IMAGES in Paediatric Cardiology

Invited article **Migliavacca F*, Dubini G**, de Leval MR***. Computational fluid dynamics in paediatric cardiac surgery. Images Paediatr Cardiol 2000;2:11-25**

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MeSH

Computational fluid dynamics Mathematical model Cavopulmonary connections Systemic-to-pulmonary shunt

Abstract

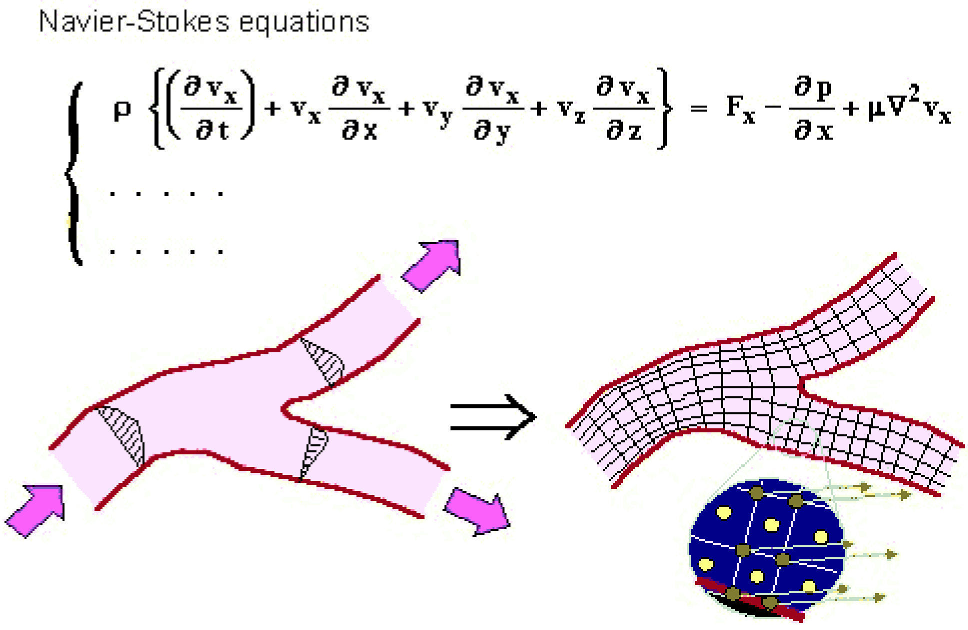
Computational fluid dynamics techniques have been applied to study both the local and the global haemodynamics created by different surgical reconstructions, currently used to treat complex congenital heart defects. These operations are characterised by competition of flows which can lead to postoperative failure of the surgical treatment. Different techniques have been used in order to improve knowledge of the global haemodynamics in patients submitted to such operations, and to devise possible optimal hydraulic designs of the connections. The adopted approach has combined highly-detailed, three-dimensional models of the connections with simplified zero-dimensional, lumped-parameter network models of the overall circulation of the patient. Three-dimensional models of the connections have been developed by means of the finite element method. Local fluid dynamics features have been analysed and then ‘incorporated’ in mathematical models able to predict some clinically relevant postoperative haemodynamic data. Results emphasise the impact of local geometry on global haemodynamics.

Article

Introduction

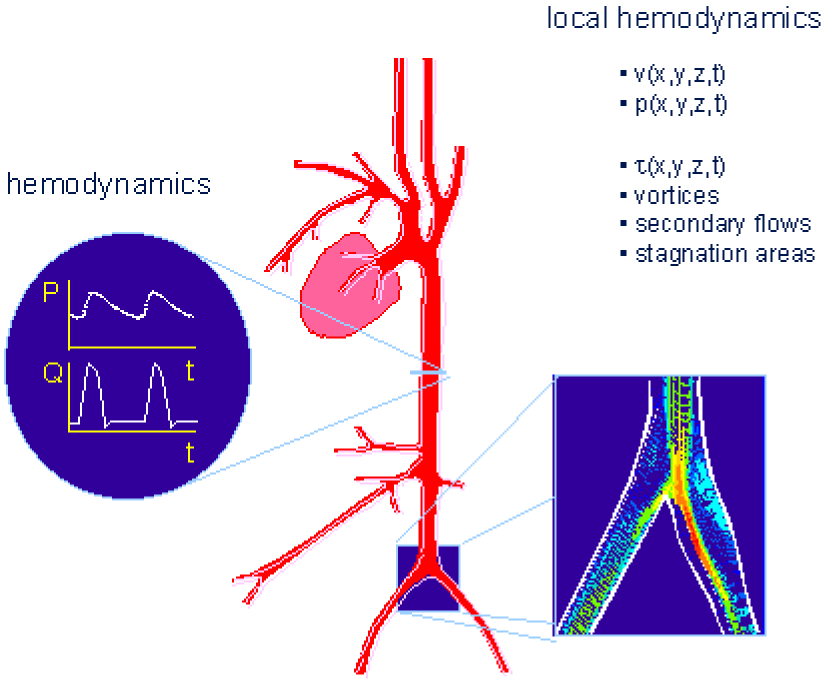
Computational fluid dynamics (CFD) techniques are among the most powerful tools available to the engineering branches dealing with the motion of fluids and exchange of mass, momentum and energy. They have recently facilitated an increasing number of applications to the human cardiovascular system but few studies have focused their attention on the fluid dynamics of the surgical reconstruction of congenitally malformed parts of the cardiovascular system. In vitro models are the alternative laboratory tools to study fluid dynamics. Advantages of CFD over in vitro models are the easy quantification of haemodynamic variables (such as flow rate, pressure, shear stress distribution) and changes in geometric and fluid dynamics parameters. Furthermore, CFD methods allow the development of 3-D models able to reproduce both the complex anatomy of the investigated region and the details of the surgical reconstruction, especially with the recent developments in magnetic resonance imaging. Basically, CFD models enable one to obtain the solution of the non-linear equations governing the motion of blood in vessels with proper boundary and initial conditions (Fig.1). CFD models have obviously some limitations and the main still lies in the knowledge of the exact boundary conditions to impose at the inlets and outlets.

