Cardiol Young 2004; 14 (Suppl. 3): 38–43

ISSN 1047-9511

Original Article

The effect of the position of an additional systemic-to-pulmonary shunt on the fluid dynamics of the bidirectional cavo-pulmonary anastomosis

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Keywords: Functionally univentricular heart; cyanotic congenital heart disease; simulation experiments

THE BIDIRECTIONAL CAVO-PULMONARY anastomosis is a well-established palliative procedure for patients with a functionally univentricular circulation. It is usually considered one step in preparation for Fontan procedure, but it may be performed as a long-term palliation for patients deemed to be at high-risk. In this subset of patients, a valuable surgical option could be to add, or maintain, an additional source of flow of blood to the lungs,[1–3] either derived from a patent but banded trunk or one protected by native pulmonary stenosis, or a systemic-to-pulmonary arterial shunt. The risk and benefits of providing an additional source of pulmonary flow after construction of a bidirectional cavopulmonary anastomosis are strongly debated.[4–7] In terms of benefit, the arterial saturation of oxygen is increased due to the greater ratio of pulmonary-to-systemic flow, arteriovenous fistulas are prevented and, as a consequence of the arterial pulsatile flow, the pulmonary arteries are stimulated to grow. The most significant drawbacks are volume overload of the functionally single ventricle, and higher pressures compared to an isolated bidirectional cavopulmonary anastomosis.

It is obvious that, assuming unchanged pulmonary vascular resistances after the bidirectional cavopulmonary anastomosis, the additional source that advantageously increases the flow to the lungs inevitably causes an increase in the pulmonary arterial pressure and, hence, in the superior caval venous pressure. If the additional flow is provided by a systemic-to-pulmonary arterial shunt, a possible solution to limit the augmentation of pressure is to create an arrangement in which the systemic-to-pulmonary shunt works as an ejector. An ejector is a device that uses a fluid jet coursing at high-velocity through an area of low velocity to produce a pumping action. Usually the jet is produced by a fluid, which may be liquid, steam or gas, operating at high pressure, that crosses a small nozzle which converts the pressure into a stream flowing at high-velocity. The pumping action starts since the jet, lowering the pressure in the suction chamber, entrains the fluid in the chamber. The advantage of such an ejector is that the rate of flow on the discharge side is increased, without increasing much the pressure on the inlet side.[8] Figure 1a shows a schematic diagram of a liquid–liquid ejector. Previous studies[9,10] showed that blood entering a systemic-to-pulmonary shunt is accelerated at the proximal anastomosis, producing a high-velocity jet that impacts on the pulmonary arterial wall downstream to the distal anastomosis. We suppose that, if the impingement of the jet and the consequent deceleration are avoided, the ejector effect may be enhanced, and pulmonary flow could increase without producing a corresponding increase in superior caval venous pressure.

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Accepted for publication July 2003

https://doi.org/10.1017/S1047951104006547 Published online by Cambridge University Press

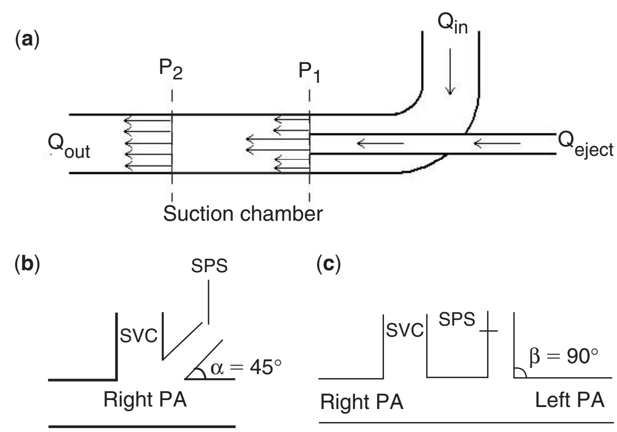

Figure 1.
*(**a**) Sketch of a liquid–liquid ejector. A high-pressure jet (Q_{eject}) crosses a small nozzle that converts the pressure into a high-velocity stream and entrains the fluid (Q_{in}) in the suction chamber. The flow rate on the discharge side (Q_{out}) increases without increasing much the pressure on the inlet side ($P_1 < P_2$). Schematic drawings of the two systemic-to-pulmonary shunt (SPS) configurations: (**b**) angled ($\alpha = 45°$) and (**c**) parallel ($\beta = 90°$). PA: pulmonary artery; SVC: superior caval vein.*

