Cardiol Young 2004; 14 (Suppl. 3): 11–19

ISSN 1047-9511

Original Article

A study of the Fontan circulation and ventricular energetics based on a model

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PATIENTS WITH CONGENITALLY MALFORMED HEARTS characterised by a functionally single ventricle are currently treated using several procedures that bypass the right heart, such as the Fontan and hemi-Fontan operations, the bidirectional cavopulmonary anastomosis, and the total cavopulmonary connection. All these options are based on the procedures introduced for palliative correction of tricuspid atresia by Fontan and Baudet in 1971.[1] Introduced with the purpose of reducing the pre-operative volume overload, the surgical task mainly consists of separating the pulmonary from the systemic circulation. Irrespective of the specific operation performed, we can call the resulting circulation the Fontan circulation.

Creation of a Fontan circulation has greatly improved the mortality rate in patients with a functionally single ventricle,[2] but in the survivors suffer a considerable morbidity and mortality over the longer term,[3,4] along with a marked decline of functionality as assessed using the categorisation of the New York Heart Association,[5] an abnormal cardiovascular response to exercise,[6] and an increasing likelihood of death with growth and ageing.[7] The reasons for these unsatisfying results can be found in the inevitable passive relationships of pulmonary pressure and flow in the absence of a second ventricle, and in an additional resistance to the venous return.[8]

These consequences of Fontan circulation, although of fundamental importance, are not the only ones affecting the outcome of the surgical procedure, especially in the long term. Thus, the state of pulmonary circulation, the pre-operative value of ventricular contractility and compliance, and the performance of the atrioventricular valve have all been reported significantly to influence the haemodynamic parameters in the midterm, such as end-diastolic volume, end-systolic volume, ejection fraction and end-diastolic pressure.[9–11] Irrespective of how these different parameters influence the known decline in cardiac performance, we know that we observe the progressive development of cardiac failure in the follow-up of patients with a functionally single ventricle. According to Katz,[12] we can describe this adverse evolution as the manifestation of an overload pathology, with starvation of energy and continuous deterioration of the pump function.

Starting from this consideration, the mechano-energetics of the functionally single ventricle appear to offer an important theoretical framework with which to understand the long-term prognosis of patients with the Fontan circulation. In particular, the increased impedance faced by the functionally single ventricle may lead to a mismatch between contractility and afterload, with a consequent reduction in mechanical efficiency that can explain the poor cardiac performance of the surgically treated patients.[13]

The importance of ventriculo-arterial coupling,[14–17] the intrinsic complexity of the cardiovascular system, coupled with its ability to establish sophisticated mechanisms for compensation, point to the need to investigate the energetic and haemodynamic behaviour of the functionally single ventricle as an integrated function of afterload, preload, and contractility as they are modified by the Fontan repair. In this work, we describe the non-linear time-varying lumped-parameter mathematical model we have developed for the Fontan circulation and ventricular function, permitting us to study the effect of parameters such as systemic and pulmonary resistances and compliances on the energetic efficiency and the haemodynamic performance of the functionally single ventricle.

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Accepted for publication July 2003

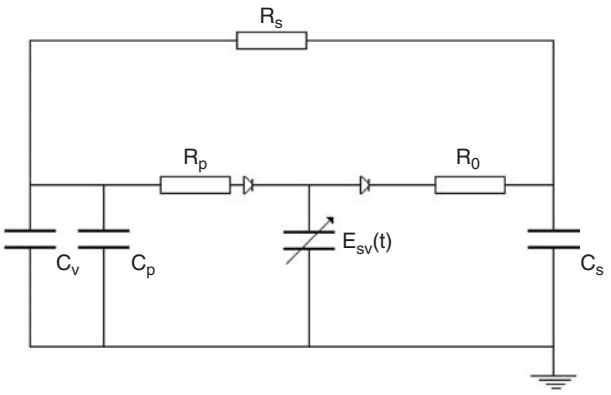

Figure 1.
A lumped-parameter model of the functionally single ventricle and of the Fontan circulation. $E_{sv}(t)$ is the time-varying systolic elastance, R_0 is the systemic characteristic resistance, C_s is the systemic compliance, R_s is the systemic resistance, C_v is the systemic venous compliance, C_p is the pulmonary compliance, R_p is the pulmonary resistance.

