Optimum Conditions for the Cardio-vascular-pulmonary System Obtained in the Irreversible Finite Speed Thermodynamics Framework

I. Oxygen flow as a function of blood speed

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Considering the diffusion of a gas through the semipermeable walls of a tube, we find the dependency between the blood speed and the flow of the gas transported by the cardio-vascular system. We show that there is no maximum point for the oxygen flow, which starts approximately directly proportional to the blood speed (for small speeds), and then reaches a plateau (for higher speeds).

Keywords: semipermeable walls , blood speed, cardio-vascular system, oxygen flow

Presently it is estimated that on the entire planet about 50% of the deaths are because of cardiovascular diseases. From this follows the extraordinary importance of the research in the Cardio-Vascular-Pulmonary field. There is hope that Cardio-Vascular Physiology [8, 13] must and can add substantially to the understanding of this extremely complex system, which plays a vital role in the life and wellbeing of animals with blood (including humans). In recent years, this research has been substantially helped in the colossal effort of understanding the complexities of the *finite speed processes* in this system by Physics [1], Biological Thermodynamics [1, 5, 6, 7, 21] and recently by the new branch of Irreversible Thermodynamics, *Finite Speed Thermodynamics* (FST) [3, 4, 16, 18, 16].

The components of the cardio-vascular-pulmonary system are viewed in TVF analogous to real thermal machines, which evidently work also with *finite speed* to produce or use mechanical power in order to obtain a useful effect. The useful effect of the cardio-vascular-pulmonary system deals with at least four fundamental aspects (for the normal/healthy functioning of the whole organism):

- passing the Oxygen from air to blood (in lungs);
- transporting the dissolved Oxygen to all the cells in the organism;
- transporting and removing the CO_2 and H_2O produced by "burning", at 37°C, the fuel (food eaten) converted to AdenosyneTriPhosphate (ATP) in the liver through the metabolism in stomach and intestines, and
- removing part of the heat produced by these biochemical reactions through the air expelled by the lungs. The rest of the heat is removed through skin (convection and perspiration-vaporization), urine and fecal matter.

We consider as an essential fact that both in thermal machines and in animal organisms, *the processes' speed being finite*, the Irreversible Thermodynamics with Finite Speed can help not only to deepen the understanding of these processes (from a thermodynamic viewpoint), but to model them quantitatively, with the hope that the models can be validated, as it was possible in FST for the very complex Stirling machines [16].

The components of the cardio-vascular-pulmonary system are viewed in FST as analogous to thermal machines not only due to the fact that both work with finite speed, but obviously *the heart* is actually a pump transporting blood (with various *finite speeds*, in different functioning regimes – just like a water pump which can work at different speeds/regimes), and *the lung* is nothing else than the "equivalent" of two piston compressors, which also work in different regimes with various *finite speeds*, just like an air compressor can work at different speeds.

The colossal research and financial efforts made today towards building an Artificial Heart and an Artificial Lung require the mechanical engineers to "enter the arena" of this research, as was recently emphasized in the Mechanical Engineering Journal (American Society of Mechanical Engineering), in the paper "Re-engineering Healthcare" [14]. The recent breakthrough of the French researchers from the CARMAT company [2], which lasted almost 15 years and used 55 million Euro to create *the first artificial heart* meant to work in a human body for about 5 years, also illustrates the colossal efforts made for building a "thermal machine" = *artificial heart*, to save and prolong millions of lives.

In the first part of the present paper we study, using the concepts and the approach of FST, the correlation between *Oxygen flow* and *blood speed*. In the second part we will determine analytically the power produced by the whole organism and we will show there exists a certain optimum blood speed and an optimum number of capillaries that maximize this power. Future research is aimed at other aspects of the cardio-vascular-pulmonary system, taking into account more and more functional and "constructive" parameters.

