Development of an anatomical carotid artery flow phantom for the calibration of Doppler ultrasound systems

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ABSTRACT
Cardiovascular diseases are responsible for over 50% of all deaths in the UK. Current measurement techniques involve non-invasive Doppler ultrasound imaging of blood velocity, however it is known that measured velocity may be in error by typically 20-60%. This paper presents the development of anatomically correct tissue equivalent vessels for calibration of Doppler ultrasound by particle imaging techniques. Patient specific arterial MRI data is used as the basis for construction of a 3D CAD model. The model was modified to simulate varying degrees of stenosis (narrowing). The arterial geometry is fabricated by Stereolithography to generate investment cast patterns from low melting point alloy. The expendable cores are then used in the construction of optically transparent models for particle image velocimetry (PIV) analysis and for agar models using an acoustically matched tissue mimic material for Doppler ultrasound measurement. Issues concerning the fabrication of models for direct comparison of Doppler and PIV data will be discussed.

KEYWORDS
Vascular, Phantom, Stenosis, CAD, Stereolithography

1.0 INTRODUCTION

Cardiovascular diseases are responsible for over 50% of all deaths, and over £2 billion per annum is spent on healthcare related to arterial disease in the UK alone. Doppler ultrasound techniques are widely used to measure blood velocity, and to assist in the measurement of degree of disease. Several blood velocity quantities are of interest. Maximum velocity is used to estimate the degree of lumen reduction within atherosclerotic plaque; volumetric flow is used to provide an indication of the effect of disease on blood flow to a specific organ, and there is an interest in the measurement of wall shear rate (velocity gradient at the vessel wall) as abnormalities in wall shear rate have been shown to be associated with the development of disease.

Because the dynamics of blood flow (haemodynamics) depend to a great extent on the geometry of the affected vessels, there is a need to develop anatomically accurate flow phantoms of the human carotid artery in order to establish the nature of blood flow in healthy and diseased vessels. Here, we describe the computer-aided design and manufacturing techniques that we have developed to fabricate phantoms based on medical images such as MRI and CT scans, and which can be used in conjunction with state-of-the-art laser velocity

measurement techniques, for example, Particle Image Velocimetry (PIV), to provide as accurate a description of the flow as possible. The phantoms are placed in a mock circulation where the flow field is interrogated to reveal the type of flow structures present and to quantify the distribution of wall shear stress within them. This work forms part of a major research project that is being carried out in a collaboration between the Universities of Liverpool and Edinburgh, the aim of which is to develop vascular phantoms for the calibration of the next generation of Doppler ultrasound probes used to screen patients with arterial disease.

In this paper we present investment casting techniques that we have developed to fabricate anatomically accurate arterial phantoms for use in experiments involving laser measurement techniques, on the one hand, and clinical Doppler ultrasound systems, on the other. This is not as straightforward as it might at first seem, as the physical and mechanical properties of the tissue and blood mimics used in each case are very different, having been developed specifically for their respective imaging modality (acoustic versus optical) [1], [2]. Moreover, in order to ensure that the flow fields in the respective phantoms are identical, dynamic similarity between the two systems must be maintained before detailed cross comparison of the flow fields in the optical and acoustic models can be made, and the flow rates and pulse frequencies used in each case must therefore be optimised. The flow field in an optical model has therefore been visualised with a view to defining the experimental conditions under which subsequent ultrasound and PIV measurements will be made.

Much of this work is novel, for example, the development of rapid prototyping techniques for the construction of anatomically correct physical models, and matching of wall motion in both acoustic and optical systems. In the longer term, we anticipate that research of this kind may help us establish the extent to which haemodynamic factors influence the onset and progression of cardiovascular disease.

2.0 METHODS

2.1 Data Collection

For this work, a sequence of MR slice images from a healthy patient was imported into Mimics (Version 6.3, Materialise BV, Leuven, Belgium). This software is dedicated to importing 2D CT or MR slice data, enabling the data to be manipulated and 3D virtual models to be generated.