The model

By taking advantage of the capabilities of modern computers, mathematic modelling now allows simulations to be made of the cardiovascular system, providing a useful tool for the comprehension of several patho-physiological phenomenons.[18] Various cardiovascular models have been proposed in the last years, depending on the particular aspect of cardiovascular dynamics being focussed on. Here, we used a well-established model of the cardiovascular system (Fig. 1) that is suitable for assessment of the inotropic and energetic reserves of the functionally single ventricle in the pressure–volume plane.[19] The lumped-parameter model, both of the heart and the Fontan circulation, has been kept as simple as possible while still permitting us to describe the main haemodynamic phenomenons, limiting the uncertainty that might be introduced by attempting to explore a greater number of parameters. The functionally single ventricle was represented by a time-varying elastance model.[20] The atrioventricular and neo-aortic valves were assumed to be ideal. The systemic arterial load was represented by a three-element windkessel model,[21] the venous arterial load was represented by a single capacitance, and the pulmonary load was represented by a two-element windkessel model. We ignored the unequivocal contribution made to the overall cardiovascular dynamics by the two atriums and the atrial septal defect, our attention being devoted mainly to the energetics and haemodynamics of the functionally single ventricle and to its coupling to the arterial network.

In agreement with Suga's theory,[19] the function of the ventricle was described by its pressure–volume relation:

$$P_{sv}(t) = e(t)[P_{es}(V_{sv}) - P_{ed}(V_{sv})] + P_{ed}(V_{sv}), \quad (1)$$

where $P_{sv}(t)$ and $V_{sv}(t)$ represented the instantaneous values of the pressure and the volume of the ventricle, P_{es} and P_{ed} represented the ventricular pressure values computed by the end-systolic and end-diastolic pressure–volume relationships; e(t) represented a smoothing function, reproducing the time course of the ventricular elastance.

The end-systolic pressure–volume relationship provided an exhaustive description of the ventricular contractile performance and was given by the relation:

$$P_{es} = E_{sv}(V_{es} - V_0), \quad (2)$$

where P_{es} and V_{es} represented the end-systolic ventricular pressure and volume, E_{sv} represented the end-systolic elastance of the functionally single ventricle, and V_0 represented the ventricular dead volume, namely the ventricular volume corresponding to zero pressure.

The end-diastolic pressure–volume relationship described the ventricular lusitropic properties and was given by the relation:

$$P_{ed} = A(e^{B(V_{ed}-V_0)} - 1), \quad (3)$$

where P_{ed} and V_{ed} represented the end-diastolic pressure and volume, and A and B represented two constants describing the passive-filling properties of the ventricle.

The stroke work, being the external work performed by the functionally single ventricle in ejecting blood through the neo-aortic valve, was defined as the area bounded by the pressure–volume loop. The pressure–volume–area was defined as the area in the pressure–volume plane that is enclosed by the end-diastolic pressure–volume relationship, the end-systolic pressure–volume relationship and the systolic part of the pressure–volume loop. The rate of consumption of oxygen by the myocardium was calculated[22] from pressure–volume–area and end systolic elastance as follows:

$$\begin{aligned}\text{Rate of consumption of oxygen} \\ = \alpha \cdot \text{pressure–volume–area} \\ + \beta \cdot \text{end systolic elastance} + \gamma\end{aligned} \quad (4)$$

where α, β and γ are the three constants.

Ventricular mechanical efficiency (η) was defined as the ratio of stroke work to the total energy supplied to and used by the heart. The efficiency of myocardial conversion (H) was defined as the ratio of pressure–volume–area to the total energy supplied to the heart.[23] Since the pressure–volume–area was defined as the total mechanical energy available for the ventricle, efficiency of myocardial conversion represented the efficiency of global conversion of biochemical into

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Table 1. Parameters for the simulation of the mechanical and energetic behaviour of the functionally single ventricle and of the systemic and pulmonary circulation.

E_{sv} (millimetres of mercury per millilitre)	6
A (millimetres of mercury)	0.5
B (per millilitre)	0.1
V_0 (millilitre)	10
R_0 (millimetres of mercury per millilitre per second)	0.01
R_s (millimetres of mercury per millilitre per second)	2
C_s (millilitres per millimetre of mercury)	0.5
C_v (millilitres per millimetre of mercury)	4
R_p (millimetres of mercury per millilitre per second)	0.2
C_p (millilitres per millimetre of mercury)	1.5
α (per millimetre of mercury)	0.00002
β (square millilitres per millimetre of mercury)	0.002
γ (millilitre)	0.17

mechanical energy. Total energy supplied to the heart was determined from the computed value for myocardial consumption of oxygen, assuming a conversion of 1 millilitre of oxygen into 20 joules.

The ventriculo-arterial coupling was assessed by the ratio between end systolic elastance and effective arterial elastance, this last being given by the ratio between the end-systolic pressure and the stroke volume of the functionally single ventricle.[14,15]

The values for the parameters used in the model were obtained from previously published studies of the functionally univentricular circulation[8,24–28] were reported in Table 1. An abnormal ventricular diastolic function was represented by a reduction of ventricular compliance.[29] A RR interval of 600 milliseconds was considered, with a systole duration of 300 milliseconds.