Figure 1. Computational techniques provide a solution of the Navier-Stokes equations



On the basis of the results, quantitative evaluation of the surgical correction can be made. This technology, which benefits greatly from the continuous improvement in hardware and software resources, enables cardiovascular experts and bioengineers to look at fluid dynamics behaviour of cardiovascular regions with increasing sophistication (Fig.2).

Figure 2. Electric analogue models of the circulation allow a description of temporal variables, such as flow and pressure tracings (left inset), in different portions of the cardiovascular system. Local haemodynamics (i.e. shear stresses, local pressures and velocities, stagnation and recirculation areas) can be studied with more sophisticated, 2-D or 3-D models (right inset)



Surgery for treatment of congenital heart diseases has improved very rapidly in recent years, as well. This can be attributed to evolutions in preoperative diagnosis of congenital heart defects, the development of new surgical techniques, and improvements in postoperative management. Surgical repair of complex heart defects may impose major reconstructive procedures, creating a totally new circulation. The most striking examples are the right heart by-pass operations (Fontan circulation)[1] used to treat a variety of complex heart defects such as hypoplastic ventricle, tricuspid atresia and univentricular hearts. These congenital defects share the common characteristic that there is functionally only a single ventricular chamber.

Herewith, CFD methodologies are applied to study the fluid dynamics created by three currently used surgical reconstructions: i) the systemic-to-pulmonary shunt or modified Blalock-Taussig shunt (MBTS), ii) the bidirectional cavopulmonary anastomosis (BCPA) and iii) the total cavopulmonary connection (TCPC).

The MBTS is used as first stage palliation in the treatment of hypoplastic left heart syndrome. It consists in using the pulmonary valve as a systemic ventricular outlet and constructing a systemic-to-pulmonary artery shunt to provide pulmonary blood flow. The BCPA lies in connecting the superior vena cava directly to the right pulmonary artery. In the TCPC, a more recent development, in addition to the superior vena cava, the inferior vena cava is connected to the right pulmonary artery, too (by means of an intraatrial tunnel). From a fluid dynamics viewpoint, the surgical repair or reconstruction should not cause blood flow abnormalities (turbulence, vortices, high shear stresses, flow separation, recirculation and stagnation areas) which may induce increased resistance, wall damage, thrombus formation, as well as energy dissipation. In addition, the surgical repair should supply the lungs with a proper blood flow, to guarantee adequate blood oxygenation.

Quantitative information gathered from CFD simulations of the cardiovascular districts involved could be used to demonstrate the potential for abnormal fluid dynamics behaviour, and suggest alternative haemodynamic designs that could be applied and validated in clinical practice. A refinement of surgical procedures could lead to further improvements in clinical results. However, it is important to underline that mathematical models should provoke, predict and not take the place of new experimental observations. Recent literature on modelling fluid dynamics of paediatric surgical procedures.

In vitro models

One of the first in vitro experiments was performed by de Leval et al[2], who reported visualisation of flow through cavities and around corners and measurements of energy losses across non-pulsatile cavities, corners, and stenoses. Those studies indicated the importance of streamlining and suggested a way in which hydrodynamic designs of the Fontan circulation might be improved: the total cavopulmonary connection. In the following years researchers studied the hydrodynamics of this new connection and compared it with the atriopulmonary connection. Low and colleagues[3] in 1993 showed that the cavopulmonary connection was found to have much lower flow losses compared to the atriopulmonary one. The study by Kim et al[4] in 1995 reached the same conclusions. Sharma et al[5] in 1996 sought to evaluate the effect of offsetting cavopulmonary connections at varying pulmonary flow ratios to determine the optimal geometry of the connection.

Recently Lardo et al[6] studied hydrodynamic efficiency between intra-atrial lateral tunnel, extracardiac tunnel and extracardiac conduit, with or without caval vein offset performed on explanted sheep heart preparations and in an *in vitro* flow loop. As regards the bidirectional cavopulmonary anastomosis as far as we know only two work have recently appeared. Sievers et al[7] proposed a modification of the standard Norwood variant of cavopulmonary connection with an extended anastomosis based on hydrodynamic results of two different in vitro models. Lardo et al[8] showed with an *in vitro* model that a bidirectional cavopulmonary anastomosis reduces fluid-energy dissipation in atriopulmonary connections, provides a physiologic distribution of total flow, and maintains some hepatic venous flow to each lung.

