# Design of Control Strategies for the $CO_2$ Removal from Blood with an Intracorporeal Membrane Device

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***Abstract*— Currently, available blood $CO_2$ control systems mostly interact with blood extracorporeally, though guiding large blood quantities out of the body is problematic and frequently causes severe complications. Here we propose a set of strategies and a method to control $CO_2$ removal and $CO_2$ concentration change in blood in an intracorporeal membrane based respiratory assistive device using a liquid sweep fluid. Control objectives are defined and concept designs are presented for the respective control loops based on measurement of accessible sensor signals and rotational speed of the blood pump as control variable. The results of this study may contribute to the development of safe and easily controllable $CO_2$ removal systems in patients suffering from respiratory insufficiency.**

***Index Terms*— $CO_2$ control; intracorporeal respiratory assistive device; $CO_2$ removal; membrane device**

## I. Introduction

Blood oxygenators serve as temporary respiratory support. Blood flows through a microporous hollow fiber bundle where $O_2$ is increased and $CO_2$ removed. Usually a gas serves as sweep fluid that flows inside the membrane's hollow fibers and transports the exchange permeates.

Currently available blood $CO_2$ control systems mostly interact with blood extracorporeally, i.e. the blood needs to exit the patient's body only to be released into the systemic circulation after passing the external oxygenator [1].

Standard membrane oxygenators need a surface area of about two square meters to achieve sufficient gas exchange for an adult patient. Consequently, due to the required size, full lung support can only be supplied extracorporeally, meaning that high blood quantities must be guided in and out of the body, which can lead to a number of complications including bleeding, thrombus formation, emboli and infection [1]. Additionally, problematic situations can arise during weaning after the treatment [2].

Intracorporeal devices for support of the respiratory function [3] with smaller surface areas can serve as assistive devices for the pulmonary function and have the advantage that the blood does not need to leave the body. However, on the other hand, the sweep fluid needs to be led in and out of the body to transport the permeates, what may cause severe problems in case of leakage, especially if the sweep fluid is a gas there is the risk of gas embolism.

A different approach is to use a liquid as sweep fluid as e.g. liquid perfluorocarbon (PFC) to sweep the fiber lumens [4]. Using liquid PFC avoids the risk of gas embolism in case of leakage. PFCs have high $CO_2$ solubility of up to 207% (v/v at 25 °C) [5] and are used as blood substitutes in clinical applications [6]. Especially if this liquid is biocompatible with blood the risk of complications in case of leakage can be significantly reduced.

What gets more complicated in case of intracorporeal devices is the control of the system and especially the $CO_2$ concentration in blood and $CO_2$ mass transfer through the intracorporeal membrane.

The aim of this work is to propose a method to control $CO_2$ removal and $CO_2$ concentration change in blood in an intracorporeal membrane based respiratory assistive device using a liquid sweep fluid.

## II. Methods

### A. System

The device consists of an intracorporeal membrane module $M_{int}$ with an integrated micro pump that transports the blood and controls the volume flow of blood through the membrane by modulation of its rotational speed $n_b$. A typical schematic is shown in fig. 1.

The sweep fluid, a liquid, runs in a closed circulation. It flows inside the hollow fibers of the intracorporeal membrane module and transports the removed $CO_2$ out of the body. Outside the body the sweep fluid is depleted from dissolved $CO_2$ in the extracorporeal membrane degasser $M_{ext}$, a pump in the circulation controls the volume flow of the sweep fluid by modulation of its rotational speed $n_s$.

In the extracorporeal membrane module, air flows inside the hollow fibers with a controlled flow rate and regenerates the sweep fluid by removing $CO_2$.

978-1-5386-5724-9/17/$31.00 

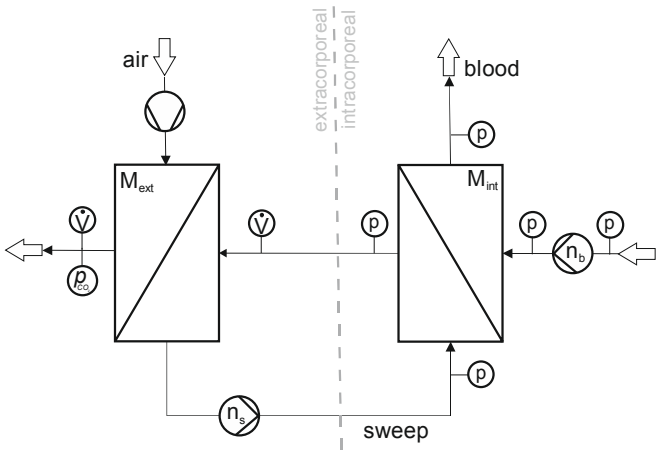



Fig. 1. Schematic of the respiratory assistive device and its functional components

*B. Control principle*

For control of $CO_2$ removal from blood three control objectives are defined with the following parameters:

1) Flow rate of removed $CO_2$ from blood $\dot{V}_{CO_2,blood}$.