The manipulation of the dataset first involved assigning threshold values to the greyscale images such that only tissue of the correct density is selected. While identification of highly radio-opaque, calcified tissues is straightforward, great care must be taken in the case of soft arterial tissue to ensure that an appropriate threshold is selected, as much of the surrounding tissue is of similar density.

Once the area of interest was identified, each individual slice was inspected and edited. Artefacts inherent in the images (as a result of turbulent blood flow, for example) were corrected at this stage, as shown in Figure 1.

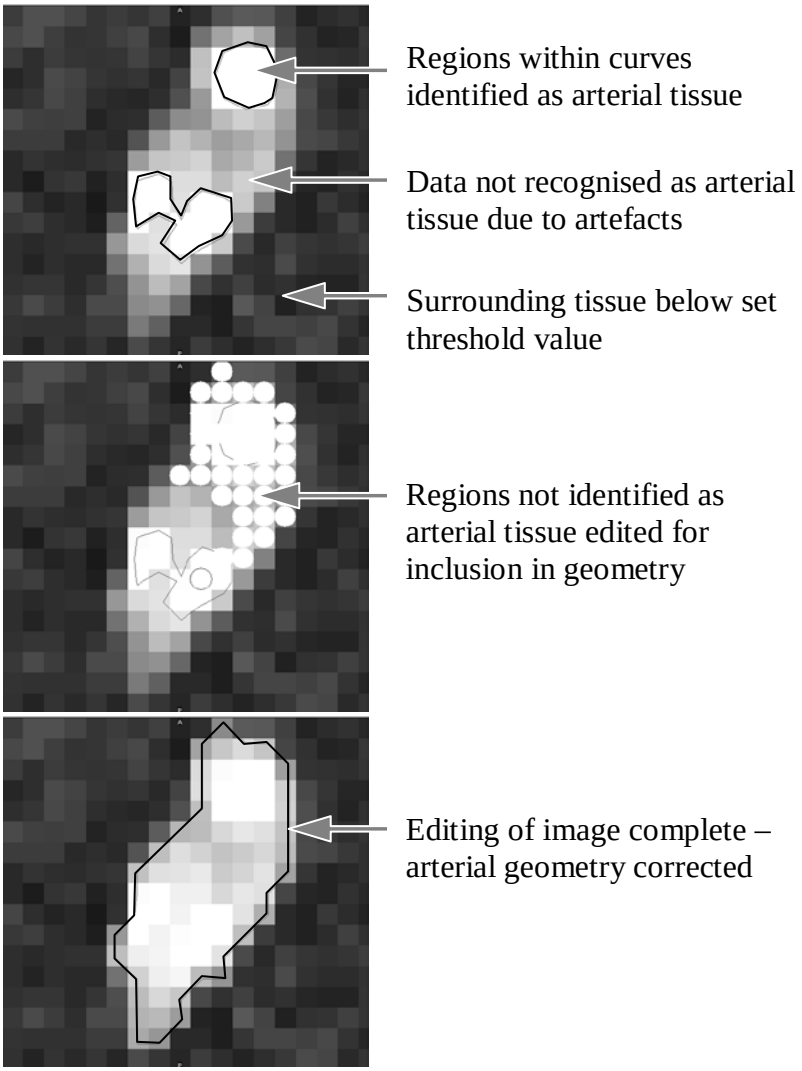

Figure 1: Individual slice correction in MIMICS.

The poor resolution of the MR system (0.5-1.0mm pixel size) required the superimposition of *NURBS* (non-uniform rational B-Spline) curves around the perimeter of each region to more accurately represent the arterial sections, as shown in Figure 2.