Computational models

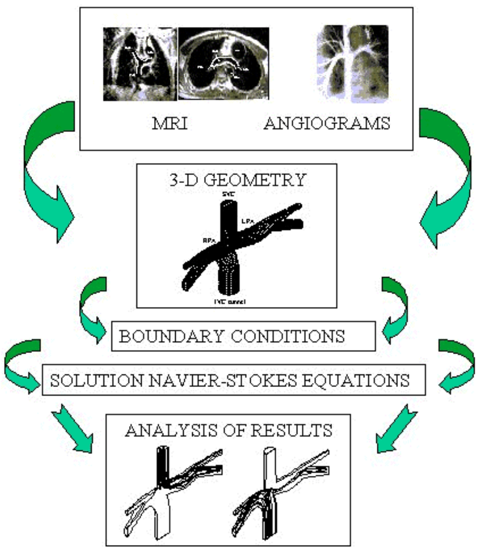
Computer generated models, based on finite element method, showed lower energy losses with the cavopulmonary than with the atriopulmonary connection[9]. Asymmetry of the superior and inferior vena caval connection was demonstrated to improve the energetics[10,11]. Local effects of pulsatile forward flow from the native pulmonary artery on blood flow repartition to the lungs were investigated as well[12,13]. Barnea et al[14,15] and Santamore and colleagues[16] developed theoretical analysis to determine the effects of blood flow distribution between pulmonary and systemic circulations on oxygen delivery in the treatment of hypoplastic left heart syndrome. Recently Pennati et al[17,18] presented a lumped parameter model of the circulation, which can represent both the preoperative and the postoperative (systemic and pulmonary) circulations in a patient with a double-outlet univentricular heart.

Mathematical models

3-D models

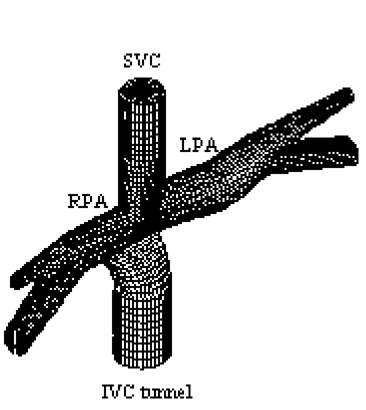
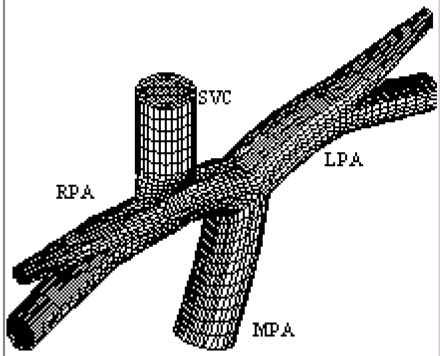
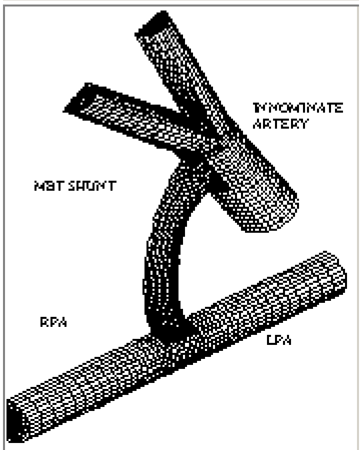
Fluid dynamics in great arteries can be described by equations of mass and momentum conservation (Navier-Stokes equations) when the non Newtonian features of blood can be ignored[19]. Basically, these are partial derivative equations, which can be analytically solved only for simple cases when the boundary conditions are properly set. CFD techniques, among which the finite element method, allow one to solve the fluid dynamic field in most of the cases where the analytical solution cannot be achieved. The first step in creating a three-dimensional CFD model is to reproduce the geometry of the investigated region and to divide a continuum in a number of simple element ('bricks') where the unknowns of the problem (pressure and velocity) will be evaluated. Geometric data are kept from angiograms, MR images, Doppler measurements, etc. Imposition of the boundary conditions (i.e. velocity and pressures at the inlets and outlets of the model) is the second step. The mathematical code adopted then will calculate the fluid dynamic field. The last step is the analysis of the results (Fig .3).

Figure 3. Methodological steps for the construction and analysis of fluid dynamics in 3-D models:
1) reproduction of the geometry of the investigated region
2) division of fluid domain continuum in a number of simple elements ('bricks')
3) imposition of the boundary conditions at the inlets and outlets
4) calculation of the fluid dynamic field
5) analysis of the results



Three dimensional models of several operations are shown in fig 4. 3-D models allow one to describe in great detail fluid dynamics in specific portions of the cardiovascular district. Effects of these features on global haemodynamics such as cardiac output and pulmonary-to-systemic flow ratio, for example, cannot be described with these models. This kind of information can be achieved with lumped parameter models or electrical analogue.

