Introduction

The incidence of death from noncardiac complications following cardiac surgery is increasing while death from cardiac complications has plateaued.\(^1\) Atherosclerotic emboli from the ascending aorta have been linked to stroke, multiorgan failure and death.\(^2\) The force of the high velocity jet of blood exiting the aortic cannula and striking the diseased aortic wall, described as sandblasting, is a recognized cause of atheromatous emboli generation.\(^1\) The aortic cannula, a seemingly simple conduit connecting the extracorporeal circuit to the patient, is an extremely critical component. The aortic cannula has the smallest cross-sectional area in the entire circuit and is consequently the point of highest flow velocity in the extracorporeal system. Aortic cannula design is a major determinant of the flow pattern and flow velocity within the patient’s aorta. While numerous studies have alluded to cannula velocity and jet flow pattern,\(^3\) none has directly measured flow velocity at the cannula exit and few have attempted to characterize the flow pattern of aortic cannulae.\(^3\)

The purpose of the study was to map the peak flow velocity of various aortic cannulae and to characterize the flow pattern of two cannula designs. We performed in vitro velocimetry measurements, using laser Doppler technique, during perfusion with five aortic cannulae at the exit site within a cast of a human aorta attached to a mock circulatory loop. In vivo examination of exit sites were also performed with the use of transoesophageal (TEE) colour flow Doppler imaging to characterize the flow pattern of two cannula designs. Cannulae were tested under continuous and pulsatile flow conditions.

Methods

A mock circulatory loop and extracorporeal circuit was assembled as shown in Figure 1. The transparent, flow-through replica of a human aortic arch was fabricated under strictly controlled conditions during harvesting and injection so that the aorta’s internal dimensions and texture were preserved. A section of the aorta, starting just distal to the aortic valve, and continuing approximately 80 mm past the ostium of the left subclavian artery was removed at autopsy. Included in the segments were several millimetres of the proximal brachiocephalic, left common
carotid and left subclavian arteries. The arch was fixed in 10% buffered formalin for 24 hours, rinsed and then filled with a silicone casting material, Dow Corning 3110 RTV (Dow Corning, Midland, MI, USA). The silicone was allowed to cure, and the tissue was dissected from the rubber luminal mould. A two-piece outer layer or investment of the luminal mould was then made by lightly coating it with petroleum jelly, and pressing it into silicone dental impression material, Reprosil (Dentsply International, Inc, Milford, DE, USA). After curing, the top half of the investment was formed by pouring more impression material over the embedded mould. This two-piece investment was then filled with a low melting point, water-soluble wax (polyethylene glycol). After cooling, the wax mould was removed. A base layer of clear RTV silicone material, Dow Corning Sylgard 184 (Dow Corning, Midland, MI, USA), was poured into a properly sized acrylic box completely filled with Sylgard. Upon full curing, the wax mould was melted from the Sylgard replica, and the lumen was washed with hot water. The proximal end of the aorta was left closed to simulate a closed aortic valve. The outlets of the three head vessels, and the distal aorta were fitted with tubing for connection to the flow system.

An eight-millimetre hole was made on the anterior surface of the cast two centimetres prior to the origin of the brachiocephalic artery, for test cannula placement. The distal aspect of the cast was attached to capacitance and resistance components previously described,7 then connected to a collection reservoir. A Sarns 8000 pulsatile roller pump (3M Cardiovascular Ann Arbor, MI, USA) and Delphin centrifugal pump (3M Cardiovascular Ann Arbor, MI, USA) were placed in parallel in the test circuit. The raceway tubing in the roller pump raceway was 1/2" internal diameter tubing. An ultrasonic flow sensor (3M Cardiovascular Ann Arbor, MI, USA) was placed in the circuit just beyond the two pumps. The pulsatile pump was adjusted to a pulse rate of 72 cycles per minute with the base flow set at 50% of the peak flow and with the pulse width set at 50% of the cycle. The delphin centrifugal pump was used for continuous flow measurements. The roller pump was used to produce pulsatile conditions for the pulsatile measurements. Beyond the flow probe a Sarns hollow fibre blood oxygenator (3M Cardiovascular, Ann Arbor, MI, USA) was placed in the test circuit. Test cannulae were placed on a five-foot length of tubing attached to the oxygenator outlet. The aortic arch vessels (brachiocephalic, carotid and subclavian arteries) were attached to 1/4" internal diameter tubing adjusted to a height of 76 cm. The circuit was primed with a 33.6% sodium thiocyanate solution in water. Microspheres, five microns in diameter, were added to the circuit to provide a medium for velocity measurement. The solution was circulated until all of the air was displaced from the circuit. The height of the tubing attached to the aortic arch vessels was further adjusted until arch vessel flow was 20% of the total flow from the extracorporeal circuit. The compliance was set by adjusting the cantilever spring to the normal adult range (between 0.67 and 2.0 ml/mmHg pressure). The total resistance was adjusted to 2000 dynes/sec/cm². The ultrasonic blood flow sensor was calibrated according to the manufacturer's recommended procedure. The flow was held constant at 2 l/min ± 0.025 l/min. When corrected for the kinematic viscosity of blood, this relates to a blood flow rate of 6 l/min ± 0.075 l/min.