2) $CO_2$ concentration in blood at the outlet of the intracorporeal membrane module $C_{CO_2,blood\ OUT}$.

3) Rate of $CO_2$ removal from blood $\frac{d\dot{V}_{CO_2,blood}}{dt}$ or concentration change $\frac{dC_{CO_2,blood\ OUT}}{dt}$.

For ***objective 1***, $\dot{V}_{CO_2,blood}$ is directly calculated from the partial pressure of $CO_2$ in the air flow, measured at the extracorporeal regeneration membrane module outlet, $p_{CO_2,air\ OUT}$, and the known air flow rate $\dot{V}_{air}$:

$$\dot{V}_{CO_2,blood} = \dot{V}_{air} \cdot \frac{p_{CO_2,air\ OUT}}{R.T_{air}} \quad (1)$$

where R is the universal gas constant, and $T_{air}$ is the air temperature.

For ***objective 2***, at steady state, the $CO_2$ balance must close at inlet and outlet of the extracorporeal membrane, giving

$$\dot{V}_{air} \cdot \frac{p_{CO_2,air\ IN}}{R.T_{air}} + \dot{V}_{sweep} \cdot c_{CO_2,sweep\ IN} = \dot{V}_{air} \cdot \frac{p_{CO_2,air\ OUT}}{R.T_{air}} + \dot{V}_{sweep} \cdot c_{CO_2,sweep\ OUT} \quad (2)$$

where the partial pressure of $CO_2$ at the extracorporeal regeneration membrane module inlet, $p_{CO_2,air\ IN} \sim 0$, and the volume flow of the sweep fluid, $\dot{V}_{sweep}$, is measured by a flow sensor.

Additionally, a direct relation exists between the partial pressure of $CO_2$ in the air and the $CO_2$ concentration in the sweep fluid at the outlet of the extracorporeal degasser membrane $c_{CO2,sweep\ OUT}$ assuming equilibrium between gas and liquid phase,

$$p_{CO_2,air\ OUT} = H_{sweep,CO_2} \cdot c_{CO_2,sweep\ OUT} \quad (3)$$

with the Henry's constant $H_{sweep,CO_2}$ for solubility of $CO_2$.

Assuming that the mass transfer process in the intracorporeal membrane is slow, equilibrium in the $CO_2$ concentrations at membrane outlet can be assumed, giving

$$c_{CO_2,sweep\ OUT} = K \cdot c_{CO_2,blood\ OUT} \quad (4)$$

which allows the estimation of $c_{CO_2,blood\ OUT}$ through the equilibrium constant K. K is usually derived from experiments.

For ***objective 3,*** the same relations can be applied to an additional control of the change rate over time. Here the control objective is to obtain setpoint gradients for transient tasks rather than setpoint concentrations or flows.

*C. Resistance coefficient for mass transfer*

A relation for the overall resistance coefficient through the membrane, $k_{tot}$, is derived from the resistances to mass transport through the membrane. A cross section through a typical hollow fiber membrane is shown in fig 2. Mass transport is limited by three resistances in series - the mass transfer resistance $R_{CO_2,blood}$ on the blood side of the membrane, the mass transport resistance $R_{CO_2,membrane}$ inside the wall of the membrane fiber, and mass transfer resistance $R_{CO_2,sweep}$ on the sweep fluid side of the membrane.