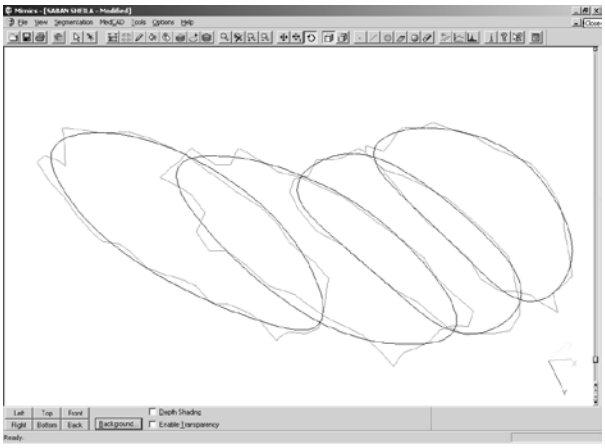

Figure 2: Superimposition of NURBS curves to multiple slices.

Once complete, the curves were exported to a solid modelling CAD system (Unigraphics).

2.2 Data Refinement & Surfacing

IGES curves were imported into the CAD software and scrutinised, one slice at a time, and closed loop splines were fitted through selected points on each curve. A 3D free-form surfacing technique was used to create solid bodies through multiple slice splines (3^{rd} order polynomial surfaces); solid sections were created with tangent surfaces and solid bodies were united together. The spline control points were refined in consultation with a radiologist (J. Wardlaw) and, after several iterations, a more realistic generic artery geometry was constructed as a baseline data set, see Figure 3.

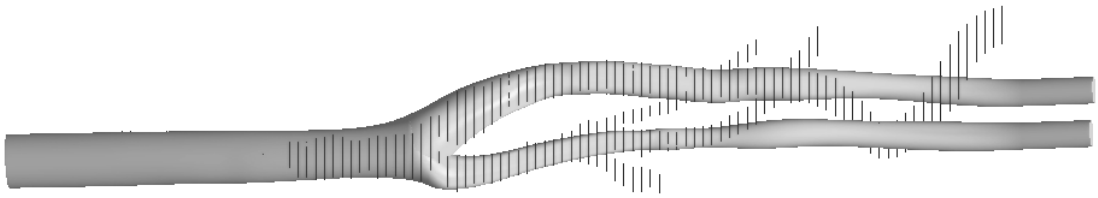

Figure 3: IGES slice curves and completed healthy baseline CAD model.

Cylindrical sections were added at each end, and their central axes aligned in parallel, to facilitate incorporation of the phantom into the flow circuit.

2.3 Addition of Stenoses

Using the above CAD manipulation techniques, the baseline generic dataset was modified to model differing degrees of disease, again in full consultation with the radiologist. The technique used by clinicians to categorise patients with disease defines the degree of stenosis in terms of the lumen diameter at two specific locations as shown in Figure 4.

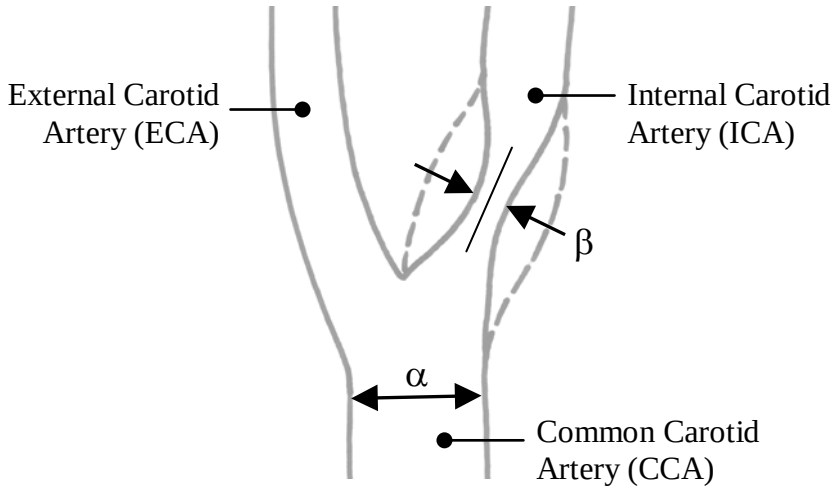

Figure 4: Sketch of carotid artery identifying locations where diameter values are measured for clinical classification of disease level.