To assess the impact of the vascular pulmonary resistance on ventricular energetics and haemodynamics, we increased it from a basal value of 0.1 millimetre of mercury per millilitre per second to a final value of 1.1 millimetres of mercury per millilitre per second, increasing by increments of 0.2 millimetre of mercury per millilitre per second.

To assess the impact of the vascular pulmonary compliance on ventricular energetics and haemodynamics, we increased it from a basal value of 0.5 millilitre per millimetre of mercury to a final value of 4 millilitres per millimetre of mercury, increasing by increments of 0.5 millilitre per millimetre of mercury.

The effects of changes in pulmonary compliance and resistance were evaluated in three characteristic states of the systemic vasculature. The simulations discussed above were repeated considering, first, a low systemic impedance with a systemic resistance of 1 millimetre of mercury per millilitre per second, and a systemic compliance of 1 millilitre per millimetre of mercury, second, a normal systemic impedance with a systemic resistance of 2 millimetres of mercury per millilitre per second, and a systemic compliance 0.5 millilitre per millimetre of mercury, and then a high systemic impedance with a systemic resistance of 4 millimetres of mercury per millilitre per second, and a systemic compliance of 0.25 millilitre per millimetre of mercury.

To assess the sensitivity of the computed haemodynamic and energetic parameters to the pulmonary vascular resistance and compliance, we calculated the slope of the linear regression with pulmonary resistance and compliance, for each considered value of the systemic impedance. A p value less than 0.05 was considered statistically significant. The strength of the statistically significant linear regression was quantified by the R^2 value.

Results

When the simulation was performed using the selected values for the parameters, the results obtained for the haemodynamic behaviour of the functionally single ventricle were as follows: end-diastolic volume: 48 millilitres, end-systolic volume: 28 millilitres, stroke volume: 20 millilitres, ejection fraction: 41.6%, end-diastolic pressure: 20 millimetres of mercury; the energetic performance of the ventricle was characterised by stroke work of 0.25 joule, pressure–volume–area of 0.35 joule, with a ventricular mechanical efficiency of 15% and an efficiency of myocardial conversion of 22%. The pulmonary circulation was characterised by a maximum pressure value of 28 millimetres of mercury. The computed ventriculo-arterial coupling was 1.10, with an effective arterial elastance of 5.40 millimetres of mercury per millilitre. In Figures 2 and 3 we show the achieved ventricular, systemic and pulmonary pressure waveforms and the pressure–volume loop.

When the value of pulmonary resistance was increased from 0.1 to 1.1 millimetres of mercury per millilitre per second, the first finding was that the pressure–volume–area, the myocardial consumption of oxygen, the efficiency of myocardial conversion, the maximum pulmonary pressure and the end-diastolic volume all decreased, maintaining greater values for a higher systemic impedances. We then noted that ventriculo-arterial coupling decreased, maintaining greater values for a lower systemic impedance. In the third place, stroke work decreased, maintaining lower values for a low systemic impedance. A higher stroke work was obtained for normal systemic impedance until we reached a value of pulmonary resistance of 0.3 millimetre of mercury per millilitre per second. For values of pulmonary resistance greater than 0.3 millimetre of mercury per millilitre per second, the higher values of stroke work were obtained for a high systemic impedance. In the fourth place,

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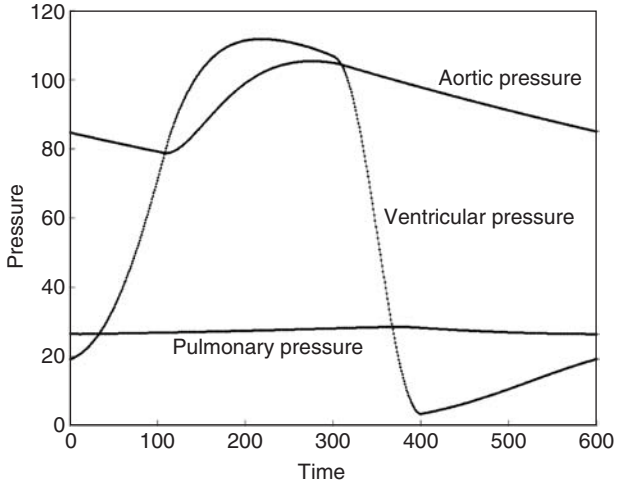

Figure 2.
Pressure waveforms computed for the values of the parameters shown in Table 1.