Fluid velocities were measured using a TSI laser doppler velocimetry system (TSI Incorporated, St Paul, MN, USA) operating in the forward

![Figure 1](https://example.com/figure1.png) Description of the test circuit
scatter mode. Velocity was measured at the point where the two laser beams converged. The measurement volume is approximated by an ellipsoid having a diameter of 50 microns and a length of 650 microns. The system included a frequency shift module which allowed the resolution of bidirectional blood flow and a counter-type signal processor (TSI Model 1980B) connected to the digital I/O ports of a Labmaster data acquisition board. The Labmaster board was mounted in a Hewlett Packard Vectra RS/20C 386 IBM compatible computer which was used for data acquisition and processing. The laser was installed on a platform attached to a computer controlled stepping motor actuated in three axes. The measurement point could be moved in 10-micron increments in each axis. The measuring window was toggled within the aortic cast in 0.5 mm increments, scanning the cast for the area of highest velocity. Velocity was measured over one minute to reduce random noise then average peak values were recorded. This allowed for the acquisition of thousands of measurements that were averaged to yield peak velocity. This technique is extremely accurate since the angle of incidence is fixed, compared with echo Doppler where the angle is estimated for each measurement. Velocity measurements during pulsatile flow were averaged over 72 pulse cycles. To validate the experimental model, four of each aortic test cannulae were tested during continuous flow. One of each of the cannulae was tested under pulsatile conditions for comparison.

Five aortic cannulae, Argyle THI Model 591081 (Sherwood Medical, St Louis, MO, USA), USCI model 1966 24 (CR Bard, Tewksbury, MA, USA), DLP Curved Tip arterial 88024 model (DLP, Inc, Grand Rapids, MI, USA), RMI ARS-024-C (Research Medical Incorporated, Midvale, UT, USA) and Sarns Softflow model 5781 (3M Cardiovascular, Ann Arbor, MT, USA) were selected for testing. All cannulae had an outer diameter of approximately 8 mm and had angled tips. Cannula internal diameter differed according to wall thickness. Four of the test cannulae were angled open tip traditional designs. One of the test cannulae was of a new design having a closed tip and four lateral windows, and flow dispersing cones situated in the blood flow path (Figure 2). This design provides a much larger exit area at the cannula tip. The cross-sectional area of the opening for each cannula is listed on Table 1. Continuous flow peak velocities were analysed for statistical significance using a Student t-test. Statistical significance was accepted at $p < 0.05$.

**Results**

The peak velocity measured for each of the tested cannulae during continuous and pulsatile flow conditions is summarized in Figure 3. Peak velocities averaged 45% higher when pulsatile flow was used. The peak velocity during continuous flow ranged from 3–9.5 times the normal aortic flow velocity (80 cm/second).8 The

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Table 1 Effective cannula exit area (mm$^2$)

<table>
<thead>
<tr>
<th>Cannula</th>
<th>Area (mm$^2$)</th>
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<td>Bard</td>
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</tr>
<tr>
<td>Argyle</td>
<td>29.2</td>
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<td>RMI</td>
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<tr>
<td>Soft Flow</td>
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**Figure 2** Cannula A, closed tip with lateral windows and flow diverting cone; cannula B, open tip design.
peak velocity during pulsatile flow ranged from 6–17 times greater than the normal peak aortic velocity. The cannula with the smallest area of opening had the highest peak velocities. The cannula designed with dispersing cones and larger openings had the lowest peak blood flow velocity during both continuous and pulsatile flow. Peak velocity measurements were found approximately 6 mm from the cannula tip in the test cannula with flow dispersing cones. Open ended cannulae typically had the highest velocity 12 mm from the cannula tip.