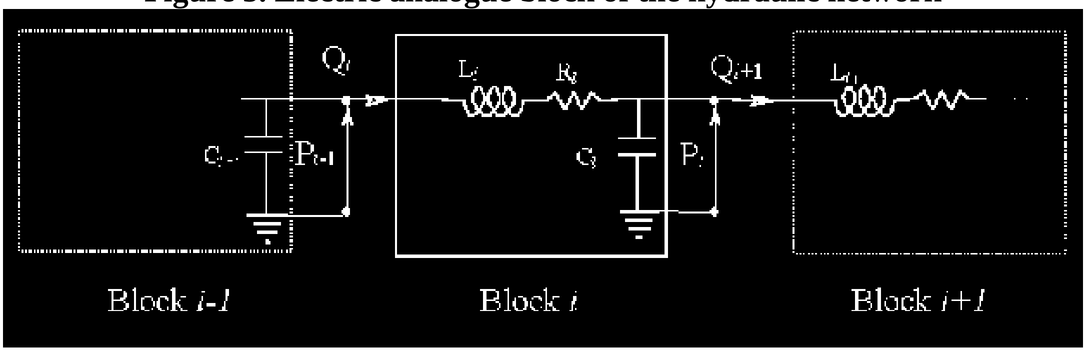
Figure 4. Finite element models of the total cavopulmonary connection (TCPC), the bidirectional cavopulmonary anastomosis (BCPA) and the modified Blalock-Taussig Shunt (MBTS)

Total cavopulmonary connection	Bidirectional cavopulmonary anastomosis	Modified Blalock-Taussig Shunt
		

Lumped parameter models

In the, literature these models can be easily found[20-23]. They have in common the fact that the circulatory system is modelled as a hydraulic network composed of resistance, inertance and compliance elements as well as non-linear resistance components that incorporate energy losses at the connections. Basically, an analogy between electricity and haemodynamics is adopted (Fig. 5). If pressure gradient is analogous to voltage, flow to current, compliance to capacitance, inertance to inductance and hydraulic resistance to electrical resistance, equations of electrical network can be applied and the effects of any changes in the system parameters calculated[24].

Figure 5. Electric analogue block of the hydraulic network



In order to achieve the required level of detail in the region where the surgical correction is performed, results from previously described three-dimensional CFD models should be included. Local haemodynamics are heavily affected by inertial effects as well as three-dimensional details of the surgical repair. Hence, apposite flow related lumped parameters should be added to the whole network[17]. These models represent the whole circulatory system as a closed loop. This allows one to evaluate the mutual interactions among all the involved haemodynamic variables. The major drawback of this approach is the necessity of using many parameters to characterise the resistance, the inertance and the compliance of the districts of the hydraulic network. Their evaluation turned out to be quite a troublesome task, due to the impossibility of measuring those data during clinical investigations and procedures, except for the pulmonary arteriolar resistance and the systemic vascular resistance. Indeed, even in literature the great majority of data comes from *in vivo* measurements on animals[25,26]. Nevertheless mathematical model studies of human arterial systemic circulation[20,27-29], of the venous systemic circulation[30,31], the pulmonary arterial circulation[23,32,33] as well as of the whole cardiovascular system[34] supply a limited amount of data and references at various levels of detail. Unfortunately, the majority of them refers to normal adults. Our approach was based on the adaptation of an existing hydraulic lumped-parameter model of the fetal circulation[35] already tested on pregnant women according to a protocol supported by extensive Doppler velocimetry measurements.

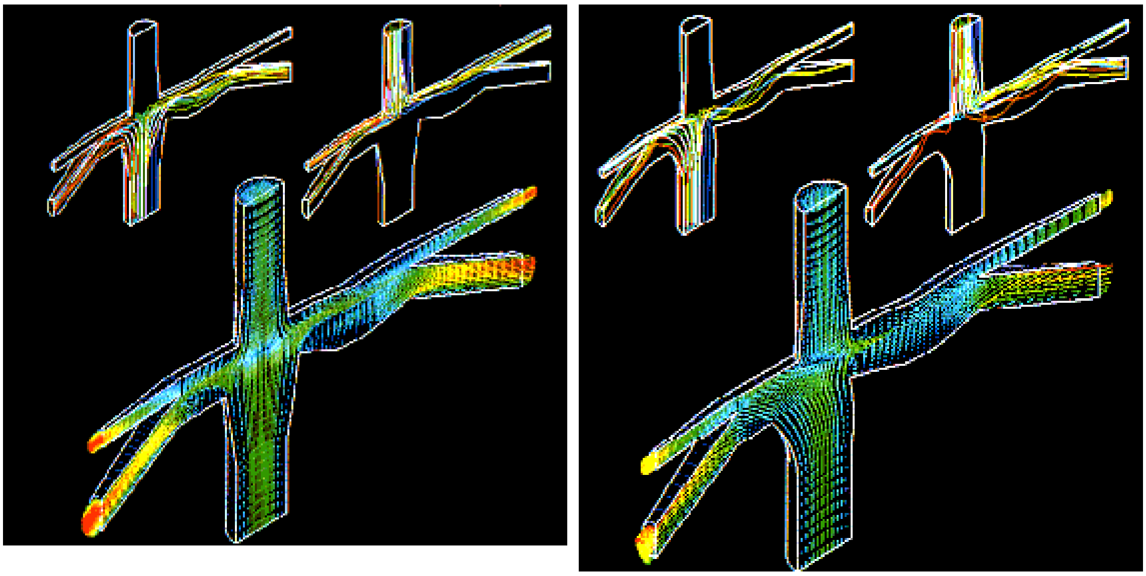
Three examples of coupling 3-D and lumped parameter models