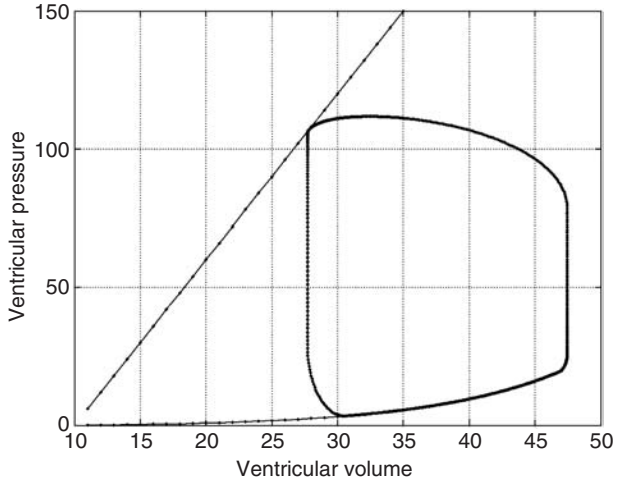

Figure 3.
Pressure–volume loop, end-systolic pressure–volume relationship and end-diastolic pressure–volume relationship computed for the parameters shown in Table 1.

ventricular mechanical efficiency decreased (Fig. 4), maintaining greater values for a low systemic impedance until we reached a value of pulmonary resistance of 0.3 millimetre of mercury per millilitre per second. Then, until we reached a value of 0.9 millimetre of mercury per millilitre per second, a greater mechanical effiicency was obtained for a normal systemic impedance. When pulmonary resistance was greater than 0.9 millimetre of mercury per millilitre per second, a greater mechanical efficiency was obtained for a high systemic impedance. Finally, stroke volume decreased, maintaining a greater value for a lower systemic impedance. The difference between the stroke volume, obtained with the three

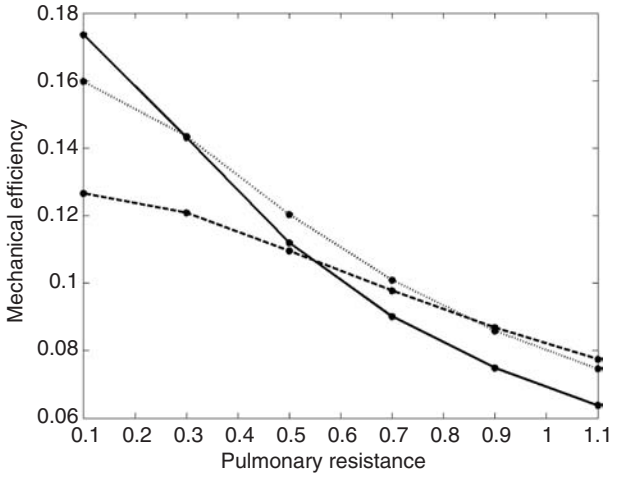

Figure 4.
Mechanical efficiency as a function of pulmonary resistance, for different systemic loads. The solid line stands for a low systemic impedance, the dotted line stands for a normal systemic impedance, and the dashed line stands for a high systemic impedance.

different systemic loads, markedly decreased for increasing values of pulmonary resistance. In Table 2, we show the range of variation of the considered haemodynamic and energetic parameters for the different systemic loads. In Table 3, we show the slope, the R^2, and the p values of the linear regressions performed between the increasing pulmonary resistance and the varying parameters. These show that the slopes of the regression lines, namely the sensitivities of the computed haemodynamic and energetic parameters to the pulmonary vascular resistance, were dependent to the systemic load. The sensitivity of ventriculo-arterial coupling, pressure–volume–area, and myocardial consumption of oxygen, increased for higher values of systemic impedance, while the sensitivity of mechanical efficiency, efficiency of myocardial conversion, stroke volume, end-diastolic volume and maximum pulmonary pressure all decreased for higher values of systemic impedance. The sensitivity of stroke work showed its maximum value for a normal systemic impedance and its minimum value for an high systemic impedance.

When the value of pulmonary compliance was increased from 0.5 to 5 millilitres per millimetre of mercury, we found, first, that the pressure–volume–area, the myocardial consumption of oxygen, the efficiency of myocardial conversion, the maximum pulmonary pressure and the ventricular end-diastolic volume all decreased, maintaining greater values for a higher systemic impedance. Secondly, ventriculo-arterial coupling increased, maintaining greater values for a lower systemic impedance. Third, stroke work decreased, maintaining lower values for a low systemic impedance. A higher stroke work value was

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Table 2. Range of variation for the different considered haemodynamic and energetic parameters, when the pulmonary resistance was increased from the minimum value (R_{pmin}) of 0.1 to the maximum value (R_{pmax}) of 1.1 millimetres of mercury per millilitre per second. Pressure–volume–area, stroke work and myocardial oxygen consumption are expressed in joules; maximum pulmonary pressure in millimetres of mercury; end-diastolic volume and stroke volume in millilitres.