Colour flow imaging of the open tip cannula design showed a concentrated jet of flow emanating from the tip directed to the lesser curvature of the aortic arch as demonstrated in Figure 4. The closed tip design with windows and a dispersing cone showed a less concentrated wider dispersion of flow emanating from the cannula that filled the entire lumen of the vessel as demonstrated in Figure 5. The flow patterns were consistent during continuous and pulsatile flow for the two cannula designs, with the exception that the patterns were more pronounced during the peak of the pulsatile phase.

Discussion

Recent published series have shown significantly increased risk of stroke in the patient of advanced age. In a 10-year review of coronary artery bypass patients Gardner reported an overall incidence of stroke of 1.7%.10 The incidence in patients less than 60 years of age was 1.3% versus 6.2% for patients older than 70 years of age. Blauth’s2 extensive series of post mortem examinations of cardiac surgery patients revealed the extent of the problem and prompted many centres to develop strategies for operating on patients with known atherosclerosis of the ascending aorta. Blauth described atheroembolic events as an emerging problem since the incidence increased from 4.5% to 48% in his series from 1982 to 1988.

Embolization of atheroma from the ascending aorta is one of the leading causes of stroke following heart surgery.11 The recent use of intraoperative echo imaging with both transoesophageal and surface probes indicates that the extent of the atherosclerotic changes in the ascending aorta is not always appreciated. Katz12 demonstrated with TEE that 18% of his series of patients older than 65 years had protruding aortic atheromas, 83% of which were not calcific and therefore not detectable by visualization or palpation of the aorta. Measures have been described for reducing the risk of emboli by not clamping or cannulating the ascending aorta in high-risk patients.

Cannula design has traditionally been based on Poiseuille’s Law (where pressure drop is proportional to cannula length divided by the radius4), minimizing the length of the cannula narrowing and expanding the cannula internal diameter by reducing wall thickness. It is generally accepted that a pressure loss or pressure gradient of more than 100 mmHg across the cannula results in excessive formed element damage.10 Brodman5 described a cannula performance index based on the product of the cannula’s outer diameter and pressure loss at a certain flow rate. Shear stress approximations have been used to evaluate cannula fluid dynamics.4 Wall shear stress measurements are greatly influenced by cannula orientation and the direction of blood flow from the cannula with respect to the wall of the vessel. Since cannula orientation is variable in the laboratory and in the clinical settings, shear stress calculations may not accurately reflect the cannula fluid dynamics. Cannula velocity is not influenced by subtle changes in position. Cannulae with high peak velocity would have the potential to produce
**Figure 4** Transoesophageal echo with colour flow imaging of the Bard THI 1966 cannula. Long axis view of the ascending aorta. Note the concentrated high velocity jet at the cannula tip.

**Figure 5** Transoesophageal echo with colour flow imaging of the Sarns Soft Flow closed tip cannula. Long axis view of the ascending aorta. Note the more diffuse and lower velocity flow pattern at the cannula tip.
the highest shear and mechanical impact stress at the vessel wall. While a good argument exists for the use of pulsatile flow during cardiopulmonary bypass, the use of pulsatile flow doubles the velocity at the cannula exit and may present a higher risk of atheroembolism generation in patients with atherosclerotic disease in the ascending aorta. Cannula exit velocity may be a more important consideration than pressure drop.

The exit site velocities of various 8 mm cannulae vary widely. The cannula designed with lateral windows and a dispersing cone produced a lower velocity and less concentrated flow pattern at the cannula exit. High velocity jets at the cannula tip have been implicated as causing the shower of atheroemboli from the wall of the aorta. A low velocity cannula should be considered particularly in older patients, especially when pulsatile flow is employed. Pressure drop has been the primary focus of cannula design and testing. Flow velocity measurement at the cannula exit should be a principal consideration in aortic cannula design and testing.

Acknowledgement

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References

2 Blauth CI, Cosgrove DM, Webb BW et al.