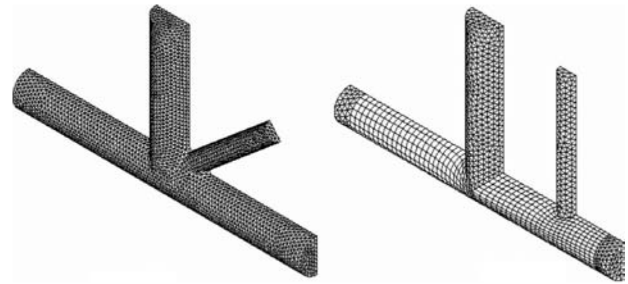

Figure 2.
Three-dimensional computational models of the angled and parallel configurations for the shunt. Due to the symmetry of the models, only one half has been meshed in order to reduce computational time.

The objective of our study, therefore, was to investigate the fluid dynamics and possible ejector effect in models of two shunts with different geometries (Fig. 1b and c), since the position and the angle of the systemic-to-pulmonary shunt may have an influence on the efficiency of the ejector. We used approaches based on both in vitro experiments and analysis of computational fluid dynamics.

Materials and methods

Experimental study

We investigated two different plastic models of a systemic-to-pulmonary arterial shunt, specifically with the shunt connected directly at an angle of 45° to the left side of the anastomosis between the superior caval vein and the right pulmonary artery (Fig. 1b), and one connected perpendicularly to the left pulmonary artery at a distance of 18 millimetres from the anastomosis between the superior caval vein and the pulmonary artery (Fig. 1c). In both models, the inner diameter and length of the shunts were 5 and 35 millimetres, respectively. The superior caval vein and the pulmonary arteries have inner diameters of 10 millimetres.

The plastic models were inserted in a purpose designed mock-loop simulator described in detail by Kull et al.[11] Briefly, the simulator permits the imposition of different rates of flow through the shunt, along with different superior caval venous and pulmonary resistances. Pulmonary resistances consist of a fixed term, simulating the pulmonary arteries, and a adjustable one, simulating the vascular bed. Different pressures and flows may be measured by piezometers or ultrasonic flowmeters. In this way, we measured pressures in the right pulmonary artery, at the shunt inlet, in the atrial reservoir and in the windkessel, mimicking the pulmonary compliance along with rates of flow in the superior caval vein and in the right pulmonary artery.

The experiments were conducted at room temperature. Water was used as the working fluid. During all tests, the level in the atrial reservoir was kept constant to guarantee an atrial pressure equal to 3 millimetres of mercury. For superior caval venous pressures, we used values about 10, 15 and 20 millimetres of mercury. A centrifugal pump supplied the flow through the systemic-to-pulmonary shunt at a steady-state ranging from 0 to 1 litre per minute, in steps of 0.25 litre per minute. We investigated values of about 3 and 9 millimetres of mercury per minute per litre for the pulmonary resistances. We analysed the results in terms of total pulmonary flow, the proportional contribute of superior caval venous flow to total venous flow, and the ratios of flows to the right and left lungs.

Computational study

Simulations[12–14] were performed to reproduce the experimental tests and improve the understanding of the local fluid dynamics in the models. Two finite volume three-dimensional models (Fig. 2) with the same dimensions of the plastic models were created and solved with a commercial code (Fluent 6.0, Fluent Inc., Lebanon, New Hampshire, USA). In the angled configuration, it was necessary to round off the corners between the caval vein, the shunt, and the pulmonary arteries in order to recreate the geometry with proper elements for the finite analysis of volumes. This slight change creates a distal cross-sectional area

