of perforators by stents C. Y.

Yang et al [6] showed that when using commercially available stents which have porosity around 80% the probability of blocking a 100 μm perforator is around 20%.

We have designed a stent that fulfills these two requirements, the prototype is depicted in fig 1; we took a Tri-Star stent provided by (Guidant Corp Temecula CA) and we modified it by welding a stainless steel patch as shown in Fig. 1(a). We also considered the possibility of using a porous graft made of PTFE (polytetra-fluoro-ethylene) which is sutured on the stent Fig. 1(b). These stents are designed for in vitro experimental purposes only, the final asymmetric stent will have a much more porous structure which will satisfy the second requirement we stated above concerning fabrication of a neuro stent.

This new asymmetric stent will require a high resolution detector in order to place it accurately not only in the transversal direction, along the artery, but also azimuthally, with respect to the center of the artery. Using a micro-angiographic detector developed by our group [11] [12], we were able to localize the asymmetry on the undeployed stent with a precision of 2 degrees [13], a precision which should be safe enough for the placement of such a device.

In order to study the effects on the flow of the new device we made a Particle Image Velocimetry (PIV) experiment in which we used different stents in the same flow conditions. The PIV method consists of tracking packets of fluorescent particles which are mixed with the carrier fluid. Using this method one can evaluate accurately the velocities in the area of interest. The main impediment is that in one PIV acquisition, velocities can be found in only one plane. Most of groups have done their studies in side wall aneurysms in straight vessels using different stents available commercially.

Using the PIV data additional quantities can be evaluated: vorticity [8], or shear stress [14]. In [14] S.C.M. Yu used a 3 degree polynomial fit of the velocity profile in a region next to the wall in order to evaluate the shear stress. Following the same idea we propose a technique which uses a function that fits one velocity component within the whole aneurysm. The advantage of this method is that it gives greater flexibility in working with velocity components, and additionally, we can use the same functions to evaluate quantities such as vorticity or the inflow in the aneurysm in a certain plane.

In this presentation we intend to study the effect that different permeabilities have on the flow modification within the aneurysm. We will assess these modifications by comparing the vectorial fields, the shear stress at the wall of the aneurysm, vorticity and the inflow and outflow at the aneurysm neck. Such a study will prove what range of permeabilities we need to use in order for the stent to be effective.

2. Materials and Methods

2.1. Fabrication of phantoms

There have been different approaches for simulating aneurysms in PIV experiments for flow modification using stents. In general phantoms of a straight vessel with side wall aneurysms made from a translucent elastomer [8] or transparent glass [9] and [14] have been used. Elastomer phantoms with side wall aneurysms placed on curved vessels were also studied

[15]. In order to mimic more accurately the cerebral arterial geometry and model aneurysmal blood flow dynamics we used the same geometry as reference [15]. The phantoms we have used are made from a translucent silicone elastomere (Sylgard 184 silicone elastomer, Dow Corning Corporation, MIDLAND MI). The geometry of the phantom is presented in the Fig. 2. s. First we made a mold from cerro (Mac Master Co., Cleveland, OH), which is a special metal alloy with a low melting point around 80°C. We melt the metal and then poured it in a metal cast. When the model hardens it is removed from the metal cast and introduced into a plastic box. The box is filled with silicone elastomere and placed in a vacuum chamber in order to remove any air bubbles. After all the air has been removed the phantom is pulled out and left for 48 hours to cure. We remove the metal by heating the model in a container with hot water. After the metal has been removed we use a solution of nitric acid to remove any small residues of the metal from the phantom.