The diameters are measured from a single side view and therefore do not take into account the non-circular shape of the cross-section. To adapt the measuring technique for this 3D application, the section area values were measured and an effective diameter calculated – by assuming the area was circular in cross section, this allowed the 2D diameter values to be taken from any view side. These effective diameters could then be used in the following formulae to calculate the degree of narrowing.

$$\% \text{ Stenosis} = \frac{\alpha - \beta}{\alpha} \times 100$$

α = Effective diameter of CCA where the lumen walls are parallel before the bifurcation.

β = Effective diameter of lumen at narrowest point (centre of stenosis).

At various locations along the length of the CCA and ICA the cross-sectional area was measured and the centroid of each section identified. These centroids could then be joined together by a central spline running through the entire geometry. With the start and end positions of the diseased region known, the centre point along the central spline could easily be found. At this point the solid CAD model of the healthy baseline dataset could be cut, slicing it perpendicular to the central spline. The area of this perpendicular flow slice could then be scaled about its centroid by different amounts to artificially simulate varying degrees of disease, using the clinical method previously described. Free-form modelling through the various sectional area curves was used to create the solid stenosis bodies, with tangent surfaces to the CCA and ICA respectively. This method was repeated, building up a group of similar models that only differed by the extent of the stenosis. This group contained five geometries: the initial healthy artery (0% stenosis), and four models with progressive levels of stenoses, i.e. 30%, 50%, 70% and 90% all accurate to one decimal place, see Figure 5.

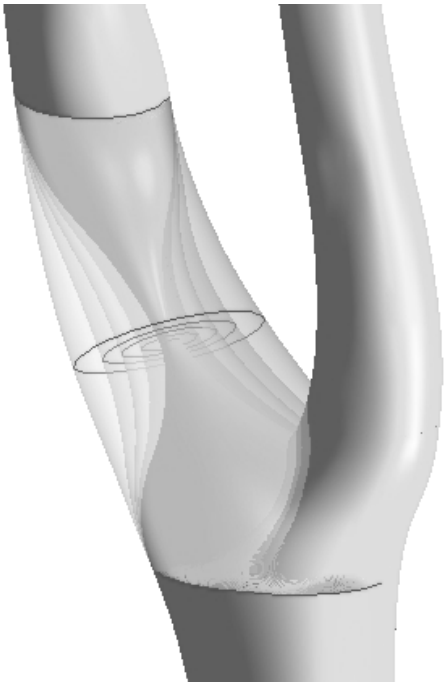

Figure 5: Manipulation of healthy CAD model to artificially simulate varying degrees of narrowing.

2.4 Production of Models

After the CAD models were complete, solid master patterns were produced by means of rapid prototyping technology. The CAD data was exported in stl file format, allowing StereoLithography Apparatus (SLA) to build life size physical prototypes from photosensitive resin, with a layer thickness of 0.025mm to minimise stair stepping effects and reduce the amount of manual finishing required. These SLA master patterns were used to create soft silicone tools that split along a pre-defined parting plane, via a vacuum casting process. Low melting point (47°C) alloy geometries were cast from the soft tools as expendable cores for subsequent use in investment casting. These required finishing and polishing by hand, see Figure 6.