Gas diffusion through the semipermeable walls of a tube

Stationary fluid

We consider a fluid at rest in a tube of constant cross-sectional area A. The wall of the tube is a semipermeable membrane: it lets a certain gas to pass through it, but not the fluid inside the tube. We delimit an infinitesimal portion of fluid of length dx and volume dV (fig. 1). In this volume

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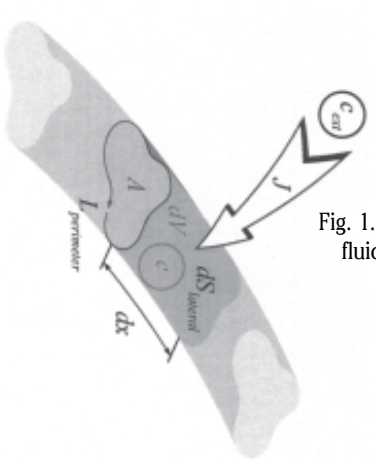


Fig. 1. An infinitezimal portion of fluid in a tube with constant cross-sectional area

of fluid we consider that a mass dm of gas is dissolved, having the *mass concentration* c:

$$c = \frac{dm}{dV} \tag{1}$$

Outside the tube we consider that the gas has mass concentration c_{ext}.

Neglecting the gas passing through the cross-sectional surfaces, the mass flux J of gas entering the dV volume through the wall is, according to Fick's law [15, 20], directly proportional to the concentration gradient across the membrane:

$$J = -K_d \nabla c, \tag{2}$$

where K_d is the membrane's *diffusivity* for Oxygen. This parameter is influenced by the membrane's structure, age etc. According to the *Linear Irreversible Phenomenological Thermodynamics* of Onsager and Prigogine [15, 20], this dependency can be generalized to any pair thermodynamic force – associated thermodynamic flux. But in this work we study only diffusion, because this is the phenomenon we are interested in.

Considering that the membrane has a constant thickness s, the concentration gradient across the membrane is:

$$\nabla c = \frac{c - c_{ext}}{s} \tag{3}$$

It follows that the gas flux passing through the membrane is:

$$J = \frac{K_d}{s}(c_{ext} - c) = K_p(c_{ext} - c) \tag{4}$$

We wrote K_p for the ratio K_d/s, which is called *permeability* and characterizes the membrane (relatively to the considered gas).

The gas flux J passing through the area $dS_{lateral}$ causes a gas flow $JdS_{lateral}$ entering the dV volume, and this leads to an increase of the gas concentration inside that volume. From mass conservation it follows that the speed of this concentration increase is:

$$\frac{\partial c}{\partial t} = \frac{JdS_{lateral}}{dV} \tag{5}$$

We define:

$$\gamma = \frac{dS_{lateral}}{dV} = \frac{S_{lateral}}{V_{total}} = \frac{L_{perimeter}}{A} \tag{6}$$

This is a geometrical feature we call the form factor of the tube.

We obtained that the speed of variation of the gas concentration is directly proportional to the gas flux entering the tube:

$$\frac{\partial c}{\partial t} = \gamma J \tag{7}$$

We also define:

$$K = K_p \gamma \tag{8}$$

Combining formulae (4) and (7), we obtain that c satisfies the following differential equation with respect to time:

$$\frac{\partial c}{\partial t} = K(c_{ext} - c) \tag{9}$$

For a tube whose geometrical characteristics (the form factor) and permeability do not vary in time, the factor K is constant. Considering the gas concentration outside the tube c_{ext} also constant in time, the equation's solution is an exponential evolution of the inside gas concentration, starting at the initial value c_{init} (when $t = 0$) and tending asymptotically towards c_{ext} as time passes:

$$c(t) = c_{init} + (c_{ext} - c_{init})(1 - e^{-Kt}) \tag{10}$$

Substituting this solution in formula (4), we obtain that the gas flux entering the volume dV at moment t is:

$$J(t) = K_p(c_{ext} - c_{init})e^{-Kt} \tag{11}$$

Fluid in motion

The solution (11) is valid regardless of the motion/position of the volume dV. Assuming that the fluid moves in the tube with a constant speed w, a volume dV would have reached distance x in the time $t = x/w$ (fig. 2). From (11) it follows that the gas flux entering that portion of the tube is:

$$J(x) = K_p(c_{ext} - c_{init})e^{-K\frac{x}{w}} \tag{12}$$

Knowing that the lateral area is $dS_{lateral} = L_{perimeter}dx$ and integrating for the tube's length L_{total}, we obtain that the total gas flow entering the tube is:

$$\dot{m} = \int_{S_{lateral}} J(x)dS_{lateral} = \int_0^{L_{total}} J(x)L_{perimeter}\,dx$$

$$\dot{m} = (c_{ext} - c_{init})D\left(1 - e^{-\frac{KL_{total}}{w}}\right), \tag{13}$$

where $D = Aw$ is the fluid volumetric flow passing through the tube.