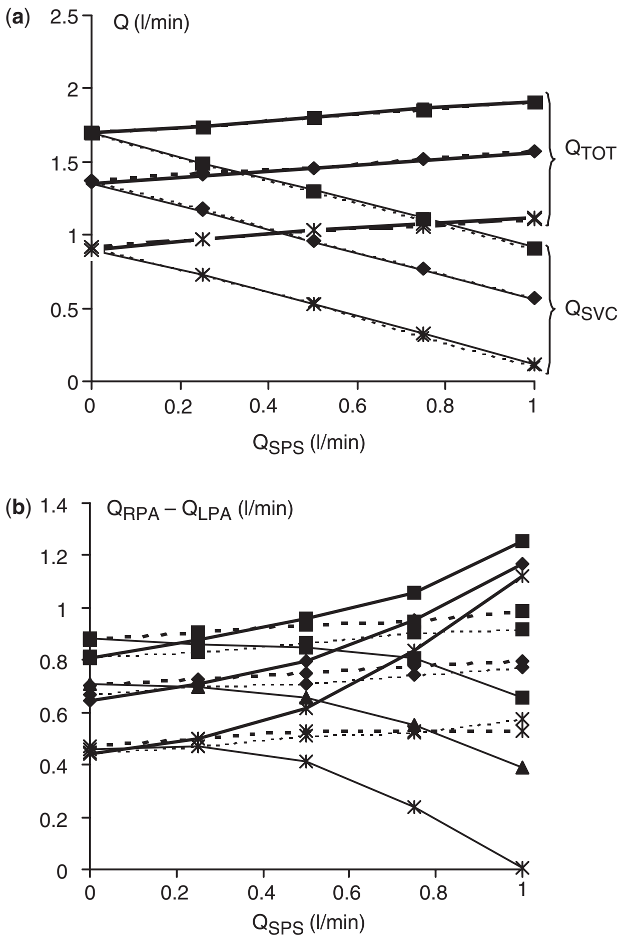

Figure 3.
*Results of experimental tests at the lower values of pulmonary resistance. (**a**) Superior caval venous (Q_{SVC}, thin line) and pulmonary (Q_{TOT}, thick line) flows versus flow through the shunt (Q_{SPS}), for the angled (full line) and the parallel shunt (dashed line) as a function of superior caval venous pressure. (**b**) Flow in the left pulmonary artery (LPA, thin line) and right pulmonary artery (RPA, thick line) pulmonary arteries versus flow across the shunt for the angled (full line) and the parallel shunt (dashed line) (✱, ◆, ■ indicate superior caval venous pressures of 10, 15 and 20 millimetres of mercury).*

of the systemic-to-pulmonary shunt that is larger than in the in vitro experiments. Pressure contours, particle pathlines, velocity vectors, and distributions of flow were evaluated in both the models.

Results

In Figure 3a, we show the influence of the flow through the systemic-to-pulmonary shunt on the flow in the superior caval vein and on the total pulmonary flow for the parallel and angled shunts at the different caval venous pressures but with the lower value for pulmonary resistance. The curves are almost identical for the two models.

As expected, higher venous flows correspond with higher pressures. Furthermore, with increasing flow through the shunt, superior caval venous flow decreases almost linearly, while total pulmonary flow increases linearly at every tested value for pressure. When flow across the shunt passes from 0 to 1.0 litre per minute, the proportion of superior caval venous flow decreases from 100% to about 48%, 36% and 10% for pressures equal to 20, 15 and 10 millimetres of mercury, respectively, for both models.

Figure 3b shows the influence of flow through the shunt on the distribution of flow in the pulmonary arteries for the two models with the lower resistance at every value of pressure tested. In the angled configuration, the flow in the right pulmonary artery increases concomitant with increase of the flow through the shunt, while the flow in the left pulmonary artery decreases at every caval venous pressure. In the parallel configuration, the distribution between the right pulmonary artery and the left pulmonary artery is almost uniform and constant for different flows through the shunt and pressures in the superior caval vein. Similar trends were observed with lower values of the measured pulmonary venous, superior caval venous, and left and right pulmonary and arterial flows versus the flow across the shunt and superior caval venous pressure in both of the shunt models when the pulmonary resistances were set to the higher value. At pressures of 10 and 15 millimetres of mercury, the percentage of flow in the superior caval vein became negative until −0.52, because of retrograde venous flow. In the angled configuration at the same pressures, the ratio of flow to the left and right lungs showed negative values even until −0.41, indicating a flow entering the model through the left pulmonary artery. In contrast, values around unity were measured for the parallel configuration as seen for lower resistances.

