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Development of a Patient-Specific FSI Model of an Arteriovenous Fistula

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Abstract

The Arteriovenous Fistula is a surgical connection between two blood vessels, which produces complex haemodynamic conditions. For a patient-specific study, a mode of imaging which captures the 3D geometry is needed. This paper shows the development of a new method using ultrasound to obtain the geometry and a subsequent FSI analysis which incorporates a vessel wall and surrounding tissue.

Introduction

End-stage renal disease (ESRD) patients require dialysis in order to filter their blood of toxins. An access point which has high flow rates does not occur naturally in the body, and therefore, an Arteriovenous Fistula (AVF) is created in the arm. The surgery consists of the vein being connected onto the artery, providing a short circuit back to the heart and encouraging the vein to remodel. The remodelling effects are a change in vessel diameter which can either be enlarging as part of the maturation phase, or narrowing due to disease. This remodelling has been shown to be highly patient-specific, and the driving variables remaining unknown [5].

In approximately 40% of patients, the vein fails to remodel appropriately, or occludes due to neo-intimal hyperplasia. Computational studies use idealised geometry with rigid walls [8, 3], however, the actual vessel walls are compliant and this simplification inaccurately describes the flow patterns and wall shear stress. Moreover, the compliance in the walls influences the surrounding structural domain, which also then exerts a force back on the fluid. Previous AVF studies have only looked at atmospheric conditions surrounding the wall, but a compressible tissue could have a large role in the vessel remodelling process [7].

Difficulty also exists in obtaining accurate patient-specific geometry for ESRD patients, as the dyes used in techniques such as MRI cannot be used with these patients; generally only 2D ultrasound is available. These geometries change significantly over a timescale of months after the AVF is created, and therefore, it would be advantageous to capture the effects in high temporal resolution as a means of investigating the driving variables for remodelling. As ultrasound imaging is cheap, portable and non-invasive, we have developed a method of reconstructing a 3D volume from 2D ultrasound images and using it to construct a patient-specific FSI model.

Methods

Ultrasound is a cheap and non-invasive way to obtain a 2D image of a vascular geometry but for CFD analysis, a three-dimensional volume must be created. We have developed a method of using a 3D camera which tracks the ultrasound probe

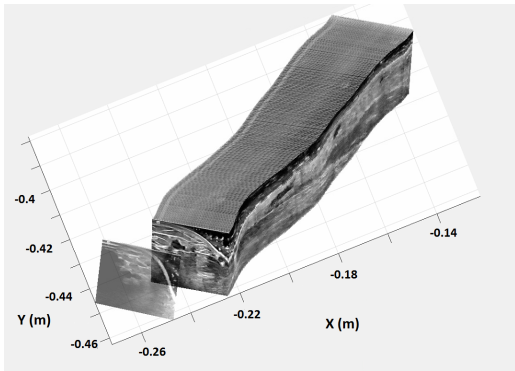

Figure 1: A set of ultrasound images stacked on each other to create a volume.

using infrared and stores the current 2D ultrasound image along with the real world position at 30 frames per second. As the probe is scanned along the patients' arm, an array of ultrasound images can be spatially put together as shown in Figure 1. The resulting volume is digitally segmented using thresholding based on greyscale intensity values.

Initially, the volume is median filtered, as this removes noise but preserves edges better than other filters. The sigmoid filter is also used, with the Beta value corresponding with the grayscale value at the gradient of the vessel wall and the Alpha value controlling the sensitivity. A region growing algorithm offered by the ScanIP software environment (Simpleware Ltd. Exeter, UK) is suitable for extracting the vessel (Figure 2) but is not for the wall and tissue. Patient specific extraction of wall and tissue from the ultrasound is possible to do manually, but various techniques will have to be employed to have semi-automatic segmenting, and for it to be reproducible. This is currently still being developed. However, the methodology is developed so that non uniform walls and tissue can be implemented into the FSI model at a later stage.

In the case described here, an average wall thickness was determined from the ultrasound images to be 10% and this was applied along the vessel as shown in Figure 3. A surrounding tissue surface was also created, ensuring that it encapsulated the entire geometry, as pictured in Figure 4.