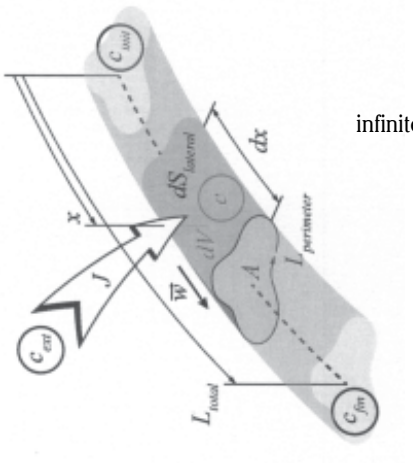


Fig. 2. The infinitezimal slice of fluid in motion

We define the non-dimensional quantity ξ as the ratio:

$$\xi = \frac{w}{KL_{total}} = \frac{D}{K_p S_{lateral}} = \frac{A}{K_p S_{lateral}} w \qquad (14)$$

We call this ratio the reduced speed of the fluid passing through the tube. The reduced speed is directly proportional to the actual speed of the fluid.

With this, the gas flow entering the tube can be written:

$$\dot{m} = K_p S_{lateral} (c_{ext} - c_{init}) \xi \left(1 - e^{-\frac{1}{\xi}}\right) \qquad (15)$$

Because each volume dV travels along the entire tube in the time $\Delta t_{total} = L_{total}/w$, after a few calculations we obtain from formula (10) the Oxygen concentration c_{fin} at the tube's exit:

$$c_{fin} = c_{init} + (c_{ext} - c_{init}) \left(1 - e^{-\frac{1}{\xi}}\right) \qquad (16)$$

Cardio-vascular application

We consider the following model of the cardio-vascular system: a fluid (blood) circulating in closed loop between two "Oxygen exchangers" with semipermeable walls (fig. 3). Each "Oxygen exchanger" is a system of parallel capillaries, which we assimilate to a single tube with constant cross-sectional area and huge lateral area. The fluid enters the pulmonary capillaries having the Oxygen concentration c_1 (low); outside the tube (in the pulmonary alveoli), the Oxygen concentration is c_H (much higher), so that as the fluid travels through the pulmonary capillaries it enriches with Oxygen due to diffusion; when exiting the pulmonary capillaries, the fluid reaches an Oxygen concentration $c_2 > c_1$; then it is transported to the systemic capillaries, outside of which the Oxygen concentration is c_L (much lower); there the blood releases Oxygen through diffusion, the Oxygen concentration dropping again to c_1; finally, the blood is transported back to the pulmonary capillaries and the cycle restarts.

The concentrations in stationary regime

For the two "Oxygen exchangers" (pulmonary and systemic, respectively) we impose the conditions: 1) the concentration c_{finH} at the exit of the pulmonary capillaries has to equal the concentration c_2 at the entrance of the systemic capillaries, and respectively 2) the concentration c_{finL} at the exit of the systemic capillaries hat to equal the concentration c_1 at the entry in the pulmonary capillaries:

$$\begin{cases} c_{fin_H} = c_1 + (c_H - c_1)\left(1 - e^{-\frac{1}{\xi_H}}\right) = c_2 \\ c_{fin_L} = c_2 + (c_L - c_2)\left(1 - e^{-\frac{1}{\xi_L}}\right) = c_1 \end{cases} \qquad (17)$$

By denoting

$$\begin{cases} \varepsilon_H = e^{-\frac{1}{\xi_H}} \\ \varepsilon_L = e^{-\frac{1}{\xi_L}} \end{cases}, \qquad (18)$$

the system of equations becomes:

$$\begin{cases} -\varepsilon_H c_1 + c_2 = c_H (1 - \varepsilon_H) \\ c_1 - \varepsilon_L c_2 = c_L (1 - \varepsilon_L) \end{cases} \qquad (19)$$