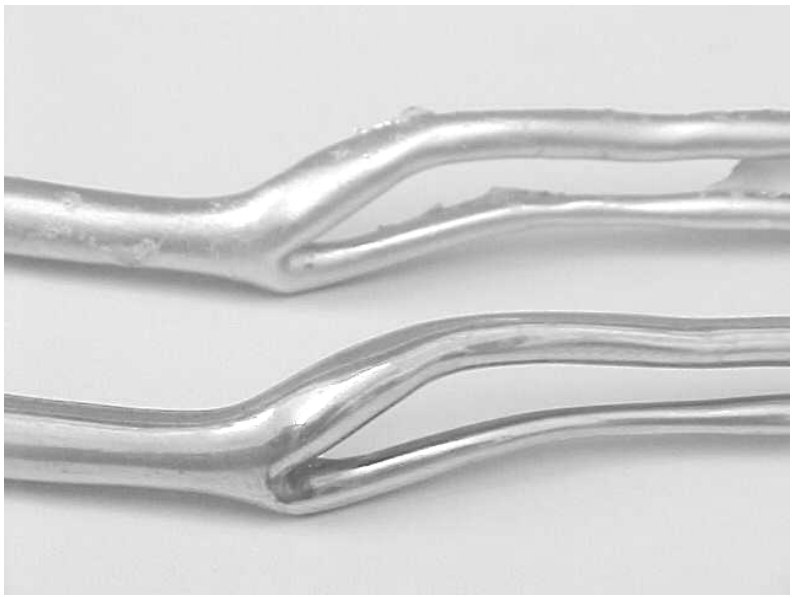

Figure 6: Low melting point expendable alloy cores for use in investment casting, as cast (top) and after hand finishing and polishing (bottom).

Each of the artery geometries needed to be cast twice, once in optical grade silicone (Sylgard 184, Dow Corning) and once in an acoustically matched tissue-mimicking agar. To ensure that the results are comparable between the two types of analysis, it is essential that each of the two boxes held and orientated the artery geometry in exactly the same way. For this reason the boxes needed to be identical in many respects, which also enabled easier manufacture of the two different box types.

The boxes were constructed from aluminium, using a square hollow extrusion for the main part with aluminium end plates. The PIV box design incorporated removable Perspex windows. This allowed the silicone to have a smooth surface to cast against whilst enabling their removal for PIV analysis, see Figure 7. The agar box design incorporated an acoustic window with a well for ultrasound gel. The two design variations were detailed in CAD and engineering drawings were produced. The parts were made by CNC machining.

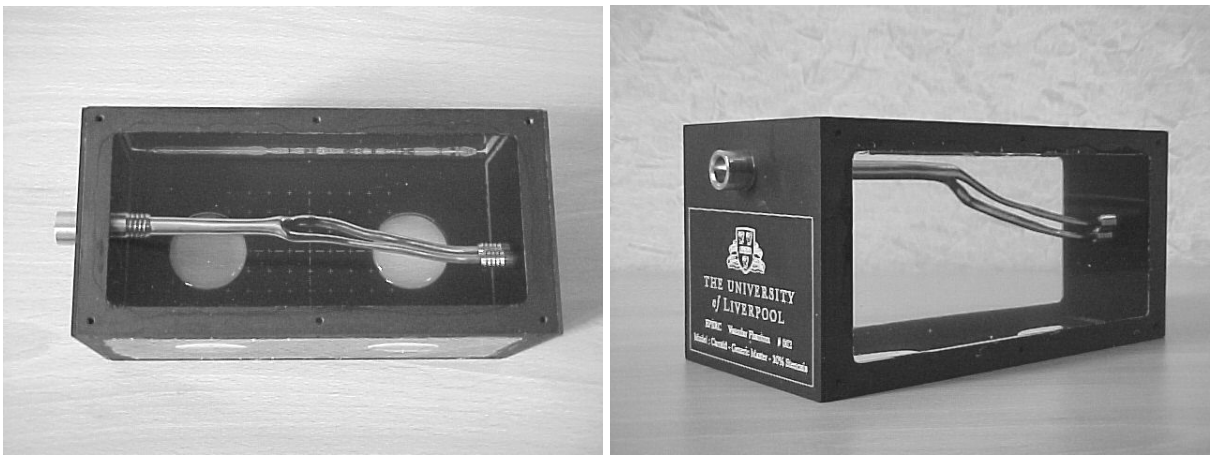

Figure 7: Optical phantom before removal of polished core, for use in PIV analysis.