2.2. Fabrication of the asymmetric stents

The asymmetric stents for this study are made by modifying commercially available coronary stents, (Penta by Guidant Corp, Temecula CA). We modified them by laser micro-welding stainless steel patches with different permeabilities using a Micro Laser Welder or by suturing a PTFE porous graft onto them. The descriptions of the meshes are given in the Table 1. The mesh count stands for the number of wires per inch. The patches are cut using the micro welding laser; in this way we ensure formation of “puddles” of melted metal on the edge that will help in welding them onto the stent. This method ensures that there will be a relatively smooth edge that will not puncture the balloon on which the stent will be crimped or scratch the phantom walls. The porosity was calculated as the ratio between the open area and the total area. Permeability was calculated using a simple approximation equation (1) given in [16]:

$$Permeability = \frac{Porosity * (area\ pore)}{32} \quad (1)$$

Standard materials including PTFE were used; in a review of the current types of grafts used in construction of covered stent S. E. Greenwald [10] presents some of the proprieties of such materials. PTFE has a good biocompatibility as well as a good compliance with the vessel wall which will reduce the possibility of appearance of intimal hyperplasia.

2.3. Particle Image velocimetry setup

We used steady flow, so that the flow was fully developed in the main vessel right before the aneurysm. The average flow rate was 414 ml/min. The fluorescent particles used had the diameter less than 20 µm; we used micro-particles from Microparticle GmbH, Dantec-Denmark. Due to the dimensions of our phantoms that are much larger than the actual clinical cases we used a solution with a higher viscosity than blood in order to mach the Reynolds number. The solution was 60% glycerin and 40% water. The viscosity of the working fluid was 10.8 centipoises and the Reynolds’s number was 136.

The experimental setup of PIV experiment is presented in Fig. 3. An Argon laser emits a continuous beam which passes through a Bragg Cell. In order to obtain a pulsed laser sheet

we place cylindrical lens collimator (CLC) in the beam. The emergent light sheet was incident on an aneurysm phantom. A CCD 1KX1K camera records the position of the same reflective particles at two different times. In front of the camera we placed a 580 nm filter to increase the signal to noise ratio. The data are recorded by a computer, integrated over many particle observations and the vector components of the velocities in the exposed plane are calculated at each point.

The vectorial field was acquired in a symmetry plan for all the phantoms including a non-stented phantom. The velocity components were calculated using a 60×60 grid with a spatial resolution of 0.3 mm. The program used for the acquisition and calculations was provided by IDT SuperVision, Tallahassee, FL. We acquired 120 frames for each acquisition; exposure time was 280 μs for the acquisition in the aneurysm plane. We took for each case three runs and averaged them.

2.4. Data analysis

From the PIV experiments we obtain a vectorial field. Associated with the vectorial field we obtain a data file which contains the (x, y) position measured in pixels and the two velocity components. We use a known distance in the image, such as vessel diameter, and calibrate the data in order to obtain the position in millimeters and velocities in mm/s.

We use these data not only to plot the vectorial fields but also to derive some quantities of interest which we can compare among the cases we studied. We calculated a function that fit one velocity component within the whole aneurysm in the plane in which the data are acquired. The advantage of this method is that it gives us greater flexibility in working with velocity components and we can use the same function to evaluate quantities such as vorticity and the inflow in the aneurysm in a particular plane. The best fit was obtained using a cosine bivariate series given by:

$$P(x, y) = \sum_{i=0}^{10} A(i) S_i(x') \sum_{j=0}^{10-i} B(j) S_j(y') \quad (2)$$

The elements of higher order are defined as:

$$S_i(x') = \cos[ix'] \quad (3)$$

because the cosine series are defined only on the interval $[0, \pi]$ we have to map our physical domain using the next scaling formula:

$$x' = \frac{x - x_{\min}}{x_{\max} - x_{\min}} \cdot \pi \quad (4)$$

The fit is done by expanding fully the expression (2). We got 67 terms which have the coefficients given by products of the form $A(i)B(j)$. The accuracy of the fit is good for our purpose; we obtained an R^2 of .83 for the untreated case and an average of .93 for the treated cases. Once we found the velocities in Cartesian coordinates we changed our coordinate

system to a polar system with the origin in the center of the aneurysm, and we recovered the velocities components with regard to the new system:

$$\begin{cases} u=P_1(x, y) \\ v=P_2(x, y) \end{cases} \iff \begin{cases} v_\varphi=P_1(\varphi, r) \\ v_r=P_2(\varphi, r) \end{cases} \quad (5)$$

