Using fluid dynamics to improve vascular access in haemodialysis

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Abstract
Sustaining viable vascular access for dialysis patients is a challenging problem. Vascular accesses become thrombosed, requiring vascular or endovascular intervention, which adds considerable cost to the health care system and inconvenience to the patient. Stenosis and thrombosis of vascular access has been linked to high access flows, particularly in relation to the cannulation site. In particular, high flow rates may cause development of intimal hyperplasia, leading to stenosis and thrombosis. Computational modelling and benchtop flow experiments to understand fistula haemodynamics and needle flow dynamics are described in this paper.

Keywords
Haemodynamics, cannulation, arteriovenous fistula, computational fluid dynamics (CFD), experiments.

Introduction
Dysfunctional vascular access is a common problem that leads to significant morbidity, mortality and cost (Mantha, 2011). An important initiating event in access failure is fluid dynamic stress, caused by unusual blood flow conditions due to the creation of a fistula, or the blood flow from the venous needle.

Our fluid dynamics laboratory in the School of Mechanical and Manufacturing Engineering at the University of New South Wales (UNSW) has a focus on studying these flow conditions, using models (computational and benchtop flow experiments) to understand how the fluid flow can contribute to vascular access complications. Figure 1 shows the procedure used in our laboratory to direct the research.

Background
Failure of vascular access, due to stenosis (blockage) is common and is estimated to cost US$1 billion/year (USRDS, 2009). This costly and life-affecting problem can be reduced by finding the reasons behind the access failure. Access dysfunction, primarily due to venous intimal hyperplasia development and stenosis formation, is mainly attributed to complex haemodynamics within the fistula (Carroll, 2011). Common arteriovenous fistula geometries cause the blood to take sudden path changes and the necessary high flow rates cause elevated shear stress. Suction of the blood by the arterial needle causes a turbulent flow around the needle, and the venous needle jets the blood stream against the vessel wall.
Brouwer suggested the time has come for research to answer the many basic questions about the best practices of cannulation so we can move cannulation from an “art” to a “science” (Brouwer, 2005) and that little research is available to use when teaching needle cannulation (Brouwer, 2011). These basic questions remain unanswered in the literature.

The most important initiating event in access failure is fluid dynamic stress, especially regions of low shear stress and turbulence (Roy-Chaudhury, 2003). In haemodynamics, wall shear stress (WSS) is the mathematical description of the blood flow effect on the endothelial cells (the thin layer of cells that line the vessel wall) and has been used extensively to analyse the haemodynamic conditions in various geometries which may lead to vascular dysfunction (for example, Rayz, 2010). The direction corresponds to the blood flow direction where $\text{WSS} = \sigma$. WSRS, (where $\sigma$ is whole blood viscosity, WSR is the wall shear rate). The units of WSS are Pa. Wall shear rate is defined as the difference between adjacent velocities in the vessel interior, divided by their distance.

WSS can be measured both computationally and experimentally (via an in vitro flow rig): however, it is difficult to gain accurate experimental measurements. We use a closely coupled series of simulations, using experiments in our dialysis flow rig to validate our computational findings. This flow rig allows very detailed measurements of the flowfield in the near vicinity of an idealised arteriovenous fistula and the flowfield around the venous and arterial needles.

Experimental models

In 2011, a new experimental set-up for flow measurement of arterial and dialysis fluid dynamics was developed at UNSW. The modular rig allows experiments to be conducted for a number of needle and fistula variations, including needle angle, orientation and placement (Figure 2). We use Laser Doppler Anemometry (LDA) and Particle Image Velocimetry (PIV) laser flow measurement systems. Currently our models have rigid (non-compliant) walls. To gain correct flow pulsatility, a modified CardioFlow 5000MR Computer-Controlled Flow Pump System, along with a number of dialysis-similar peristaltic pumps, are used.