In the following examples of TCPC, BCPA and MBTS models linked with lumped parameter model of the whole circulation (or the pulmonary circulation for he case of TCPC) are shown. Conceptually the TCPC coupled model is simpler than the others. Indeed, aim of this connection is to minimise the energy dissipation at the anastomoses and a local haemodynamics description is sufficient to address the clinical question. BCPA and MBTS models have different clinical questions. The former aims to know the quantity of forward flow from the native pulmonary artery to leave at the time of surgery in order to have a good blood lung perfusion avoiding caval hypertension; the latter aims to know which is the best size of the interposition shunt to have a satisfactory blood lung perfusion. For such models, local haemodynamic description (i.e. only 3-D models) is unsatisfactory and a global description of the whole cardiovascular system by means of a lumped parameter model is necessary.

TCPC

3-D models have been previously described[10,11]. An example of the discretised model was shown in Figs.3 and 4. Those studies showed that the best configuration to have minimal energy losses with optimal flow distribution between the two lungs was obtained by enlarging the inferior vena caval anastomosis towards the right pulmonary artery. Pulmonary resistances play an important role in blood flow distribution to the lungs. To investigate their effects a simple mechanical lumped parameter model was added to the 3-D model. The two lungs and the left atrium represented it. Only resistance components appeared in that model as the simulations were performed with the assumptions of steady flow. Data of the parameter were taken from catheterisation. Velocity vector plots and particle paths of two different configurations are reported in Fig.6. A collision of the two caval streams is present when the two veins are aligned. On the contrary, when an offset towards the right lung is present, most of the inferior caval flow is diverted towards the right pulmonary artery. This findings suggest to modified the surgical procedure[11].

Figure 6. Velocity vector plots and particle paths in two different models of TCPC: no offset between caval vein (top) and right insertion of inferior anastomosis (bottom)

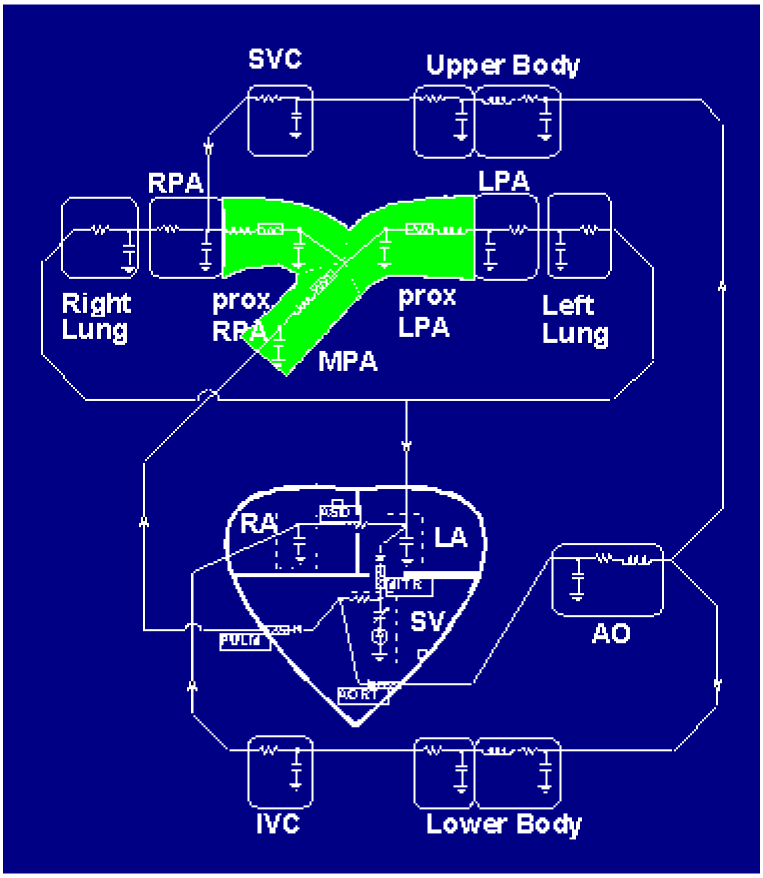


BPCA

In Fig. 7 the case of the BCPA circulation[17,18] is depicted as example. The systemic circulation consists of an RCL block representing the ascending aorta and the two parallel upper and lower body branches. Each parallel branch is assumed to be divided into three stages, which represent the arterial compartments, the venous compartments of the upper and the lower body and the venae cavae, respectively. The upper body branch is connected with the pulmonary circulation by means of an end-to-side anastomosis between the superior vena cava and the right pulmonary artery, while the lower body branch is connected to the heart with a compliance element representing the right atrium. The modelled pulmonary circulation looks rather similar to the systemic one, but it is conceptually different. More precise detail is required at the level of pulmonary arteries in order to incorporate properly the effects due to the surgical repair. A RC block (right and left lung) representing the right and left lungs with their veins completes each pulmonary branch, which eventually delivers blood into the left atrium.

