Hemodynamics, and in particular Wall Shear Stress (WSS), is thought to play a critical role in the progression and rupture of intracranial aneurysms. Wall motion is related to local biomechanical properties of the aneurysm, which in turn are associated with the amount of damage undergone by the tissue. The underlying hypothesis in this work is that injured regions show differential motion with respect to normal ones, allowing a connection between local wall biomechanics and a potential mechanism of wall injury such as elevated WSS. In a previous work, a novel method was presented combining wall motion estimation using image registration techniques with Computational Fluid Dynamics (CFD) simulations in order to provide realistic intra-aneurysmal flow patterns. It was shown that, when compared to compliant vessels, rigid models tend to overestimate WSS and produce smaller areas of elevated WSS and force concentration, being the observed differences related to the magnitude of the displacements. This work aims to further study the relationships between wall motion, flow patterns and risk of rupture in aneurysms. To this end, four studies containing both 3DRA and DSA studies were analyzed, and an improved version of the method developed previously was applied to cases showing wall motion. A quantification and analysis of the displacement fields and their relationships to flow patterns are presented. This relationship may play an important role in understanding interaction mechanisms between hemodynamics, wall biomechanics, and the effect on aneurysm evolution mechanisms.

Keywords: Image Registration, Wall Motion Estimation, Computational Fluid Dynamics, Aneurysm Rupture, Compliant Models

1. INTRODUCTION

Intracranial aneurysms are pathological dilatations of cerebral arteries, which tend to occur at or near arterial bifurcations, mostly in the circle of Willis. The optimal management of unruptured aneurysms is controversial and current decision-making is mainly based on considering their size and location, as derived from the International Study of Unruptured Intracranial Aneurysms (ISUIA). Although there is little doubt that arterial and aneurysmal walls do move under physiologic pulsatile flow conditions, there is no accurate information on the magnitude and other motion characteristics required to understand the interaction between hemodynamics and wall biomechanics. Although the visualization of aneurysmal pulsation seems to be possible with the advent of 4DCTA gated imaging techniques, there are a number of imaging artifacts related to motion of bony structures which bring questions about the accuracy and reliability of this method. It is thought that the interaction between hemodynamics and wall mechanics plays a critical role in the formation, growth and rupture of aneurysms. However, since there are no reliable techniques for measuring flow patterns in vivo, various modeling approaches were considered in the past. Hitherto, most CFD methods assume rigid walls due to a lack of information regarding both wall elasticity and thickness. Moreover, in order to perform simulations that account for the fluid-structure interaction, it is also necessary to prescribe the intra-arterial pressure waveform, which is not normally acquired during routine clinical exams. In this paper, wall motion is quantified by applying image registration techniques to sequences of Digital Subtraction Angiography (DSA). To study the effects of wall motion...
compliance on the aneurysmal hemodynamics, the obtained wall motion is directly imposed to the 3D model derived from the medical images. This allows to study the patient’s hemodynamics using personalized models including geometry and boundary conditions closer to reality than previous rigid models. In the next section, the methodology employed for wall motion estimation and its inclusion in CFD simulations is explained. Section 3 presents the results obtained for the selected patients from the database. Finally, in Section 4 advantages and drawbacks of the methodology are discussed as well as future research lines.

2. MATERIALS AND METHOD

2.1. Dataset and image acquisition

Table 1 presents an anatomical description of each patient used for the experiments, and information on image acquisition parameters. Each patient considered in this work underwent conventional transfemoral catheterization and cerebral angiography using a Philips Integris Biplane angiography unit. As part of this examination, a rotational acquisition was performed using a six-second constant injection of contrast agent and a 180-degree rotation at 15 frames per second over 8 seconds. These images were transferred to the Philips Integris Workstation and reconstructed into 3D voxel data using the standard proprietary software, which was used for generating a 3D anatomical model. Biplanar dynamic angiogram at 7.5 Hz / 2.0 Hz was subsequently performed using a six-second contrast injection for a period of at least 6 seconds. In addition, an expert neuroradiologist (CP) measured the diameters D1 and D2 (maximum height and width, respectively) on these projection views. This information was then used to establish the pixel size and to quantify wall motion.