The conventions that we use through this paper will be that the x component of the velocity is denoted with **u** and the y component is noted by **v**. using the new velocity components we derived the shear stress tangential to the wall of the aneurysm within the plane in which we performed the acquisition according to the formula:

$$\sigma_{\varphi r}=\mu \left(\frac{\partial v_\varphi}{\partial r} + \frac{1}{r \sin \theta} \frac{\partial v_r}{\partial \varphi} - \frac{v_\varphi}{r} \right) \quad (6)$$

where θ is the azimuthal angle which for our case is 90° (z axis is perpendicular on the plane of acquisition) and φ is the polar angle which is measured as indicated in Fig. 2.

Another quantity of interest we can calculate using the fit we derived is vorticity. By evaluating equation (7) we can estimate the flow conditions within the aneurysm. Ideally we would like to achieve very low vorticity areas in order to ensure ideal conditions for stasis and hence induce thrombosis:

$$vorticity = \left(\frac{\partial v}{\partial x} - \frac{\partial u}{\partial y} \right) \cdot \hat{k} \quad (7)$$

The final part of the analysis is to evaluate the inflow in the aneurysm relative to the untreated case. In figure 2 we consider the curvature between distal and proximal points, the arc (PD), and we intend to calculate the flux passing through. Considering the fact that the inflow through a surface is given by the scalar product between the velocity vector field and the surface vector we can evaluate the flow rate using the flowing expression:

$$Q = \vec{S} \cdot (v \cdot \hat{i} + u \cdot \hat{j}) = S \cdot \vec{n} \cdot (v \cdot \hat{i} + u \cdot \hat{j}) = d \cdot l \cdot v_{normal} \quad (8)$$

In the above equation v and u are the x and y component of the velocity and S is the surface of the mesh through which the fluid passes, n is the normal to S, l the length of the arc (PD) in fig 2 and d is the thickness of the plane in which we acquired the data. Due to the fact that for the laser sheet incident onto the phantom we used the same collimation the thickness d is the same for all the studies, and for convenience we will consider to be equal with one. The particle velocity components are given by the fit we found while the normal is found by evaluating the geometry in figure 2.

According to Darcy's law [15]:

$$Q = - \frac{K}{\mu} \frac{\Delta P}{\Delta L} \quad (9)$$

Q is the flow rate through a porous medium is proportional with pressure drop P across that permeable medium. L represents the thickness of the permeable medium, μ is viscosity and K is permeability. For the same pressure drop and same viscosity we will expect that the flow will decrease linearly with the ratio (permeability of the mesh)/(thickness of mesh) considering that the flow is fully developed at the aneurysm neck.

3. Results

The PIV results are presented in Fig. 4. The legend on top reflects only one component of the velocity namely the u component. The scaling is done by tacking the interval between the maximum and the minimum velocity and dividing it into ten smaller intervals, for each interval we assign a different color. The experimental data is plotted using TecPlot, (TecPlot Inc Bellevue, WA), which is a program associated with the acquisition software. In the first row, right column we present the vectorial field obtained after we fitted the data, to emphasize the difference between experimental results and the fitted ones we represented the plots using a white background. The untreated aneurysm contains the entire dome while the treated ones have the area of the neck excluded due to the presence of the asymmetric stent.

A more accurate evaluation of the changes produced by the placement of different asymmetric stents can be done by making the histogram of the absolute value of the velocity within the aneurysm for the cases we studied. The results are presented in Fig. 5. On the x axis we plotted the magnitude of the velocity using a bin interval of 0.05 mm/s. The velocity magnitude was found as the square root of the sum of square the velocity components. We plotted the histogram of velocities for the untreated aneurysm in the range 0 to 4 mm/s using a bin interval of 0.05 mm/s for comparison with the treated cases. The upper line in the treated aneurysm cases represents the maximum of the first peak (400×400 case), which corresponds to small velocity contributions. The lower line in the same figure corresponds to the maximum in the second peak (PTFE case), associated with contributions from the higher velocities.