The surfaces of each part (vessel, wall and tissue) are created in ScanIP and exported as an STL file and then imported into ICEM for meshing. The inlets and outlets have to be clipped slightly so that the vessel and wall are in plane and then capped so a volume can be created. The vessel and wall are meshed separately, with refinement based on curvature proximity; 1.4M and 66k tetrahedral elements respectively to achieve grid inde-

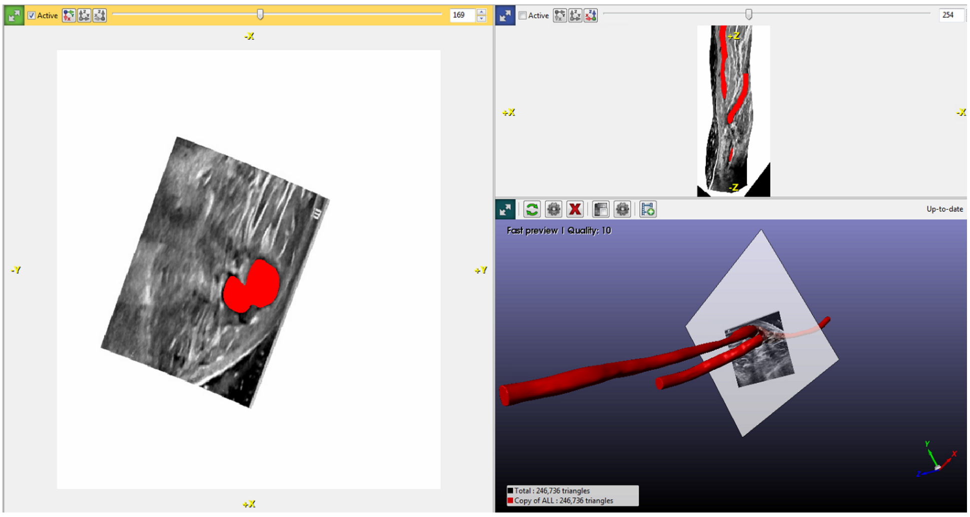

Figure 2: ScanIP software is used to segment the ultrasound image (left) to create the vessel (bottom right).

pendence. The mesh was refined from 2.6M, 1.4M and 800k tetrahedral elements in the fluid domain so that convergence was achieved with minimal amount of cells needed, based on flow velocity and pressure. The tissue zone is meshed independently to create 150k tetrahedral elements.

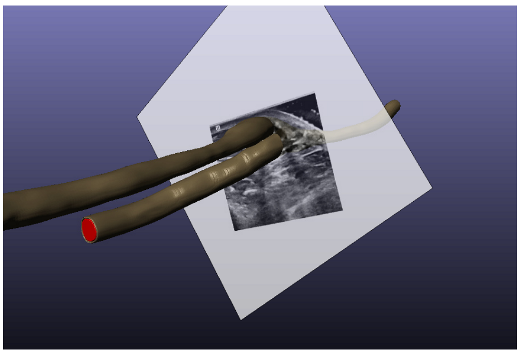

Figure 3: Creating a wall based off the ultrasound image and applied along the vessel.

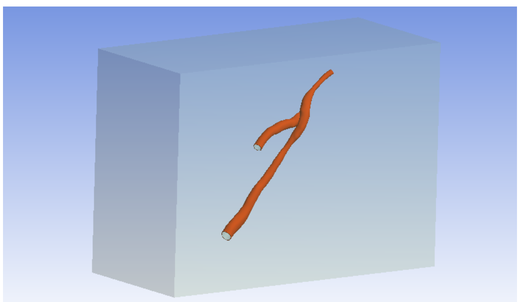

Figure 4: A surrounding tissue that is of uniform density.

FSI was modelled using Fluent, ANSYS and system coupling. One-way is set up in this initial study to check that convergence is reached between the fluid and structure and that the loads are not excessive or unrealistic.

For the fluid component, a SIMPLE scheme was used for pressure-velocity coupling and second-order upwind scheme for the spatial discretization. Blood is modelled as an incompressible, Newtonian fluid with a dynamic viscosity of 3.5×10^{-3} Pa s and a density of 1060 kg/m^3. The artery inlet and outlet are given velocity profiles which have been extracted from duplex ultrasound shown in Figures 5 and 6. It should be noted that for this particular patient, the artery outlet has reversed flow throughout the whole cardiac cycle, and therefore, all the flow exits out the vein.

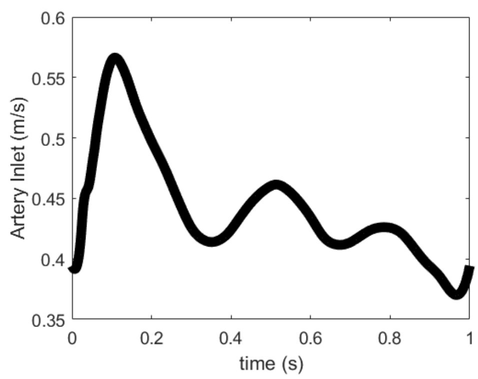

Figure 5: The artery inlet velocity which is taken from Doppler Ultrasound for one cardiac cycle.