From this system of two equations with two unknowns we can find the concentrations c_1 and c_2 at which the system stabilizes in stationary regime:

$$\begin{cases} c_1 = \dfrac{c_H (1 - \varepsilon_H)\varepsilon_L + c_L (1 - \varepsilon_L)}{1 - \varepsilon_H \varepsilon_L} \\ c_2 = \dfrac{c_L (1 - \varepsilon_L)\varepsilon_H + c_H (1 - \varepsilon_H)}{1 - \varepsilon_H \varepsilon_L} \end{cases} \qquad (20)$$

With the notation $\Delta c_{HL} = c_H - c_L$, these can be written in a simpler form:

$$\begin{cases} c_1 = c_H - \Delta c_{HL} \dfrac{1 - \varepsilon_L}{1 - \varepsilon_H \varepsilon_L} \\ c_2 = c_L + \Delta c_{HL} \dfrac{1 - \varepsilon_H}{1 - \varepsilon_H \varepsilon_L} \end{cases} \qquad (21)$$

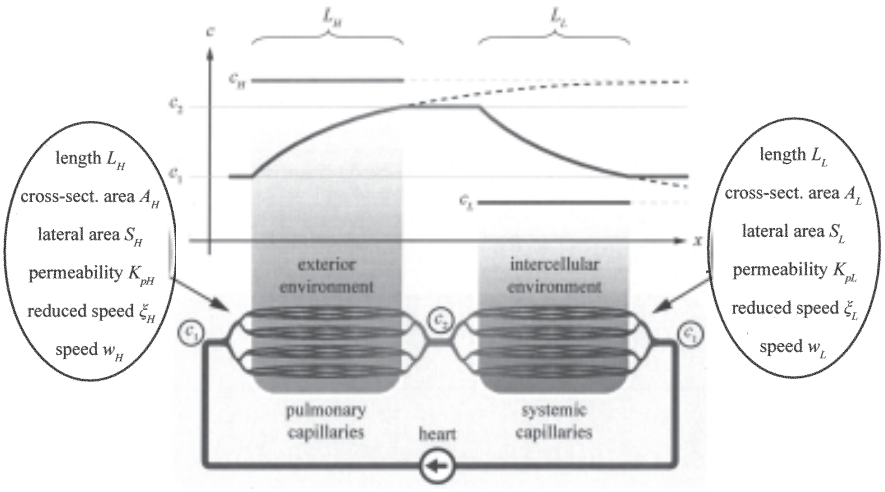
The oxygen flow in stationary regime

Knowing the concentrations c_1 and c_2, we use the formula (13) to obtain the oxygen flows in stationary regime:

$$\begin{cases} \dot{m}_H = (c_H - c_1) D (1 - \varepsilon_H) \\ \dot{m}_L = (c_L - c_2) D (1 - \varepsilon_L) \end{cases}$$

$$\begin{cases} \dot{m}_H = \Delta c_{HL} D \dfrac{(1 - \varepsilon_H)(1 - \varepsilon_L)}{1 - \varepsilon_H \varepsilon_L} \\ \dot{m}_L = -\Delta c_{HL} D \dfrac{(1 - \varepsilon_L)(1 - \varepsilon_H)}{1 - \varepsilon_H \varepsilon_L} \end{cases} \qquad (22)$$

The two oxygen flows are equal and have opposite signs (which was to be expected, because in stationary regime the system does not accumulate and does not produce oxygen – how much oxygen enters from the pulmonary



Fg. 3.The model of the cardio-vasculary system
the cycle restarts

alveoli, exactly that much is released into the intercellular environment).