<table>
<thead>
<tr>
<th>Patient</th>
<th>Frame rate (Hz)</th>
<th>Aneurysms</th>
<th>Location</th>
<th>Type</th>
<th>Motion</th>
</tr>
</thead>
<tbody>
<tr>
<td>#1</td>
<td>7.5</td>
<td>1</td>
<td>ICA</td>
<td>Lateral</td>
<td>Pulsation</td>
</tr>
<tr>
<td>#2</td>
<td>2.0</td>
<td>1</td>
<td>ICA</td>
<td>Bifurcation</td>
<td>Pulsation (dome and bleb)</td>
</tr>
<tr>
<td>#3</td>
<td>2.0</td>
<td>1</td>
<td>ICA</td>
<td>Bifurcation</td>
<td>Pulsation</td>
</tr>
<tr>
<td>#4</td>
<td>7.5</td>
<td>1</td>
<td>Basilar</td>
<td>Terminal</td>
<td>Change in artery angle</td>
</tr>
</tbody>
</table>

2.2. Wall motion estimation

In order to estimate the wall motion, image registration, which establishes correspondences between points in two different images (so called fixed and moving), was applied to the series of 2D images. To this end, an implementation of Thirion’s demons algorithm was applied. This method belongs to the broader class of optical flow methods, suitable for intramodality registration and small deformations which is precisely the scenario in this work. During the contrast injection the image intensity changes, reflecting the contrast agent distribution inside the vessels. This constitutes a problem for optical flow methods since they assume constant intensity of moving points, and if this change in intensity is not compensated by some means, there would be a potential source of errors in the motion estimation. To cope with this problem, the moving image intensity was modified to match the fixed image intensity by using histogram matching methods. The image registration method differs from that used in our previous work because it was found that optical flow methods showed to be more sensitive than techniques based in information theory for retrieving small deformations in this type of images. This could be explained by the local matching properties of optical flow methods that make them more robust to poor image quality and late appearance of arteries than global similarity metrics. For example, the difference caused by an artery present in the moving image and absent in the fixed one could have a much stronger effect in the global metric than a small artery deformation in another part of the image, and the optimizer could choose to compensate the difference caused by the artery instead of the wall deformation.

For each series of DSA images, a set of landmarks were manually delineated in the first frame, and subsequently propagated by using the transformations derived from the image registration procedure. The complete series was registered to the initial reference frame. Thus, wall motion was estimated from the propagated landmarks
by calculating the distance between corresponding points. A distribution of displacement vectors was obtained for the complete set of landmarks. The wall motion was then analyzed by grouping landmarks according to their location in the aneurysm. For example, for sequence #1, landmarks were grouped into a) artery, b) bleb and c) dome. In general, the landmark motion is contaminated by artifacts caused mainly by arteries absent in the first frame but appearing later as a consequence of the contrast injection. This phenomenon makes the boundary of the vessel difficult to distinguish and the local motion estimation fails. Therefore a landmark selection is needed, which was done by visual inspection of the motion curves and keeping only the curves showing cardiac periodicity.

Once the useful landmarks are selected it is necessary to remove the remaining low-frequency components caused by changes in image intensity not compensated by the histogram matching. These components are easy to identify by looking at the Discrete Fourier Transform (DFT) of the landmark motion, and were removed by using a simple high pass Butterworth filter with cutoff frequency at 0.3 Hz. Figure 1 shows an example of wall motion curve obtained from DSA sequences before and after filtering. The original signal presents a periodic behavior with a continuous increase in the mean value, and can be considered to be the sum of a zero mean periodic signal and another one of very low frequency. The low frequency component could be explained as caused by the contrast injection that produces a continuous increase in the intensity interpreted (incorrectly) as motion by the optical flow. This signal description becomes also evident from the power spectrum plot shown in Figure 1 (b): the peak at the lowest frequency corresponds to the change in mean, the one in the middle is centered at the frequency of the periodic signal and the last one explains changes in movement faster than the heart rate. Figures 1 (c) and (d) show the resulting wall motion curve after the filtering process along with its corresponding power spectrum without the peak at low frequency.
2.3. Flow simulations