Using formula (6) and fit (1) we calculated the shear stress tangent to the aneurysm wall in the plane where we acquired the data. The results are presented in Fig. 6. The angle on the x axis represents the orientation with regard to the x axis indicated in fig 2. For our case we consider that the polar axis is oriented perpendicular on the plane in which we acquired the data, we also consider the origin to be contained in the plane of acquisition where $\theta = 90^\circ$, so $\sin(\theta)$ is equal with one. The azimuthal angle ϕ is measured according to Fig. 2, zero is considered to be at the dome and 180° around the neck of the aneurysm. For comparison purposes in the same picture we plotted the shear stress for the PTFE case. In the plot Fig. 6(b) we represent the values for all four stents. The flat region corresponds to the neck of the aneurysm, between 118 and 187 degrees, the shear stress has been considered to be zero due to the fact that in the central plane in which we acquired the data the stent is covering the neck protecting it from exposure to the impinging jet from the main vessel.

In Fig. 7 we present a field in which we calculated the vorticity by evaluating the contour plots. To quantify the vorticity magnitude we used a gray scale color code. White denotes

the maximum value and black is the minimum. It is difficult to quantify by how much the vorticity has decreased when we compare the stented cases. We can do an accurate evaluation of the plot contour by building a histogram such that on the x axis we have the vorticity and on the y axis the number of pixels in the contour plot having that particular vorticity. On the upper row of Fig. 9 we represented the histogram of the untreated case; in the lower row we represented the histograms for the treated cases. The horizontal line represents the maximum number of pixels with low vorticity for the most porous asymmetric stent (PTFE covered stent). The bin numbers of the histogram were 0.05 s^{-1} for the treated cases, and 1 s^{-1} for the untreated case.

A last quantity to evaluate is the inflow. Using equation (8) we can derive the distribution of the velocity normal to the curved surface of the parent vessel. The results of this analysis are presented in Fig. 10–12. In Fig. 10 we plotted the normal components at the curve in the plane in which we acquired data. Fig. 11 and Fig. 12 provide a standard plot of the same data. By integrating these profiles we can determine the inflow and the outflow.

We integrated over the flows in Fig. 11 and Fig. 12, then we divided by the arc length over which we performed the integration in order to find the inflow per unit arc length in this particular plane. In the same way we calculated the outflow. We also calculated the difference between the total inflow and outflow. The results are presented in Table 2. In Fig. 13 we represented the inflow versus the ratio permeability of the mesh/thickness of the mesh and we fitted the data with linear equation.

4. Discussions

Analyzing the vectorial fields in Fig. 4 we can see the difference that the usage of the asymmetric stents produces, even for the most permeable mesh the drop in velocity is about two orders of magnitude. In Fig. 4(a) and (b) for the untreated aneurysm the inflow is directed toward the distal wall causing a large vortex within the dome. In this region the wall is exposed to high mechanical factors which could be responsible for rupture and subsequent hemorrhage. The next results, presented in Fig. 4(c), (d), (e) and (f) are for the treated aneurysm using different asymmetric stents. As can be seen from the pictures all the meshes had a radical effect in changing the flow pattern as well as the magnitude of the velocities. Additionally the vortex than can be seen in the untreated case is totally removed and it is replaced by two smaller vortices. The first vortex is situated right next to the aneurysm neck and the second is located toward the dome. This pattern is maintained in general for all the stented cases but it loses its strength compared with the most permeable stent, namely PTFE covered stent Fig. 4(c). In the 400×400 case, Fig. 4(f), the second vortex is so weak that it can not be even identified incompletely in the vectorial field acquired. These results suggest that the exchange of momentum is very small between the two vortex structures such that the second virtually vanishes. In all the stented cases next to the wall we have very slow motion of the fluid. These areas which are totally decoupled from the main flow are areas in which we appear to have flow stasis. Flow stasis is the main goal we want to achieve, in such areas we have a great probability that stable thrombus will form.