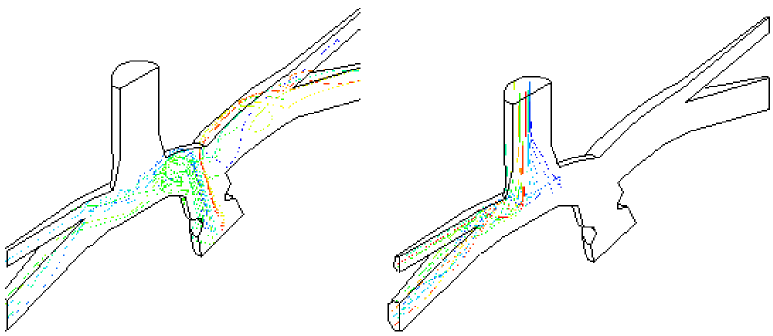
In some cases (MPA, pRPA, pLPA) the dissipative term (the resistance) is also related to the local haemodynamics and depends on the volume flow rate. 3-D models have been previously constructed and effects of different pulmonary forward flows, of presence of tubular or discrete stenosis as well as of inclination of native pulmonary trunk have been taken into considerations[12,13].

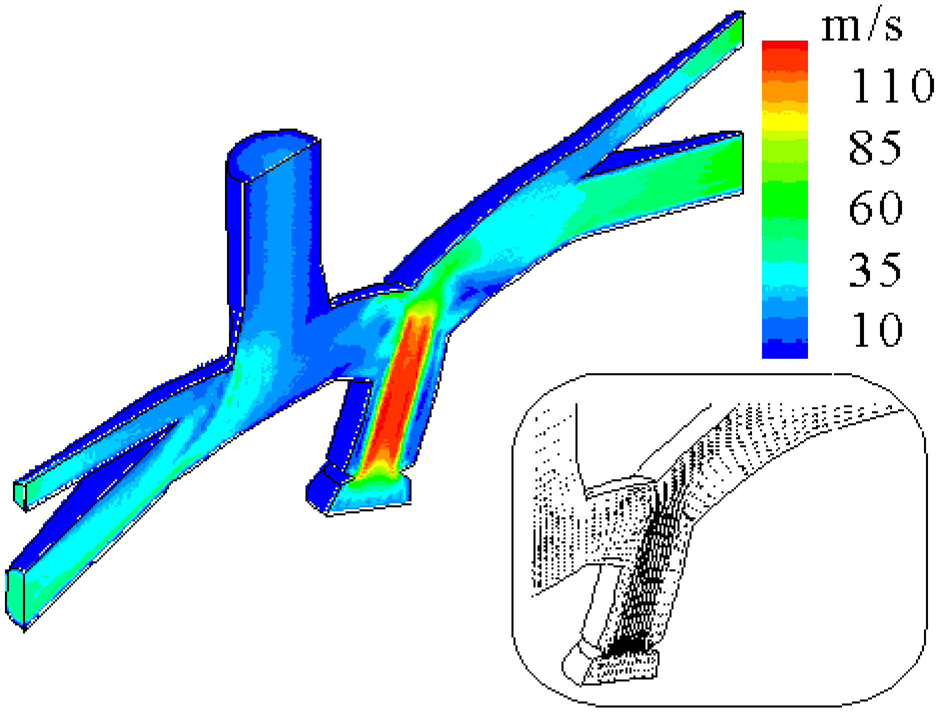
Figure 7. Lumped parameter model of BCPA circulations. Green area includes results of 3-D models



In Fig. 8 velocity contours and particle paths in different models of BCPA are reported as examples. 3-D models of BCPA emphasise the impact of local geometry on fluid dynamics[11]. Results of lumped parameter model, which are fairly in agreement with clinical data, suggest that a proper regulation of the pulmonary outflow obstruction could guarantee adequate postoperative haemodynamics at each value of pulmonary arteriolar resistance[17,18].