	Low systemic impedance		Normal systemic impedance		High systemic impedance	
	Value for R_{pmin}	Value for R_{pmax}	Value for R_{pmin}	Value for R_{pmax}	Value for R_{pmin}	Value for R_{pmax}
Ventriculo-arterial coupling	2.048	1.043	1.136	0.747	0.600	0.481
Mechanical efficiency	0.174	0.064	0.16	0.075	0.127	0.077
Efficiency of myocardial conversion	0.198	0.083	0.221	0.125	0.237	0.163
Pressure–volume–area	0.284	0.075	0.380	0.116	0.474	0.186
Stroke work	0.249	0.051	0.275	0.069	0.254	0.088
Myocardial oxygen consumption	1.433	0.804	1.720	0.927	2.003	1.137
Maximum pulmonary pressure	26	31	28	33	30	34
Stroke volume	26	8	21	8	15	8
End-diastolic volume	48	26	49	29	50	34

Table 3. Slopes, R^2 and p values for the linear regression between the different considered haemodynamic and energetic parameters and the pulmonary resistance. The slopes are expressed in: millilitres per second per millimetre of mercury for mechanical efficiency, efficiency of myocardial conversion and ventriculo-arterial coupling; joules per millimetres of mercury per millilitre per second for pressure–volume–area, stroke work and myocardial oxygen consumption; millilitres per second for maximum pulmonary pressure; square millilitres per millimetre of mercury per second for end-diastolic volume and stroke volume.

	Low systemic impedance			Normal systemic impedance			High systemic impedance		
	Slope	R^2	p	Slope	R^2	p	Slope	R^2	p
Ventriculo arterial coupling	−1.013	0.977	<0.001	−0.4	0.996	<0.001	−0.123	0.993	<0.001
Mechanical efficiency	−0.111	0.963	<0.001	−0.082	0.988	<0.001	−0.052	0.993	<0.001
Efficiency of myocardial conversion	−0.107	0.977	<0.001	−0.1	0.991	<0.001	−0.077	0.991	<0.001
Pressure–volume–area	−0.206	0.915	<0.001	−0.27	0.954	<0.001	−0.302	0.986	<0.001
Stroke work	−0.193	0.903	0.004	−0.209	0.946	0.001	−0.173	0.982	<0.001
Myocardial oxygen consumption	−0.620	0.916	<0.001	−0.809	0.954	<0.001	−0.905	0.987	<0.001
Maximum pulmonary pressure	5.271	0.937	<0.001	5.074	0.969	<0.001	4.123	0.993	<0.001
Stroke volume	−17.092	0.932	<0.001	−12.937	0.968	<0.001	−7.761	0.992	<0.001
End-diastolic volume	−21.603	0.933	0.002	−20.440	0.968	<0.001	−17.364	0.992	<0.001

obtained for a normal systemic impedance until we reached a value for pulmonary compliance of 4 millilitres per millimetre of mercury. For values of pulmonary compliance greater than 4 millilitres per millimetre of mercury, the higher values of stroke work were obtained for a high systemic impedance. We then found that mechanical effiicency decreased (Fig. 5), maintaining lower values for a high systemic impedance. Higher values for mechanical efficiency were obtained for a low systemic impedance until a value for pulmonary compliance of 2 millilitres per millimetre of mercury was reached. For values of pulmonary compliance greater than 2 millilitres per millimetre of mercury, a greater value for mechanical efficiency was obtained with a normal systemic impedance. We also found that stroke volume decreased, maintaining greater values for a lower systemic impedance. In Table 4, we show the range of variation of the considered haemodynamic and energetic parameters for the different systemic loads. In Table 5, we illustrate the slope, the R^2, and the p values of the linear regressions performed between the increasing pulmonary compliance and the varying parameters. The slopes of the regression lines, namely the sensitivities of the computed haemodynamic and energetic parameters to the pulmonary vascular compliance, were again dependent to the systemic load. The sensitivity of maximum pulmonary pressure, pressure–volume–area and myocardial oxygen consumption increased for higher values of systemic impedance, whereas the sensitivity of mechanical efficiency, efficiency of myocardial conversion, stroke volume, ventriculo-arterial coupling and ventricular end-diastolic volume all decreased for higher values of systemic impedance. The sensitivity of stroke work was found to be the same for a low and a normal

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systemic impedance, but lower for a high systemic impedance.

Conclusions

It is clear that both the systemic and pulmonary loads are significant in determining the haemodynamic and energetic behaviour of the functionally single ventricle. The simulations we performed showed a prevailing effect of the systemic load on the haemodynamic parameters such as end-diastolic volume, stroke volume and maximum pulmonary pressure, with a significant modulation due to pulmonary resistance and compliance. The greater the systemic impedance, the worse the haemodynamic performance of the functionally single ventricle, with lower values for stroke volume and higher values for end-diastolic volume and maximum pulmonary pressure. The haemodynamic performance, fundamentally determined by the systemic load, was then affected in different ways by pulmonary resistance and compliance. Thus, an increase in pulmonary resistance and compliance resulted in a decrease of stroke volume. The pulmonary pressure was increased by a lower pulmonary compliance and by a higher pulmonary resistance value. With increasing pulmonary resistance and compliance, end-diastolic volume decreased. Each haemodynamic parameter, for each systemic loading condition, appeared to be more sensitive to changes in pulmonary resistance than compliance.