It follows that the oxygen flow transported by the cardio-vascular system is:

$$\dot{m} = \Delta c_{HL} D \frac{\left(1-e^{-\frac{1}{\xi_H}}\right)\left(1-e^{-\frac{1}{\xi_L}}\right)}{1-e^{-\left(\frac{1}{\xi_H}+\frac{1}{\xi_L}\right)}} \quad (23)$$

This can be written as:

$$\dot{m} = \underbrace{\Delta c_{HL} \frac{\left(K_{pH}S_{lateralH}\right)\left(K_{pL}S_{lateralL}\right)}{K_{pH}S_{lateralH}+K_{pL}S_{lateralL}}}_{\dot{m}_{max}} \cdot \underbrace{(\xi_H+\xi_L)\frac{\left(1-e^{-\frac{1}{\xi_H}}\right)\left(1-e^{-\frac{1}{\xi_L}}\right)}{1-e^{-\left(\frac{1}{\xi_H}+\frac{1}{\xi_L}\right)}}}_{f(\xi_H,\xi_L)} \quad (24)$$

The first factor represents the oxygen flow that would be transported if the membranes were adjacent (not separated by blood) and the oxygen diffused through them directly from air to the intercellular environment. This is the maximum oxygen flow which the cardio-vascular system would be able to provide:

$$\dot{m}_{max} = \Delta c_{HL} \frac{\left(K_{pH}S_{lateralH}\right)\left(K_{pL}S_{lateralL}\right)}{K_{pH}S_{lateralH}+K_{pL}S_{lateralL}} \quad (25)$$

If the two membranes are identical (with the same permeability K_p and the same area $S_{lateral}$), the maximum Oxygen flow becomes:

$$\dot{m}_{max} = \frac{1}{2} K_p S_{lateral} \Delta c_{HL} \quad (26)$$

The factor 1/2 is there because the oxygen has to pass through two semipermeable membranes and consequently the flow is halved compared to the case when the oxygen has only one membrane to permeate.

The second factor in formula (24) is non-dimensional and captures the influence of the blood speed. The function $f(\xi_H, \xi_L)$ is always positive and subunitary, tending to 1 when ξ_H and $\hat{\imath}_L$ tend to infinity. This means that the oxygen flow tends to the maximum values as the blood speed w tends to infinity.

The ratio between the actual oxygen flow transported by the cardio-vascular system and the maximum oxygen flow which would be transported if the length of the system were zero can be regarded as the efficiency of the cardio-vascular system:

$$\eta_{cardiovascular} = \frac{\dot{m}}{\dot{m}_{max}} = f(\xi_H, \xi_L) \quad (27)$$

If the semipermeable membranes are identical, the reduced speed has an unique value ξ and the function $f(\xi)$ becomes:

$$f(\xi) = 2\xi \frac{e^{\frac{1}{\xi}}-1}{e^{\frac{1}{\xi}}+1} \quad (28)$$

In the following part we will analyze only this case, when the semipermeable membranes are identical.

Studying the blood speed

The function (28) depends on ξ (which is proportional to the blood speed w) and has no maximum with respect to its variable. Its graph is shown in figure 4.

It can be seen that when the blood speed is 0, the oxygen flow is also 0. Then the oxygen flow increases with the blood speed, although limited by the maximum value $\dot{m}_{max}$ = the oxygen flow which would result by plain diffusion through the two semipermeable membranes stuck one to the other (with no blood between them). The

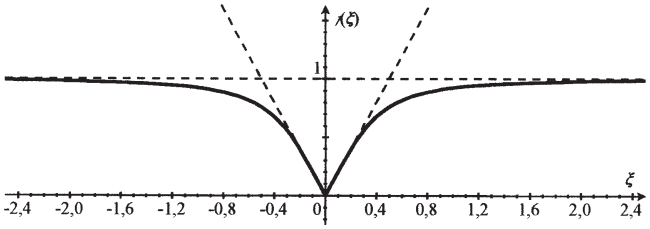


Fig. 4. The function $f(\hat{\imath})$, which (according to (28)) captures the dependency between the Oxygen flow and the reduced blood speed (14)

presence and circulation of blood always lead to an oxygen flow lower than this value – reaching it asymptotically if the reduced speed grows over $\xi \approx 1.5$.

The function $f(\xi)$ is symmetrical with respect to the origin: this means that negative blood speeds lead to the same oxygen flow as positive ones (exactly as expected, since the model works the same if the flow is reversed).

The formula (24) for the oxygen flow shows that it increases directly proportional to $\Delta c_{HL} = c_H - c_L$: so the oxygen flow increases when the atmospheric oxygen concentration c_H increases (when the patient has an oxygen mask applied) or when the oxygen concentration in the interstitial fluid c_L decreases (increased physical effort).