Figure 8. Velocity contour and vectors at systolic peak and particle paths in BCPA models with native stenosed pulmonary artery pointing towards the left (at left) or the right (at right) lung



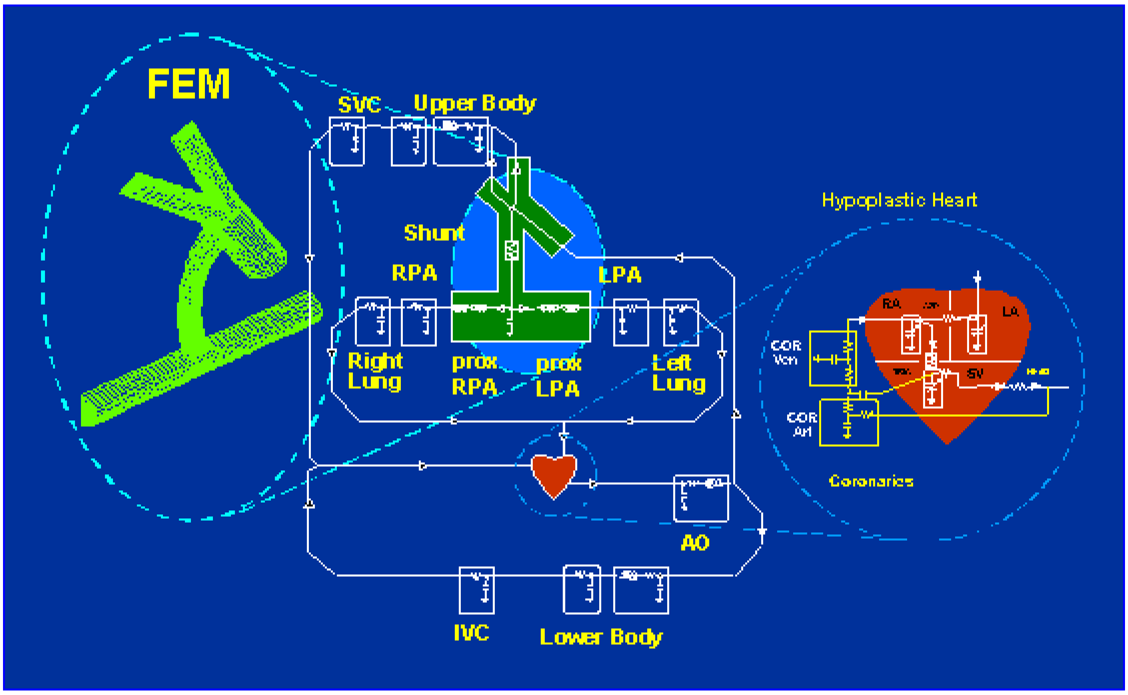


MBTS

As for the BCPA, a lumped parameter model was developed for the MBTS circulation (Fig. 9). Using allometric relations, some parameters of the previous model were scaled for a lower body surface area.

Hypoplastic left heart was modeled and effects of size of the shunt on local fluid dynamics were investigated by means of the 3-D model. Skewed velocity profiles in the proximal part of the shunt were demonstrated (Fig.10)[36] as well as the pressure-flow relationship in the shunt (Fig.11).[37] Effects of local haemodynamics in the parameters representing the shunt and the proximal pulmonary arteries were included as for the BCPA model (shaded area in Fig. 9). The lumped parameter model allowed one to study the effects of changes in pulmonary and vascular resistances on fluid dynamics. Figure 12 shows changes in cardiac output index and pulmonary-to-systemic flow ratio (Q_P/Q_S) as function of shunt diameter in a simulation for a hypothetical patient with a body surface area of 0.33 m^2, heart rate of 120 beats per minute, pulmonary and vascular resistance of 2.3 and 21.9 Woods·m^2, respectively. Influence of systemic vascular resistance on the two previous quantities is reported as well for a shunt size of 3.5 mm.

Figure 9. Lumped parameter model of MBTS circulations. Dark green area includes results of 3-D models



Validation of computational models

Validation and set-up of computational models requires data which can be obtained with different methodologies either of clinical routine, such as Doppler velocimetry, catheterisation exams, or not, as magnetic resonance imaging exams or in vitro experiments. For the studies presented, clinical data obtained from catheterisation were utilised to construct the 3-D models and set-up the parameters of the lumped models. An in vitro model of the MBTS circulation was appositely designed to test the reliability of CFD computations.[38] Comparison between the two methodologies was quite satisfactory with difference smaller than 10%. Magnetic resonance imaging was utilised on a patient previously submitted to TCPC. CFD simulations, with the geometry obtained from the MR scan, were performed and good agreement in results with in vivo data was obtained.[39]

Conclusions

Our approach combined highly-detailed, three-dimensional models of the connections with simplified zero-dimensional, lumped-parameter network models of the overall circulation of the patient. Local fluid dynamics features of the connections were analysed and then 'incorporated' in mathematical models able to predict some clinically relevant postoperative haemodynamic data. This led to the quantification of blood flow distribution into the lungs and between the systemic and pulmonary circulation in the bidirectional cavopulmonary anastomosis[18] as well as the modified Blalock-Taussig shunt.[36,37] Results from the total cavopulmonary connection simulations also suggested to modify the surgical technique by means of the insertion of a lateral patch in the right side of the inferior anastomosis.[2]

Figure 10. Velocity contour speed plots at four different instants of the cardiac cycle in the symmetry plane of the 3 mm shunt model. Red colour means high velocity, while blue low velocity