The vessel properties for the wall in this model are given as density of 1060 kg/m^3 and hyper-elasticity Yeoh 3rd order model with constants of 76.3×10^2, 3.697×10^5, 5.31×10^5. The tissue properties had a linear elasticity with youngs modulus of 1 MPa and Poissons Ratio of 0.3. These properties are taken from literature [2], but in future work will be determined specifically from each patient, using techniques based on ultrasound elastography. While still in an early stage of development, and predominately used for examination of breast tissue, the mechanics can also be applied to the arm. An elasticity map is created by measuring the shear wave propagation by pushing the ultrasound probe into the tissue [6].

The model was run for three pulse cycles to remove any transient effects and a timestep of 5×10^{-2} was chosen to keep the max Courant number less than 1. With 8 processors and SSD for reading/writing, the simulation ran for a total of 14 hours.

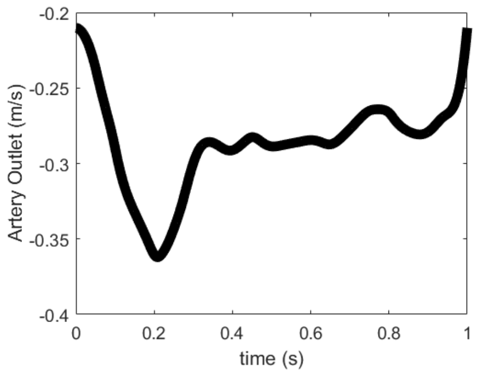

Figure 6: The artery outlet velocity which is taken from Doppler Ultrasound for one cardiac cycle. The flow in this part of the artery is reversed.

Results

In order to understand the remodelling process, the flow, wall and structural features are analysed. Based on the methodology prescribed above, this is possible on a patient-specific level.

The fluid dynamic results can be examined for regions where there are recirculation zones and flow separation. These are typically where the vein joins the artery, and are an indication of areas where non-uniform remodelling happens. Additionally, in regions of narrowing due to intimal hyperplasia, the velocities increase and become a jet-like formation as seen in Figure 7. This has further implications downstream where the flow can be highly disturbed as shown in Figure 8.

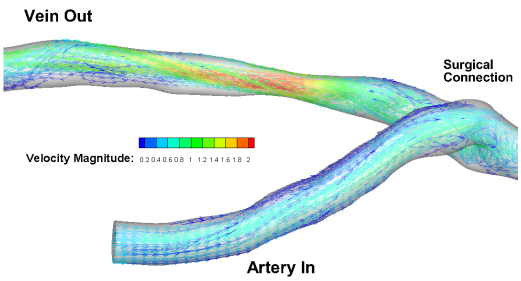

Figure 7: Streamlines in the artery and vein showing disturbances in the flow due to the geometry.

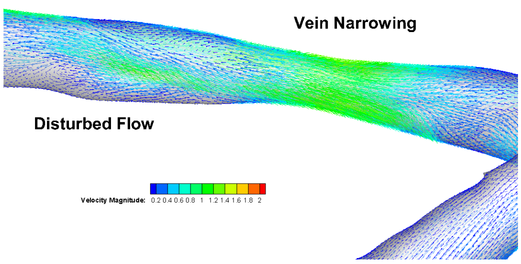

Figure 8: The velocity vectors at the peak of the cardiac cycle (systole). These are high disturbed in the region just after a narrowing.

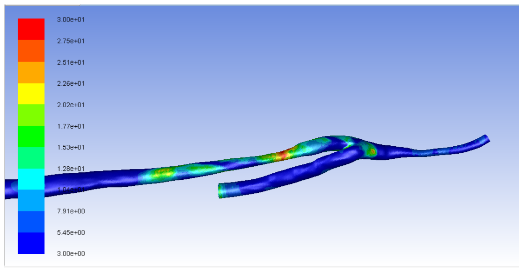

Figure 9: The WSS is examined as a metric for determining possible sites of remodelling.

The wall shear stress (WSS) is also an important flow metric and previous research has investigated the magnitude, gradient, time-averaged and oscillatory effects. These have been shown to be correlated with the activation of the cells lining the inner wall of the vessel, the endothelial cells [4]. While they have a role in the remodelling process, the mechanisms involved are not yet fully understood. With patient-specific geometries, the WSS effects are different to idealised geometries and also vary between patients. This makes it difficult to achieve WSS metrics that can be categorised into good or bad. Moreover, as the walls are compliant, studies have shown that the temporal stress phase angle, which is characterised by the WSS and circumferential stress, should be included [1]. This metric is part of the future work and requires an FSI model to evaluate the non-uniform circumferential stresses in the wall.