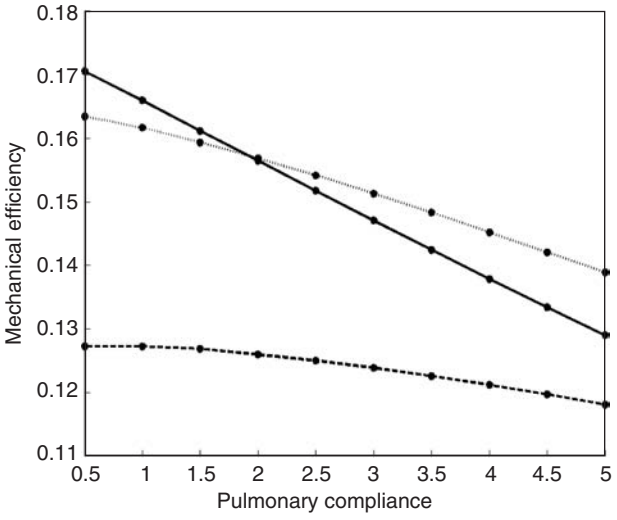

Figure 5.
Mechanical efficiency as a function of pulmonary compliance, for different systemic loads. The solid line stands for a low systemic impedance, the dotted line stands for a normal systemic impedance, and the dashed line stands for a high systemic impedance.

The ventriculo-arterial coupling appeared to be largely determined by systemic impedance, except for the condition of a low systemic impedance, when ventriculo-arterial coupling markedly decreased for increasing values of pulmonary resistance (Figs 6 and 7). Ventriculo-arterial coupling has been extensively investigated in normal and failing human hearts so as to establish its role in the efficiency of left ventricular work and the transfer of energy. The optimal stroke work is predicted when the effective arterial elastance is equal to the ventricular end-systolic elastance, with a ventriculo-arterial coupling of 1. The maximal mechanical efficiency is reached when the end-systolic elastance is twice as large as the effective arterial elastance, with a ventriculo-arterial coupling of 2. With failing ventricular function, specifically an ejection fraction of less than 40%, the end-systolic elastance has been found to be less than one-half of effective arterial elastance,[30]

Table 4. Range of variation for the different considered haemodynamic and energetic parameters, when the pulmonary compliance was increased from the minimum value (C_{pmin}) of 0.5 to the maximum value (C_{pmax}) of 5 millilitres per millimetre of mercury. Pressure–volume–area, stroke work and myocardial oxygen consumption are expressed in joules; maximum pulmonary pressure in millimetres of mercury; end-diastolic volume and stroke volume in millilitres.

	Low systemic impedance		Normal systemic impedance		High systemic impedance	
	Value for C_{pmin}	Value for C_{pmax}	Value for C_{pmin}	Value for C_{pmax}	Value for C_{pmin}	Value for C_{pmax}
Ventriculo-arterial coupling	1.889	2.050	1.076	1.168	0.580	0.626
Mechanical efficiency	0.171	0.129	0.159	0.133	0.127	0.115
Efficiency of myocardial conversion	0.200	0.152	0.224	0.184	0.241	0.211
Pressure–volume–area	0.289	0.162	0.397	0.239	0.504	0.332
Stroke work	0.247	0.138	0.282	0.172	0.266	0.181
Myocardial oxygen consumption	1.446	1.066	1.773	1.299	2.093	1.587
Maximum pulmonary pressure	31	18	34	19	36	20
Stroke volume	24	18	21	16	15	13
End-diastolic volume	47	37	50	40	52	43

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with a ventriculo-arterial coupling less than 0.5. In the performed simulations, ventriculo-arterial coupling was near 2 for a low systemic impedance and low values for pulmonary resistance, suggesting an attempt to maximise mechanical efficiency. The ventriculo-arterial coupling was near 1 for a normal systemic impedance, suggesting an attempt at maximising stroke work. Ventriculo-arterial coupling was near 0.5 for a high systemic impedance, indicating a mismatch between the ventricle and the circulatory network, and a failing ejective ventricular performance since the ejection fraction was near to 30% independently of the values of pulmonary resistance and compliance.

When we consider the behaviour of the energetics of the functionally single ventricle, the prevailing effect of the systemic load became clear with regard to the myocardial consumption of oxygen, the pressure–volume–area, and the efficiency of myocardial conversion (Figs 8 and 9), showing greater values for greater systemic impedances. The greater systemic load, however, was sustained only at the price of a greater total mechanical energy as shown by the pressure–volume–area, and a greater myocardial consumption of oxygen. At all events, the efficiency of myocardial conversion increased with the load in order to maintain the consumption of oxygen as low as possible. The increase of pulmonary resistance and compliance diminished the values of these parameters as determined by systemic impedance, with a greater sensitivity to the changes in pulmonary resistance.