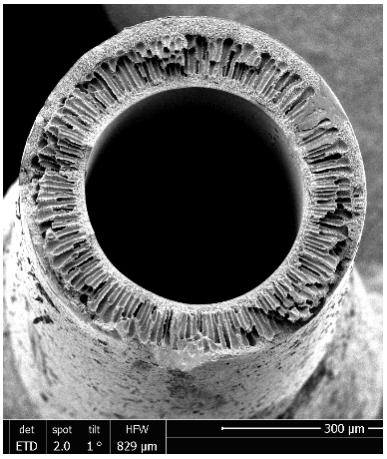



Fig. 2. SEM cross section of a typical hollow membrane fiber

With the resistance coefficients $k_i$ defined as

$$k_i = \frac{1}{R_i} \quad (5)$$

follows

$$k_{tot} = \frac{1}{\frac{1}{k_{CO_2,blood}} + \frac{1}{k_{CO_2,membrane}} + \frac{1}{k_{CO_2,sweep}}}. \quad (6)$$

where $k_{CO_2,blood}$, $k_{CO_2,membrane}$, and $k_{CO_2,sweep}$ represent the $CO_2$ mass transfer coefficents in the blood boundary layer, through the membrane and the sweep fluid boundary layer, respectively.

With the membrane area $A_{membrane}$ the flow rate $\dot{V}_{CO_2}$ through the membrane can be written as

$$\dot{V}_{CO_2} = A_{membrane}.k_{tot}.(C_{CO_2,blood} - C_{CO_2,sweep}) \quad (7)$$

The resistance coefficient $k_i$ for the mass transfer resistances can be derived using the dimensionless Sherwood number Sh representing the ratio of convective to diffusive mass transport. For blood that means

$$k_{blood} = \frac{Sh.D}{d_h}. \quad (8)$$

The diffusion coefficient D accounts for the effective facilitated diffusive transport of $CO_2$ in blood e.g. considering both $CO_2$ and $HCO_3^-$ diffusion as described in [7]. The characteristic length $d_h$ is derived from the membrane fiber packing density, e.g. by

$$d_h = \frac{4.A}{\sum U_i}. \quad (9)$$

where A is the cross section of the flow and $\sum U_i$ are all circumferences of A.

The Sherwood number Sh can be derived from flow, geometric and material properties using the relation

$$Sh = a.Re^b.Sc^c.f(geometry) \quad (10)$$

with the Reynolds number Re representing the ratio of inertial to viscous forces, the Schmidt number Sc representing the ratio of momentum to diffusive transport and the constants a,b,c which can be empirically determined from experiments through multivariable regression. f(geometry) is defined as

$$f(geometry) = \left(\frac{d_h}{L}\right)^e \quad (11)$$

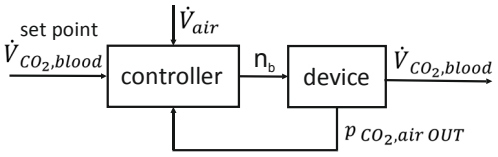
with fiber length L.

## III. Results

### *A Main control loops*

Figure 3 shows the control loops for the whole device including both membranes for control objectives 1 and 2. Air flow rate and the set point of the control parameter are inputs, the controllers determine the required blood pump rotational speed, and from the measured values - $p_{CO_2,air\ OUT}$ and for criterion 2 also $\dot{V}_{sweep}$ – the controller determines the real parameter value and adapts the required blood pump rotational speed accordingly. As controllers standard PI or PID controllers are implemented.

*a)*




*b)*

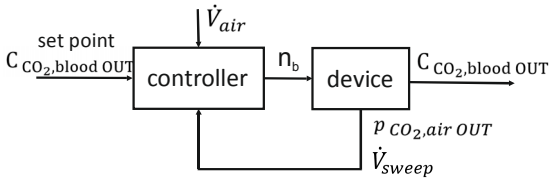



Fig. 3. Schematic of control circuits for a) objective 1 and b) objective 2. As controllers standard PI or PID controllers are implemented.

### *B Control of the intracorporeal membrane*

Assuming co-current flow through the membranes, the following dependencies can be formulated:

- Flow rates of blood and sweep fluid through the intracorporeal membrane are linked by the corresponding hydraulic pressure differences between membrane module in- and outlet on the blood side $\Delta p_{blood,M_{int}}$, and the sweep fluid side $\Delta p_{sweep,M_{int}}$. The pressure differences must be equal for stability of the fibers and to avoid damages that can lead to mass flows across the membrane wall.

$$\Delta p_{blood,M_{int}} = \Delta p_{sweep,M_{int}} \quad (12)$$

- The pressure difference in the blood flow across the membrane is a function of the rotational speed of the pump, as shown in fig. 4. With the experimentally determined characteristic behavior $f(n_b)$, measurement of pressure differences across both membrane and pump, $\Delta p_{blood,pump}$, can also be used to detect any problems as e.g. clotting in the blood flow.