To gain optical access, any sections of the model where measurement is required are machined from (cast) acrylic, which are then polished by machine and by hand, until a high-quality finish results in a high degree of clarity. To gain accurate data from the optical measurement techniques, the refractive index of the fluid and the test section material is matched. It is also necessary to match, to some extent, the viscosity of the test fluid to that of blood, to both simulate blood flow and to lubricate the pump.

Computational models

Computational fluid dynamics (CFD) models are also developed. Models use the Reynolds Averaged Navier Stokes equations (and sometimes the filtered Navier Stokes equations) to calculate the blood flow using a computer model. In CFD, the geometry is represented by a series of smaller elements on a mesh, and the equations are iteratively solved at each point. A high level of detail can be gained from the model, though it is critical for the computer model to be verified and validated for accuracy. In our models, the fluid is modelled as constant viscosity and density, Newtonian flow, and the models are assumed, currently, to have rigid walls. Also, the fluid is assumed to have isothermal properties, as blood temperature in the physiological system is highly regulated by the body. Verification of the model involves confirming that mesh independence is achieved by comparing swirling zone lengths, maximum velocities reached, and comparing meshes of different densities. Iterative convergence and transient convergence are checked by similar means.

Cannulation

The needles used in dialysis influence the fluid dynamics of the access. Suction of the blood by the arterial needle causes a turbulent flow and the venous needle jets the blood stream against the access wall (Figure 3). The venous needle (15G sharp with backeye) is oriented towards one wall causing the high-velocity flow jet to impact the lower wall of the blood vessel, causing WSS values that are above values known to cause cell damage. Downstream of the venous needle, where the flow jet entrains the normal flow, there is a region of flow instability and turbulence and high-shearing forces can be observed (Unnikrishnan, 2005). In our previous work, we have shown that the effects of needle location and angle on the vessel WSS can be significant (Fulker, 2013). Our results indicate that inserting the venous needle at a low angle in the...
vessel, inserting the arterial needle centrally within the vessel away from the swirling zone caused by the venous needle, and using lower pump flow rates may minimise vessel damage.

Very little research is available to use when teaching the art of needle cannulation (Brouwer, 2011), and our discussions with Renal Society of Australasia (RSA) nurses demonstrates the range of techniques used at different centres. Our work may assist in determining if any particular technique has an advantage over another.

Plastic cannulae are also under investigation by our team, with initial results showing interesting variations between the fluid flow for the metal needle and the plastic variation. In particular, we note a higher jet flow velocity and greater blood flow exiting from the side holes.

**Fistula haemodynamics**

The geometry of the fistula may also create physiologically unusual haemodynamic conditions which may cause access failure. As the arteriovenous anastomosis has a complex geometry, blood flow with relatively high Reynolds characteristics can be found causing turbulent flow and there are regions of high WSS along the venous wall, with significant flow separation and swirling zones (Lee, 2007). To investigate this, a side-to-side fistula was modelled, both computationally and experimentally.

Our results showed that as the flow moves through the feeding artery there is acceleration of the blood flow and in the region around the anastomosis the blood experienced significant flow disturbance (swirling and unsteady flow behaviour). The venous region of the fistula was found to contain a large swirling zone and two stagnation points were observed, one at the toe (the distal side of the anastomosis) and the other along the venous wall. Very low WSS regions were noted in the draining vein, which, in clinical studies, has been shown to correspond to failure from thrombosis and stenosis formation (Van Tricht, 2005). Elevated WSS was found in the feeding artery, toe and along the venous wall nearby the toe. These elevated levels of WSS will promote further dilatation of the vessel in attempt to achieve homeostasis.

**Conclusions**

The combination of computational and experimental models allows our laboratory to study fluid dynamics problems of interest to the renal community. Collaboration with our clinical colleagues provides the research questions and we hope to provide results that will be of use in prolonging vascular access. Information regarding the haemodynamics of cannulation and fistula flow have been presented here as examples and we have found that shallow needle angles and lower pump flow rates may minimise vessel damage. Our results will allow recommendations to be made to both dialysis unit staff and vascular surgeons, which may lead to less vascular access failure.

**References**