Table 5. Slopes, R^2 and p values for the linear regression between the different considered haemodynamic and energetic parameters and the pulmonary compliance. The slopes are expressed in: millimetres of mercury per millilitre for mechanical efficiency, efficiency of myocardial conversion and ventriculo-arterial coupling; joules per millilitre per millimetre of mercury for pressure–volume–area, stroke work and myocardial oxygen consumption; square millimetres of mercury per millilitre for maximum pulmonary pressure; millimetres of mercury for end diastolic volume and stroke volume.

	Low systemic impedance			Normal systemic impedance			High systemic impedance		
	Slope	R^2	p	Slope	R^2	p	Slope	R^2	p
Ventriculo-arterial coupling	0.035	0.985	<0.001	0.019	0.947	<0.001	0.01	0.962	<0.001
Mechanical efficiency	−0.0093	0.999	<0.001	−0.0059	0.999	<0.001	−0.0028	0.985	<0.001
Efficiency of myocardial conversion	−0.011	1.000	<0.001	−0.0089	0.999	<0.001	−0.0067	0.999	<0.001
Pressure–volume–area	−0.028	0.990	<0.001	−0.035	0.994	<0.001	−0.038	0.997	<0.001
Stroke work	−0.024	0.990	0.004	−0.024	0.995	0.001	−0.019	0.997	<0.001
Myocardial oxygen consumption	−0.084	0.992	<0.001	−0.105	0.993	<0.001	−0.114	0.994	<0.001
Maximum pulmonary pressure	−2.701	0.973	<0.001	−3.156	0.970	<0.001	−3.524	0.967	<0.001
Stroke volume	−1.334	0.997	<0.001	−0.969	1.000	<0.001	−0.556	1.000	<0.001
End-diastolic volume	−2.196	0.996	0.002	−2.110	0.997	<0.001	−1.866	0.997	<0.001

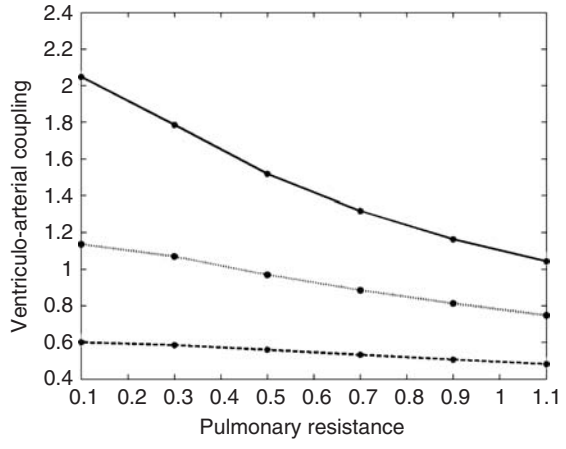

Figure 6.
Ventriculo-arterial coupling as a function of pulmonary resistance, for different systemic loads. The solid line stands for a low systemic impedance, the dotted line stands for a normal systemic impedance, and the dashed line stands for a high systemic impedance.

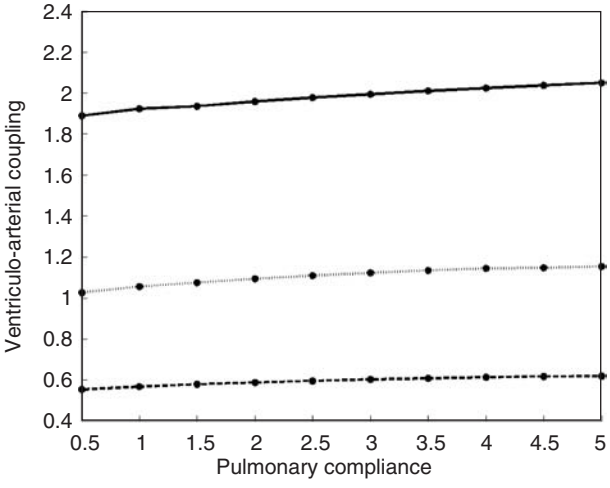

Figure 7.
Ventriculo-arterial coupling as a function of pulmonary compliance, for different systemic loads. The solid line stands for a low systemic impedance, the dotted line stands for a normal systemic impedance, and the dashed line stands for a high systemic impedance.