$$\Delta p_{blood,M_{int}} = f(n_b),\ \Delta p_{blood,pump} = f(n_b) \quad (13)$$

- The pressure difference in the sweep flow across the membrane is a function of the sweep flow rate which is controlled by the rotational speed of the sweep flow pump, $n_s$. These relations can be experimentally determined for the established circulation loop, with eqns. 12 and 13 this gives

$$n_s = f(\Delta p_{blood,M_{int}}) = f(n_b) \quad (14)$$

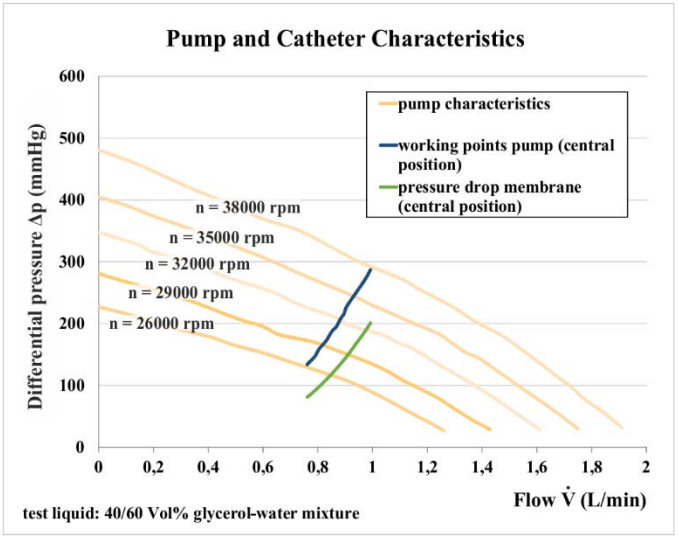



Fig. 4. Blood pump characteristics and resulting pressure differences across pump and membrane

With these dependencies the control loop for the intracorporeal membrane can be integrated into the main control loops, as shown in fig. 5 for control objectives 1 and 2.

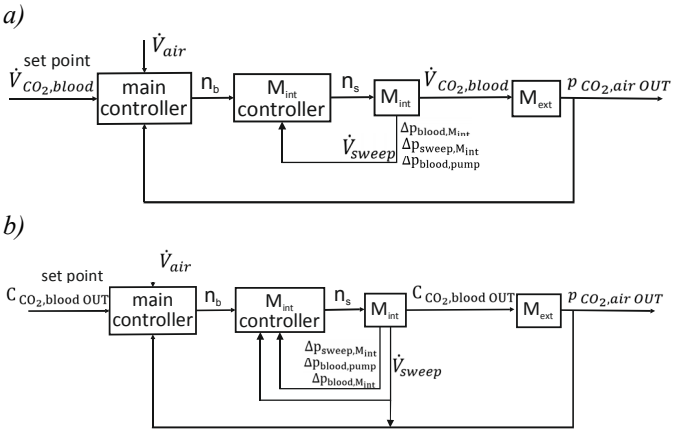



Fig. 5. Schematic of the complete control circuit for a) control objective 1 and b) control objective 2. As controllers standard PI or PID controllers are implemented.

## IV. Discussion

For clinical application, a control strategy for the $CO_2$ removal is essential. Among other key design features of such a controller, robustness and simple calibration based on physical parameters is favorable compared to other approaches such as soft sensing, black box concepts etc. In the concept design presented in this paper we have kept the number of sensors at a low limit. Pressure sensors, flow sensors and $CO_2$ concentration measurement are essential and the signals can also be used for alarm and failure management which was not presented here. The control loops needed are simple, yet effective as they act directly on the $CO_2$ mass transfer from blood to the transfer liquid and the regeneration cycle. So far, this study is theoretical, as a next step we are going to implement a dynamic simulation model to derive suitable PI/PID parameters and to test various control situations based on the objectives 1, 2 and 3 proposed in this paper.


## Acknowledgment

Research partly supported by the Austrian Research Promotion Agency (FFG), project 'LiquiClear', project number 850713 and project 'MILL – Minimal Invasive Liquid Lung' project number: 857859. We acknowledge Simone Spitzer (TU Wien) for the preparation of the SEM picture.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