3.0 DISCUSSION

The baseline data set was constructed in CAD by manipulation of IGES curves derived from patient specific MR data. This baseline data set was modified, by a process of iteration and according to specialist knowledge, until a generic model of a typical healthy carotid geometry was formed. Out of necessity, the geometry was simplified, both for practical reasons, and to meet the specific objectives of the project. Small branches after the main bifurcation were removed, and the vessels smoothed and straightened out. In addition, short cylindrical sections were added to each end along with a simple coupling arrangement to facilitate incorporation of the models into their respective flow rigs.

Despite the considerable deviation between original patient MR data and the final flow phantoms, we believe the resulting models are representative of the human carotid artery in health and disease, and that the fabrication technique described in this paper represents a considerable advance on existing vascular phantom design and manufacture. Previously, CNC machining has been used to create aluminium tools for casting expendable cores, thus limiting the geometries to being planar with circular artery sections, [3], however, the use of vacuum casting to create soft silicone tools from SLA master patterns enabled non planar geometries to be cast with artery sections which retained their true, non circular shape.

While this phantom construction method allows optical and acoustic models to be made from the same cores, there is a practical limitation of the physical models when considering highly stenosed geometries at their original, life size scale. Because the 90% stenosed carotid geometry had a cross sectional area of around 1-2 mm^2 at its narrowest point, manual handling during finishing and polishing of both the SLA master pattern and the expendable alloy cores often resulted in brittle fracture, particularly at low temperatures. Highly stenosed alloy geometries were also more prone to the effects of creep if left unsupported from the tool.

Future work includes quantifying the manual finishing deviations between the SLA master patterns and the low melting point alloy expendable cores by laser scanning for comparison. The optical silicone has been observed to degrade over time, discolouration and shrinkage effects will be of future interest to enable accurate comparison between the optical and agar

models, indeed all of the flow phantoms may have an optimum shelf life. A similar study of Femoral artery geometries is underway.

4.0 CONCLUSIONS

The use of CAD, rapid prototyping and vacuum casting techniques has enabled multiple anatomically accurate carotid artery flow phantoms to be created with relative ease from the same initial MR data set. The flow visualisation performed on the optical models prior to PIV analysis illustrated the complex 3D flow patterns that exist even in healthy geometries, see Figure 8. The final CAD models can be readily imported into other virtual analysis tools, for example Computational Fluid Dynamics (CFD), allowing further comparison of the flow field in the same arterial geometry to be made.

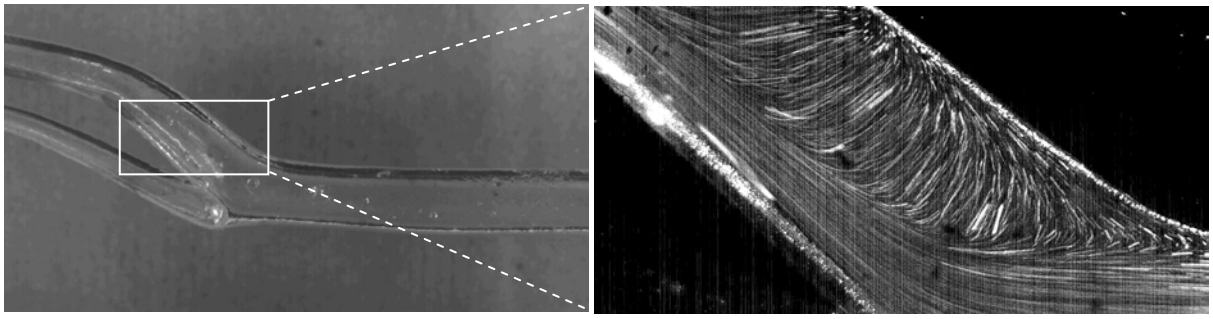

Figure 8: Optical phantom (left), showing internal and external carotid arteries from a healthy individual, and laser illumination of the internal carotid artery (right), showing the complex flow patterns that exist - even in a healthy individual - during diastolic flow.

5.0 REFERENCES

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