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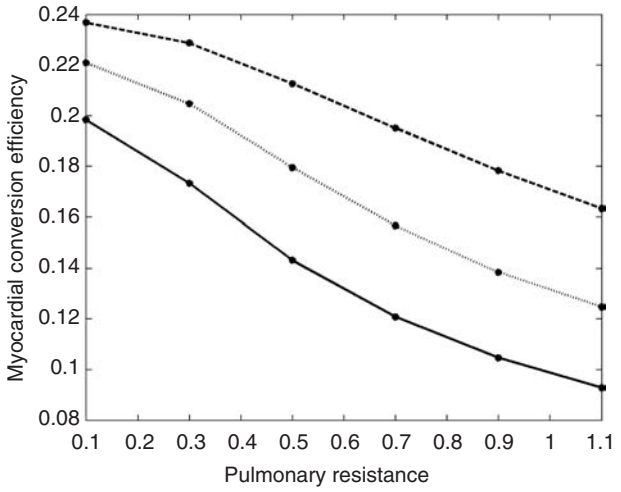

Figure 8.
Efficiency of myocardial conversion as a function of pulmonary resistance, for different systemic loads. The solid line stands for a low systemic impedance, the dotted line stands for a normal systemic impedance, and the dashed line stands for a high systemic impedance.

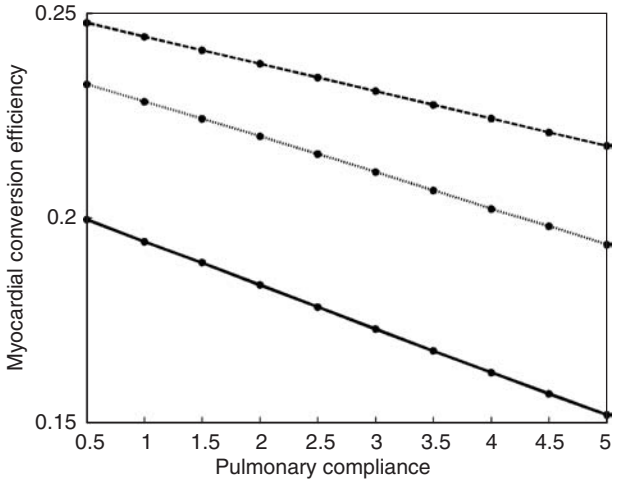

Figure 9.
Efficiency of myocardial conversion as a function of pulmonary compliance, for different systemic loads. The solid line stands for a low systemic impedance, the dotted line stands for a normal systemic impedance, and the dashed line stands for a high systemic impedance.

The interdependence between stroke work and mechanical efficiency (Figs 4 and 5) and the systemic and pulmonary loads was more complex. Even for these parameters, a clear decrease was obtained for increasing values of pulmonary resistance and compliance, but we no longer found a prevailing effect of systemic impedance. In particularly, the higher values for mechanical efficiency and stroke work were obtained for a combination of low systemic impedadance, low pulmonary resistance and compliance.

The difference between pressure–volume–area and stroke work, and then between the efficiency of myocardial conversion and mechanical efficiency, is given by the amount of mechanical energy available but not used by the ventricle to perform the external work and to guarantee the cardiac output. This difference is indicated as ventricular potential energy.[19]

The results we have obtained suggest that, in the Fontan circulation, the interplay between systemic and pulmonary load is of fundamental importance in determining the amount of total mechanical energy available to be converted into external work. Moreover, when considering the effects of the pulmonary network on ventricular energetics, we have to evaluate separately the effect of viscous resistance and of vascular compliance. We found that, rather than the net value of the pulmonary impedance being important, of greater effect were the single values of resistance and compliance.

The influence of systemic and pulmonary impedances on the haemodynamic and energetic behaviour of the functionally single ventricle, however, appeared ambiguous. The haemodynamic performance was supported by lower values of systemic impedance and pulmonary resistance. A lower value of pulmonary compliance supported the pulmonary circulation, guaranteeing a lower maximum pulmonary pressure, but damaged the ejective performance, decreasing stroke volume. A lower value of both pulmonary resistance and compliance supported the ventricular energetics, with higher values of mechanical efficiency and efficiency of myocardial conversion. A lower value of systemic impedance, in contrast, supported mechanical efficiency and damaged efficiency of myocardial conversion. These considerations suggest there will be a major difficulty, in terms of the cardiovascular system, in identifying an unambiguous point of optimal behaviour in the non-physiological Fontan circulation as it declines progressively towards overt heart failure.

Even if further investigations are needed in order to clarify the importance of ventricular lusitropic and inotropic performance,[31,32] the knowledge of the specific role of the systemic impedance, and of the higher incidence of changes in pulmonary resistance, rather than compliance, on the energetic and haemodynamic behaviour, as evidenced by the performed simulations, may guide our future clinical treatment of patients converted to the Fontan circulation.

Acknowledgments

We thank Ms Francesca Pardini for skilful secretarial assistance and Mr Emiliano Rocca for excellent technical support.

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