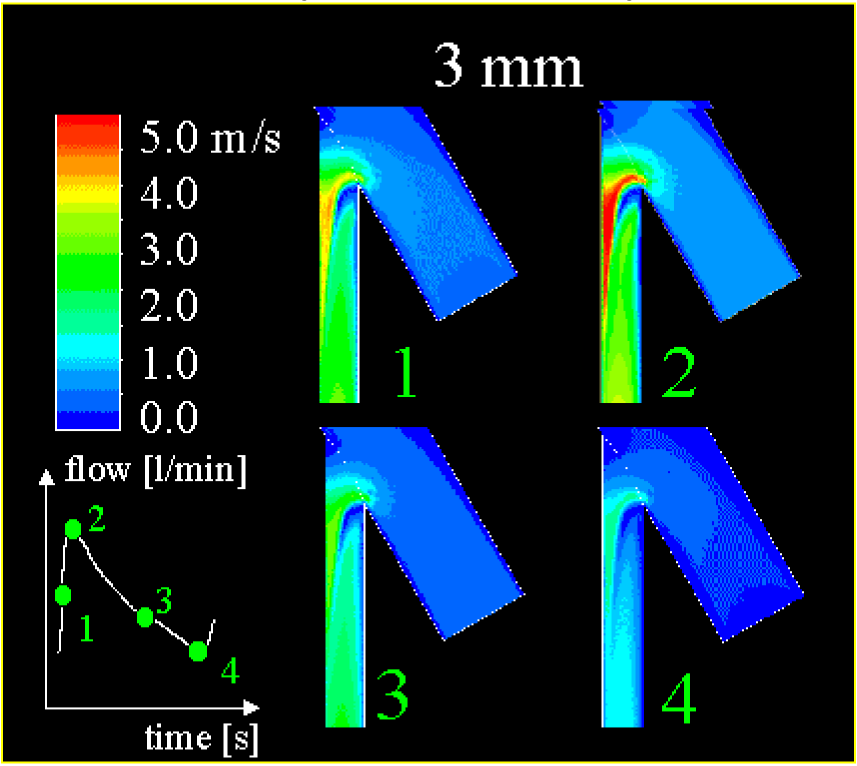


Figure 11. Pressure-flow relationship in systemic-to-pulmonary shunt as function of shunt size evaluated by means of 3-D steady models

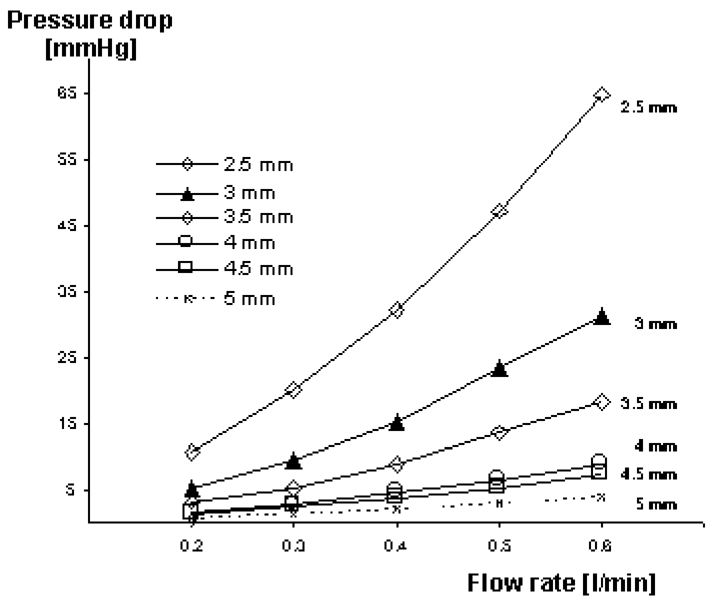
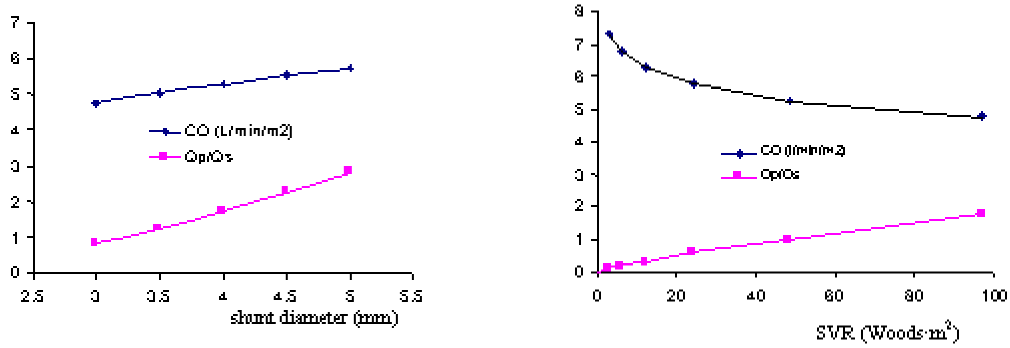


Figure 12. Cardiac output index and Q_P/Q_S as function of shunt size (graph on left) and of systemic vascular resistances (graph on right) for a 3.5 mm shunt



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