The results are confirmed by the contour plots of the vorticity Fig. 7. In the first plot of the untreated aneurysm we can see the main vortex rotating clockwise (negative values are dominating). Next than we have the treated cases where the two main vortices can be visualized very clearly: the main vortex rotating counter clockwise located at the entrance (white color) and the secondary vortex within the dome. Note that the magnitude has also decreased by two orders of magnitude.

The vectorial field analysis is not sufficient to make a comparison between the treated aneurysm cases; the maximum velocity has not changed dramatically from the most permeable mesh to the least one. We can explain this behavior on the bases that the total flow according to Darcy's law, equation (9) is directly proportional with the permeability of the media but is also inversely proportional with the thickness of the mesh. The thickness of the mesh is twice the diameter of the wire, thus by going to finer meshes the ratio permeability/thickness is not changing as dramatically. As can be seen from vectorial field the maximum velocity didn't change by much, but we need to know how many particles are moving with this velocity when we change among the different permeabilities. To verify this aspect we refer to histogram from Fig. 5. For the untreated case we took for the absolute velocity only the values contain in the interval 0–4 mm/s which we used for the treated cases. As one can see the count of particles within this region is not higher than 15. On the lower row we plotted the four treated cases. In the histogram we have two important regions one corresponding to high velocities and one for small ones. The higher velocity contribution is given by the particles moving in the vortex at the aneurysm neck, while the lower velocities correspond to the rest of the aneurysm. We will analyze how the distribution changes in the two areas. As the permeability decreased from 114 to 12 μm^2 the first part of the distribution increased from 90 counts for the PTFE mesh to more than 120 for the 400×400 mesh. The range of values for the velocity remained about the same around 3.5 with fluctuation of less than 3%. An important aspect is the behavior of the second peak of the distribution. As permeability decreases the peak decreases indicating a drop in the number of the particles moving with high velocities

In Fig. 6 we represented the evaluated tangential component of the shear stress in the plane of acquisition. The tangential shear stress dropped by two orders of magnitude when we compare the untreated case with a treated one. For the untreated case the maximum values that we calculated are about 12 dynes/cm² at a Reynolds number of Re=136 and are located as the distal wall and at the aneurysm neck. It can be seen the shear stress decreases with the decrease of the permeability at the distal wall. The 400×400 mesh has even a more dramatic change making the shear stress change its sign at the distal part of the aneurysm, this is due to the fact that the mesh has changed the direction of the vortex within the aneurysm. Regarding the rest of the aneurysm there are small oscillations of the shear stress which are due apparently to the jets perpendicular to the plane in which we acquired the data.

By analyzing vorticity we can get a measure of the tangential forces that act within the plane in which we acquired the data. Vorticity is a direct indication of the degree of agitation and of the mechanical stress that within the aneurysm dome. The contours plots in Fig. 7 represents a map of the vorticity. From the histogram in Fig. 8 we can see that for the untreated cases here are high negative values which go down to -75 s^{-1} . The lower row of

the same figure represents the treated cases. It can be seen that the areas with low vorticity practically doubled for the 400×400 mesh when compared to the PTFE case.

Final part of the analysis is concerned with the study of the inflow and outflow at the aneurysm neck. We study the inflow only in one plane. First we calculated the normal velocity to the parent vessel. We represented the calculated values in Fig. 10. There is a large difference between the untreated case and the stented cases. There is a narrow inflow region for the untreated case at the distal point and a large outflow which spreads over all the aneurysm neck. In the stented cases there is an extended inflow and a reduced outflow closely placed, they correspond to the small vortex that we can see in figure 4 for all the treated cases. When we use stents the area of inflow is extending and produces a redistribution of the mechanical factors over a larger region. In Fig. 11 and Fig. 12 we plot the inflow as a function of position on the arc PD. By integrating these profiles we can find the total inflow and the outflow. In Table 2 we have represented the actual inflow and outflow per unit of arc length of the aneurysm neck. In general the inflow per unit length decreases as the permeability decreases. However the outflow does not follow the same rule for all the meshes. On the fourth column we represent the balance between the total inflow and total outflow. For the untreated case there is a huge difference between the two quantities. This result indicates that there is a larger inflow in another plane other than the one in which we acquired the data. A more complete analysis of the 3D flow may be done by performing a 3D PIV study.