In this one-way FSI model, the cross-sectional stresses in the tissue can be observed in Figure 11. As the geometry is non-planar, only a portion of the geometry is in view. Snapshots of the tissue stress have been taken at systole and diastole of the cardiac cycle for emphasis of the travelling pulse wave in the vessel. At the start of the cycle, the stress is mostly distributed uniformly in the artery wall and tissue with peaks at the connection of the vein and artery. Very little stress is in the narrowing found in the vein for both the wall and tissue. At systole (0.2s), much larger stresses are distributed unevenly in the vein wall and greater intensity in the surrounding tissue.

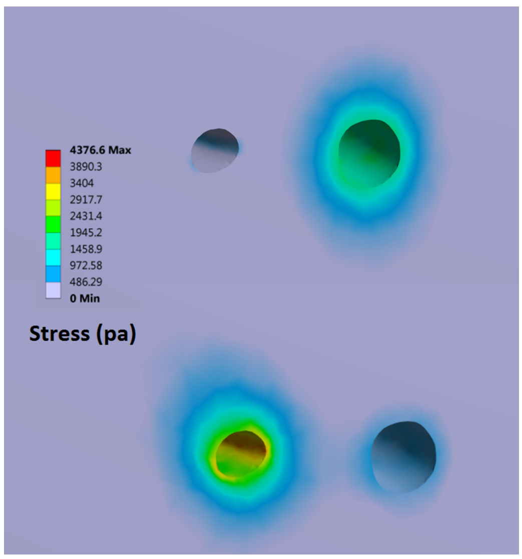

Figure 10: A slice view of the artery (right) and narrowed vein (left) showing the distributed stress in the tissue at diastole (top) and systole (bottom).

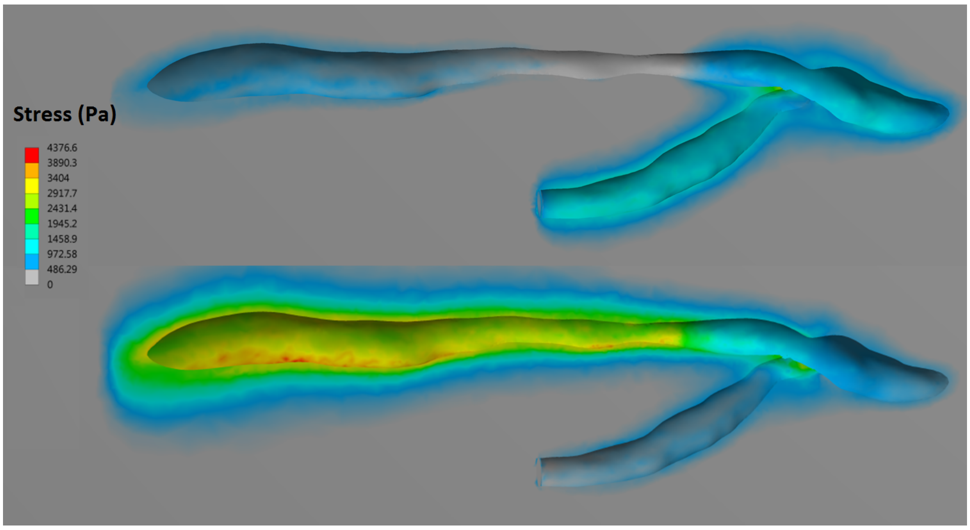

Figure 11: A cross-sectional view of the geometry showing the stress in the tissue and wall at diastole (top) and systole (bottom).

In a slice orientation (Figure 10), the vessel is shown to be non-symmetric with the narrowed vein (left) and artery (right). This affects the circumferential stress, even if the internal pressure was uniform.

The tissue is modelled as uniform in this case, but from the ultrasound scans, the density and properties will vary along the arm, especially where scarring from the surgery forms. Furthermore, the vein in some patients can be located directly next to the skin, which could produce structural and haemodynamic conditions from the asymmetry which have not been accounted for.

As the geometry is unique to each patient, it is possible for the remodelling to occur at different rates and stages and thus a sensitivity study for each driving variable will be conducted. We can establish a baseline of the metrics before the AVF is created and then at intervals as the geometry remodels. This will be done over many patients to obtain statistical significance.

Conclusion

A method of extracting patient specific geometry from an ultrasound based imaging system has been developed. This is integrated to provide Fluid and Structural domains to be coupled together for FSI analysis.

The FSI model is currently one-way, such that, only the fluid forces are exerted on the wall and then transferred into the surrounding tissue. But in reality, the fluid and structural domains are highly coupled and both the tissue and wall exert a force back into the fluid domain. Additionally, this also changes the shape of the vessel which requires remeshing and re-solving. Future work sets to achieve a patient-specific two-way FSI model that gives a detailed analysis into the variables involved with not only the cardiac cycle, but a longer remodelling time-frame. This will take into account a combination of flow patterns, WSS metrics and structural stresses.

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