Some results of our computational simulations are shown in Figure 4. The figure illustrates the pathlines of various virtual particles inserted at the superior caval vein and the inlet to the systemic-to-pulmonary shunt. Different colours indicate the different particles. Simulations are shown for the angled and the parallel configurations. The computational models correctly predicted the measured flows in the superior caval vein and the pulmonary flow, along with the ratio of flows in the lungs, apart from the situation for the simulated condition in the angled configuration with a flow through the shunt of 1.0 litre per minute, when a higher flow was calculated through the left pulmonary artery.

Discussion

According to our experimental results (Fig. 3a), no significant differences occur in the various studies

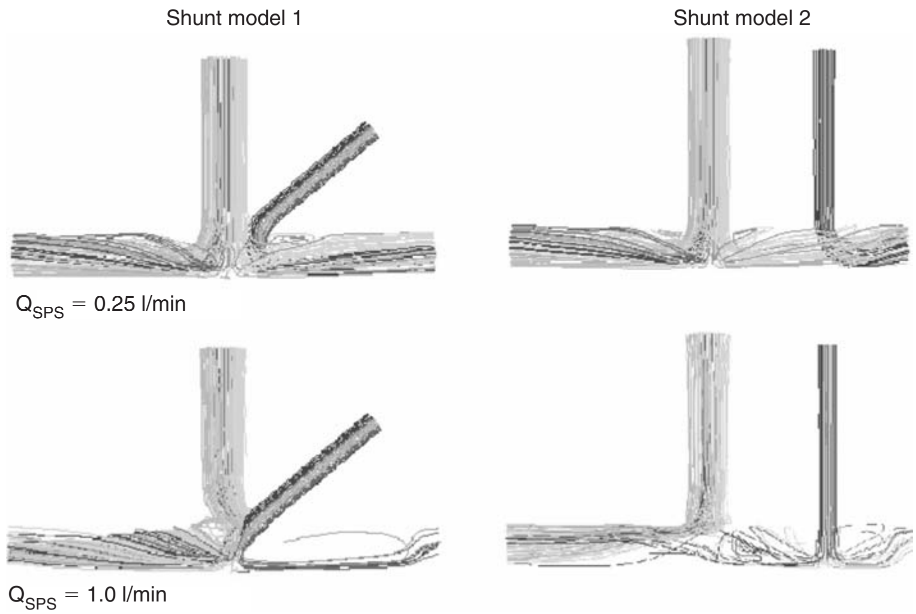

Figure 4.
Results of the computational fluid-dynamic models. The paths of particles are shown for the angled and the parallel shunts at the two rates of flow across the systemic-to-pulmonary shunts (Q_{SPS}). The different colours represent the different particles entering the superior caval venous and the systemic-to-pulmonary shunt.

flows for either the angled or parallel configurations. Different angles and positions of the systemic-to-pulmonary shunt for the conditions of testing, therefore, do not affect the flow from the superior caval vein at the investigated values through the shunts. The decrease in the flow through the superior caval vein associated with increasing flow across the shunt indicates that the two flows are in competetion. In any case, the very low values measured for superior caval venous flow are a consequence of the fixed pressures at the inlet to the superior caval vein during the experimental tests. The in vivo situation probably corresponds to smaller decreases of caval venuos flow coupled with increases of caval venous pressure.[5,6]

The configuration of the shunt, nonetheless, does influence the partition of flow for high values across the shunt of greater than 0.5 litre per minute (Fig. 3b). In the parallel model, flow through both pulmonary arteries increases slowly with an increase in flow through the shunt whereas in the angled model, an increase in flow through the shunt results in a marked increase of flow in the right, but a decrease of flow in the left pulmonary arteries. At pressures of 10 and 15 millimetres of mercury in the superior caval vein, flow in the left pulmonary artery even became negative at high pulmonary resistances as a consequence of non-physiological recirculation of flow through the pipes mimicking the pulmonary arteries. Indeed, for the sake of simplicity of the mock-loop simulator, the resistances of both lungs were lumped in a single, adjustable, component located downstream to a bifurcation of the pulmonary arteries.[11] Hence, for increased pulmonary resistances, the model preferred to recirculate flow into the pulmonary arteries instead of towards the atrial reservoir.