In order to compute intra-aneurysmal flow patterns, personalized models of blood vessels were constructed from 3D rotational angiography (3DRA) images. The segmentation procedure was based upon the use of deformable models by first smoothing the image through a combination of blurring and sharpening operations, followed by a region growing segmentation and isosurface extraction. This surface was then used to initialize a deformable model under the action of internal smoothing forces and external forces from the gradients of the original unprocessed images.\(^6\) The anatomical model was subsequently used as a support surface to generate a finite element grid with an advancing front method that first re-triangulates the surface and then marches into the domain generating tetrahedral elements.\(^9\) The blood flow was mathematically modeled by the unsteady Navier-Stokes equations for an incompressible fluid:

\[
\nabla \cdot v = 0, \rho \left( \frac{\partial v}{\partial t} + v \cdot \nabla v \right) = -\nabla p + \nabla \tau, \tag{1}
\]

where \(\rho\) is the density, \(v\) is the velocity field, \(p\) is the pressure, and \(\tau\) is the deviatoric stress tensor. The fluid was assumed Newtonian with a viscosity of \(\mu = 4\) cPoise and a density \(\rho = 1.0\) g/cm\(^3\). The blood flow boundary conditions were derived from PC-MR images of the main branches of the circle of Willis obtained from a normal volunteer. Traction free boundary conditions were imposed at the model outflows. At the vessel walls, no-slip boundary conditions were applied. Since the velocity of the wall was estimated through imaging techniques, wall compliance is implicitly included in the simulation process. As it was not possible to determine the shape of the distension waveform at low sampling rates, it was assumed as a first order approximation, that it followed the flow waveform. Such waveform was scaled locally to achieve the measured displacement amplitude in each of the regions considered. The walls were assumed to move in the normal direction, and such motion was directly imposed to the 3D mesh derived from the volumetric medical images. The grid was updated at each time step by a non-linear smoothing of the wall velocity into the interior of the computational domain.\(^10\)

3. RESULTS

For wall motion analysis, each aneurysm was divided (depending on the topology) in up to three regions: bleb, dome and artery. Figure 2 shows the result of applying the wall motion estimation method to the different sequences selected for analysis. Next to each plot there is a picture showing the corresponding vessel segmentation into the regions aforementioned.

Sequence #1 corresponds to an aneurysm 12mm \(\times\) 9mm in size, located in the left internal carotid artery at the bifurcation with the left posterior communicating artery, and with a small bleb close to the artery. Visually, this aneurysm presents only pulsatile motion. The acquisition frame rate is 7.5 Hz which allows to obtain a waveform resembling a the typical curve of arterial pressure. According to the experiments the artery moves more than the dome and bleb. On the other hand the amplitude of motion for dome and bleb is similar which could be interpreted as the bleb just being “dragged” by the dome or having wall motion too small to be detected. Another interesting observation is that the phase difference between aneurysm and artery wall movements is close to 180° with the maxima of one curve coinciding with the minima of the other one.

In sequence #2 there is a lateral aneurysm 19mm \(\times\) 18mm in size located in the left internal carotid artery and containing a bleb. The sampling frequency in this case is too low (2 Hz) to describe a curve like the one seen in sequence #1. Thus the curves have a sinusoidal pattern. However, interesting conclusions can be derived from the wall motion analysis. First, and differently from the previous case, the bleb presents the highest motion amplitude with a clear difference with respect to the dome. Therefore the bleb is not just “dragged” by the dome but undergoes a deformation as well, which can be confirmed by visual inspection.

Sequence #3 corresponds to an aneurysm of the left internal carotid artery at the bifurcation with the left posterior communicating artery. As in the previous case the acquisition frame rate is low and the motion curves present a sinusoidal pattern as well. The motion analysis was divided into aneurysm and artery since there is not bleb in this case. The movements of both parts are in phase and have similar amplitudes.
Finally, sequence #4 contains a basilar aneurysm with a large rigid movement of the artery and no distinguishable pulsation in the aneurysm. It is important to quantify this motion pattern since this could potentially change the distribution of forces inside the aneurysm and therefore the risk of rupture. As the frame rate is 7.5 Hz, the curve has a more realistic shape like in sequence #1.