In Fig. 13 we represented a plot of inflow per unit length of the aneurysm length as a function of ratio permeability/thickness of the mesh. We can see that the points are following almost a linear behavior. The deviation from the linear behavior is due to the fact that the mass is not totally conserved within the plane in order for the Darcy equation to hold true. However the results show that in our case for a permeability decrease of 100 μm the inflow per unit of arc length drops with 73%.

5. Conclusions

We have applied a new method to derive quantities of interest from PIV data. Even permeabilities as high as 114 μm^2 proved out to be very effective in reducing the stress on the aneurysm wall, however they did not succeed in lowering the vorticity in the aneurysm dome a condition essential for creating thrombosis. The 400×400 mesh which has a permeability of 12 μm^2 has successfully fulfilled both requirements. Our results show that asymmetric stents can be effective in aneurysm treatment. However, more extensive studies are necessary for a more complete understanding of the processes associated with flow modification within the aneurysm.

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Fig. 1.
Asymmetric Stents

Fig. 2. Aneurysm Model (All the units are in mm, angle ϕ is measured with respect to the x axis in degrees, point D is the distal side of the aneurysm neck and P will denote the proximal side, D_v -diameter of the vessel, R_1 - radius of the aneurysm, R_2 - Radius of the curvature)

Fig. 3.
The experimental setup for PIV acquisition

Fig. 4. PIV results. Vectorial fields for untreated and treated aneurysms. The positioning and orientation is the same as that presented in Fig. 2. In (a) we plotted the vectorial field obtained from the experimental data for the untreated case, and in (b) for comparison purposes we plotted the vectorial field obtained using the fitted profiles. The arrows indicate the inflow and the outflow direction of the fluid. The other plots are only the experimental results for the stented cases: (c) PTFE Asymmetric Stent, (d) 250×250 Asymmetric Stent, (e) 200×200 Asymmetric Stent, (f) 400×400 Asymmetric Stent

Fig. 5. Histogram of absolute velocity. We plotted the velocities on the x axis and the number of particles having that particular velocity on the y axis.

Fig. 6.
Shear Stress evaluation at the wall of the aneurysm within the plane of acquired data

Fig. 7.
Contour plots of the vorticity in the plane in which the data were acquired. For the untreated case the gray scale is between $(-100, 100)\text{s}^{-1}$ while for the treated cases is between $(-1, 1)\text{s}^{-1}$.

Fig. 8.
Histogram for the evaluated vortices within the aneurysm dome for the untreated case

Fig. 9.
Histogram for the evaluated vortices within the aneurysm dome for the treated case

Fig. 10. The radial components of the velocities at the aneurysm neck for different cases (for aneurysm geometry see figure 2). The axis represents the actual position in mm with regard to the center of the aneurysm.

Fig. 11.
Comparison of the radial velocity component for the untreated and a treated aneurysm.

Fig. 12.
Comparison of the radial velocity component for the treated aneurysms.

Fig. 13.
Inflow dependence on the ratio K/l

Table 1

Parameters of the meshes used for creating the asymmetry

Mesh count	Wire diameter(μm)	Width opening(μm)	Porosity	Permeability (μm^2)
PTFE	78	106	0.34	114
200×200	53	74	0.33	57
250×250	41	61	0.36	41
400×400	28	36	0.31	12

Table 2

Permeability K(μm^2)	Inflow/(unit arc length)	Outflow/(unit arc length)	Total flow	K/l
Untreated	17.60	23.21	134.33	-
120	1.17	2.28	0.57	0.74
57	0.73	0.77	1.24	0.53
41	0.65	0.96	0.21	0.51
12	0.32	0.30	-0.85	0.22