The data obtained with the computational simulations show a good agreement with the experimental results, since the trend of the investigated variables was correctly produced with respect to flow across the shunt. The distribution of flow for values of 0.25 litre per minute across the shunt was approximately equal for the left and right pulmonary arteries. The pathlines for the panels (Fig. 4, top panels) show that, in both models, the flow through the shunt is likely to go to the left pulmonary artery, while most of the superior caval venous flow perfuses the right pulmonary artery. This behaviour is in accord with previous clinical angiographic investigations.[1] The effect of the angle of the shunt is lacking due to the low velocity of flow interdicting the stream that crosses from the superior caval venous and perfuses the pipe representing the right pulmonary artery. Conversely, at flows, of 1 litre per minute, the flow in the angled configuration goes more to the right than to the left pulmonary artery, with no particle from the superior caval venous flow entering the left pulmonary artery (Fig. 4, left bottom panel). This produces marked inequality in the distribution of flows between the pulmonary arteries with the angled configuration. In contrast, the flow both from the superior caval vein and across the systemic-to-pulmonary arterial shunt supply both pulmonary arteries with the parallel arrangement (Fig. 4, right bottom panel).

The results of simulation for the angled model at shunt flows of 1 litre per minute differ quantitatively from the experimental results. This is compatible with the difference between the true geometry and the model mesh as described in Material and methods, since the outlet of the shunt was a little larger in the computational model than in the plastic one. Hence, velocities of the streaming jet simulated from the systemic-to-pulmonary shunt were lower, producing less flow to the right pulmonary artery compared to the experimental measurements.

We must emphasise four major drawbacks of the present study. First, the mock-loop simulator needs partially re-designing in order to obtain a more physiological splitting of the resistances in the two pulmonary arteries. Second, the working fluid used in the steady-state experiments was water instead of blood. Third, we performed only steady-state experiments and simulations. Pulsatile conditions could affect local fluid dynamics, and different behaviours could occur during the diastolic and systolic periods. Fourth, we used rigid walls as models of the bidirectional cavo-pulmonary anastomosis and shunt, while both the native vessel and the synthetic shunt have distensible walls.

Concerning the first point, our computational model allows us to make some inferences concerning the repartition of flow with high-pulmonary resistances correctly split. We performed preliminary computational steady-state simulations with higher resistances in the branches of the pulmonary trunk. The results show the same trend for the total and the superior caval venous flows versus the flows through the shunt as for the lower resistances, but the distribution of flow is now uniform in both models. This suggests that, at low pulmonary resistances, the local geometry of the shunt plays a role in determining the ratio of flow to the different lungs, whereas at higher values, the ratio is influenced only by vascular resistance.

As regards the working fluid, we expect that the use of blood instead of water will further decrease the ejector effect. Indeed, the higher viscosity should decrease the role of the local hemodynamics with respect to that of viscous resistances such as the vascular pulmonary resistances. Further investigations are in progress to assess the possible influence of distensibility and pulsatility on the local fluid dynamics.

In conclusion, our study has shown that, at the examined conditions and geometries, no evident ejector effect occurs in a bidirectional cavo-pulmonary anastomosis with an additional systemic-to-pulmonary shunt. This is due to the velocities of streaming of the systemic-to-pulmonary shunt being too low when compared with the velocities in the "suction chamber" (Fig. 1a). According to the computational model, the maximum velocity in the systemic-to-pulmonary shunt was lower than 100 centimetres per second coherent with an inner diameter of 5 millimetres and a rate of flow of 1 litre per minute. In vivo Doppler measurements,[10,15] in contrast, had indicated higher cycle-averaged velocities of up to 400 centimetres per second in the shunt, both due to high rates of flow, or smaller diameters, and skewing of the velocity profile at the proximal anastomosis.[10,16] Hence, a greater ejector effect should occur in vivo, and may explain the relative low superior caval venous pressures observed in some studies.[1] Furthermore, idealised geometries were adopted for both the plastic and computational models. In reality, the angles between the superior caval venous and systemic-to-pulmonary shunt and the right pulmonary artery could be different, with the angle of the shunt larger than 45°, or the superior caval venous not being perpendicular, but we believe that, in these conditions the ejector effect would not be significantly improved.

Acknowledgements

We are grateful to Marcia Emmer and Mjriam P. Rubbens, from the Technical University of Eindhoven, and Luigi Baglioni from the Politecnico di Milano, for excellent technical help in performing the mechanical tests.

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