Figure 3 displays the results of CFD simulations for sequences #2 and #3. For these sequences a compliant and a rigid simulation were carried out. In the case of compliant simulations, wall motion was applied in a differential manner, i.e. each region of the anatomy was forced to move according to the obtained displacements described before. In order to compare the results obtained under different conditions, the following characterization of the wall shear stress (WSS) was used: the necks of the aneurysms were manually selected on the anatomical models, and a region of approximately 1 cm of the parent vessel (denoted proximal parent vessel) from the neck of the aneurysm was identified. At each time instant, the average WSS magnitude was computed in the proximal parent vessel region and used to normalize the WSS in the aneurysm and to identify regions of elevated WSS. Visualizations of the WSS distribution at five selected instants during the cardiac cycle and for all the wall motion conditions of sequence #2 reveal a region of elevated WSS in the dome of the aneurysm. Although this region covers a small area of the aneurysm (~1.4%), it contributes to a significant fraction of the total shear force (~8%), i.e. it is subject to a concentrated shear force (percentage of the shear force applied in these regions divided by the percent area of the regions). A graph of the WSS obtained with the rigid-wall model and the compliant model with maximum differential wall motion is shown in Figure 3 (b). This figure shows that rigid models tend to overestimate the WSS compared to the compliant models. However, the overall WSS distributions obtained with the different wall models have similar appearances and characteristics in spite of small local deviations.

Visualizations of the WSS distributions at the five selected instants during the cardiac cycle for sequence #3 are presented in Figure 3 (c). The visualizations reveal a stable region of elevated WSS from the neck to the dome of the aneurysm. This region covers approximately 2% of the area of the aneurysm. It is interesting to note that compliant models tend to yield slightly larger areas of elevated WSS. This may be due to lower WSS values in the proximal parent artery for distending vessels.

4. DISCUSSION AND CONCLUSIONS

In a previous work, it was pointed out the use of low acquisition frame rates (2 Hz) as one of the drawbacks in the methodology, and it was also suggested that the use of a higher sampling rate could help obtain better quality waveforms. In this work, two sequences (#1 and #4) acquired at 7.5 Hz were processed besides other sequences at 2 Hz (#2 and #3). The wall motion curves corresponding to sequences at high frame rate look more realistic than the ones acquired at low frame rate, and resemble the typical blood pressure waveform. This result confirms that the measured values correspond to the a real displacement of the vessel wall and are not caused by the noise present in the images, which constitutes a qualitative validation of the methodology as well. From a point of view of CFD simulations, the possibility of obtaining realistic motion curves is quite important as this can be imposed directly to deform the geometry without the need of prescribing any particular waveform as before. The use of frame rates higher than 7.5 Hz should improve additionaly the quality of the results, but the cost-benefit ratio between frame rate and patient irradiation remains an issue to be researched.

One of the problems of the method applied in this work is the need for manual selection of landmarks showing artifact-free motion patterns. The appearance of vessels confounding the boundary where a landmark is placed cannot be avoided, therefore an automatic method to select landmarks according to their motion pattern must be developed. This would allow to obtain a more continuous description of movement along the boundary of the aneurysm and curves more representative of the regional movement through the combination of point-to-point measurements.

Besides aneurysm pulsation, another motion pattern was found in basilar aneurysms consisting of an periodic angular movement of the feeding artery. Such pattern could potentially impact the flow inside the aneurysm by distributing the input jet on a larger area and therefore have an influence on the risk of rupture. CFD simulations including this movement pattern is an interesting research topic and is considered as a future work to this study. Another interesting finding was the phase difference between artery and aneurysm for patient #1. This is not a
Figure 2. Estimated wall motion for the selected sequences. Each analyzed region is labeled with different colors: bleb in red, dome in blue and artery in green.
consequence of the phase shift produced by the filtering, because the same difference can be found in the initial curves. This could be explained like an effect of the extra vascular environment, which also moves during the cardiac cycle and could be modifying the movement of the aneurysm by means of physical interactions. This is only an plausible explanation for the observations and additional experiments must be carried out to identify the cause of this behaviour.

To the authors' knowledge, this is the first time that wall motion is quantitatively measured in aneurysms, being the amplitude of wall motion lower than 0.5 mm for the cases analized in this work. As pointed out at the beggining, the amplitude and pattern of motion can be connected to hemodynamic patterns and risk of rupture. Therefore, this study should be extended to a larger population in order to gather sufficient statistics to establish such relationships.

In summary, so far there is a methodology to estimate wall motion in aneurysms and impose it to the 3D geometry aiming to create personalized CFD simulations. The next challenges are the development of a more user-independent method (to avoid the landmark selection step), the study of motion patterns on a larger population in order to establish relationships with risk of rupture, and the investigation on the effects that the artery displacement has on the internal force distribution.

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