If the blood speed w is small, ξ decreases, the exponentials grow very much, their ratio tends to 1 and the oxygen flow becomes approximately directly proportional to 2ξ (so with the blood speed):

$$\dot{m} \approx \dot{m}_{max} 2\xi = \Delta c_{HL} A w, \quad \text{for } w << \frac{K_p S_{lateral}}{A} \quad (29)$$

This corresponds to the first part of the graph, where one can see the approximately linear dependence if the reduced speed drops below $\xi \approx 0.25$.

This analysis shows that there exists no blood speed maximizing the oxygen flow transported by the cardio-vascular system – if the blood speed increases, so does the oxygen flow (although over some value the increase becomes non-significant, because the oxygen flow reaches a plateau). The optimal blood speed in the cardio-vascular system might be set by constructive factors, such as limiting the pressure differences, limited mechanical strength of the blood vessels etc.

Conclusions

The cardio-vascular-pulmonary system role is to transport oxygen from a higher concentration (in air) to a lower concentration (in the intercellular environment). In other words, it accomplishes something which would have happened naturally, only it does it with greater speed. Because of this, the cardio-vascular-pulmonary system can be studied in a relevant way only within a theoretical framework which takes into account the processes' speeds. The Finite Speed Thermodynamics is exactly the appropriate conceptual framework best suited for such a study.

The blood speed in the cardio-vascular system cannot be optimized for maximizing the oxygen flow, which increases permanently with the blood speed. However, we remark two regimes:

-when the blood speed is low, the oxygen flow is approximately directly proportional to it;

-when the blood speed is high, the oxygen flow is approximately constant (equal to the value it would have

if the transport happened directly through the two adjoined membranes, without any blood between them).

It remains to be determined the actual value of the reduced blood speed for various organisms, to see in which region of the curve in figure 4 they function. This can be done easily, given that the product $K_p S_{lateral}$ from the definition of the reduced speed (14) has been computed for many organs [8, p. 181], and the average blood flow D is also known.

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Nomenclature

A [m^2] = cross-sectional area of a tube
c [kg/m^3] = mass concentration of a gas in a point inside a tube
c_1 [kg/m^3] = Oxygen mass concentration at the entrance of pulmonary capillaries
c_2 [kg/m^3] = Oxygen mass concentration at the entrance of systemic capillaries
c_{ext} [kg/m^3] = gas mass concentration outside a tube
c_{fin} [kg/m^3] = gas mass concentration when exiting a tube
c_{init} [kg/m^3] = gas mass concentration when entering a tube
Δc_{HL} [kg/m^3] = difference between gas mass concentration in the outside environment and in the intercellular environment
D [m^3/s] = fluid (blood) volumetric flow
e (non-dimensional) = natural logarithms' base
γ [m^{-1}] = form factor of a tube (defined in formula 6)
H (as index) refers to pulmonary capillaries
J [$kg/s/m^2$] = gas mass flux passing through the walls of a tube
K [s^{-1}] = constant of proportionality characterizing the diffusion of a gas through a membrane (defined in formula 8)
K_d [m^2/s] = gas diffusivity through a semipermeable membrane
K_p [m/s] = gas permeability through a semipermeable membrane
L (as index) refers to systemic capillaries
$L_{perimeter}$ [m] = circumference of a tube
L_{total} [m] = length of a tube
$\dot{m}$ [kg/s] = gas mass flow entering a tube through its walls
$\dot{m}_{max}$ [kg/s] = maximum Oxygen flow that can be transported by the cardio-vascular system (when there is no blood and the semipermeable membranes are adjoined)
s [m] = thickness of a semipermeable membrane which is the wall of a tube
$S_{lateral}$ [m^2] = lateral surface area
t [s] = time
Δt_{total} [s] = time for a fluid to pass through a tube
V_{total} [m^3] = volume of a fluid inside a tube
w [m/s] = fluid (blood) speed
x [m] = curvilinear coordinate along the axis of a tube
$\eta_{cardiovascular}$ (non-dimensional) = cardio-vascular system's efficiency (formula 27)
ξ (non-dimensional) = fluid's reduced speed (defined in formula 14)

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