PERFUSION SYSTEM CONTROLLER STRATEGIES DURING AN ECMO SUPPORT

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ABSTRACT

In this work modelling and control of Perfusion system is presented. The Perfusion system simultaneously controls the partial pressures during Extra Corporeal Membrane Oxygenation (ECMO) support. The main Problem in ECMO system is exchange of Blood Gases in the Artificial Lung (Oxygenator).It is a highly Nonlinear Process comprising time-varying parameters, and varying time delays, it is currently being controlled manually by trained Perfusionist. The new control strategy implemented here has a feedback linearization routine with time-delay compensation for the Partial pressuresof Oxygen and Carbon dioxide. The controllers were tuned robustly and tested in simulations with a detailed artificial Lung (Oxygenator) model in Cardiopulmonary bypass conditions. This Automatic control strategy is proposed to improve the patient's safety by fast control reference tracking and good disturbance rejection under varying conditions.

KEYWORDS

Cardio Pulmonary Bypass (CPB), Blood Gas Analyser (BGA), Extra Corporeal Membrane Oxygenator (ECMO)

1. INTRODUCTION

ECMO support has been established as a routine treatment used for the patients whose Heart or Lungs is not working properly and is most often used for Bypass procedures, and Heart transplantations. The ECMO procedure is similar to a Heart-Lung bypass used during open-Heart surgery. Extracorporeal Membrane Oxygenation can take over the function of the child's Heart and / or Lungs for a limited time until the child recovers from the initial cause of the failure [4]. Figure 1 shows a new born having respiratory failure is placed in ECMO setup in ICU. The most frequent use for ECMO has been with new born infants with lack of a fully functioning respiratory system or other birth defect,although it may also be helpful in selected cases of severe heart failure in adults and children. During CPB the functions of the Heart and Lungs are taken over by the ECMO setup there by guaranteeing a bloodless, motionless operating field. In Extra Corporeal Membrane Oxygenator the blood is circulated from the venous to the arterial side of the patient's vascular system by a pump in the machine. Additionally an Oxygenator takes over the function of the Lungs by the addition of fresh Oxygen (O_2) to and the removal of Carbon dioxide (CO_2) from blood. This Gas exchange is achieved by a gas flow that is in close contact to the Blood pumped through the Oxygenator. This Gas mixture and its partial gas pressures initiate a diffusion process in the Oxygenator, which in turn determines the amount of Oxygen and Carbon dioxide on the arterial side of the patient. The Gas flow and the mixture of Gas that flows

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through the Oxygenator are currently still being manually adjusted to have correct partial gas pressures in the arterial blood [5].

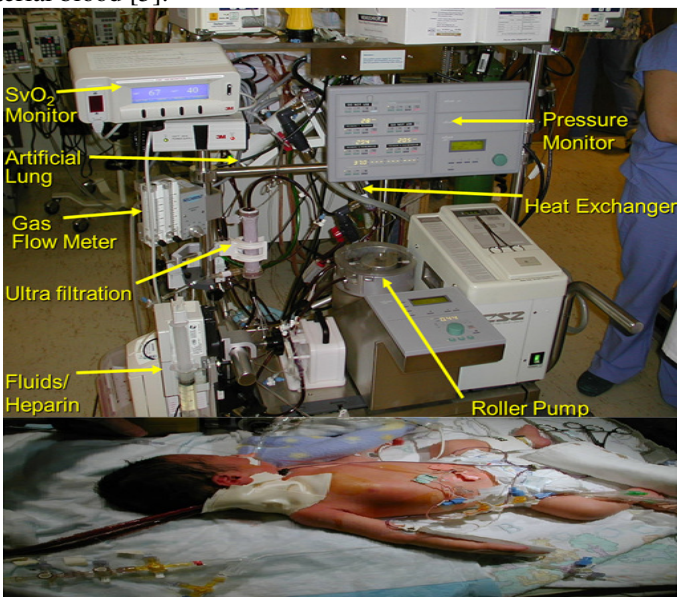

Figure 1. ECMO Setup

This manual control Process can lead to errors, which may cause tissue or nervous cell damage. An automatic blood gas control strategy is suggested to improve the patient's safety by fast control reference tracking of reasonable set points and a good disturbance rejection, which means keeping the blood gas values in physiological ranges during disturbances. In addition to that, the automatic control is suggested to remove workload from the perfusionist staff, and may thereby help to increase the amount of correct decisions in situations of extreme workload.Figure 2 shows the circuit diagram of an ECMO setup.

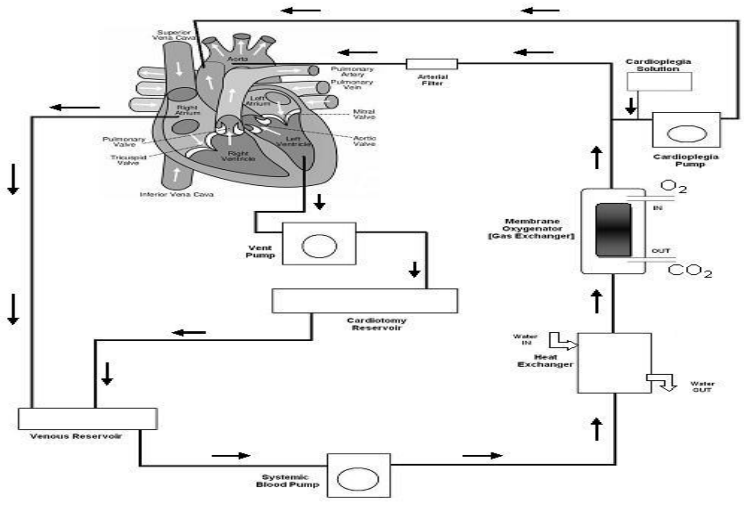

Figure 2.Complete Circuit Diagram of ECMO

2. ECMO MODELLING

As the exchange and transport of oxygen and carbon dioxide are complex biological processes Comprising nonlinearities, time delays, uncertainties, and time-varying parameters, detailed system knowledge is needed for the development of a suitable feedback control [1]. For that reason a model of the blood gas transfer process was developed as follows. The system subject to control was divided into three subsystems: the gas mixer; the Oxygenator, and the Blood Gas analyzer (BGA) as shown in thefigure 3.

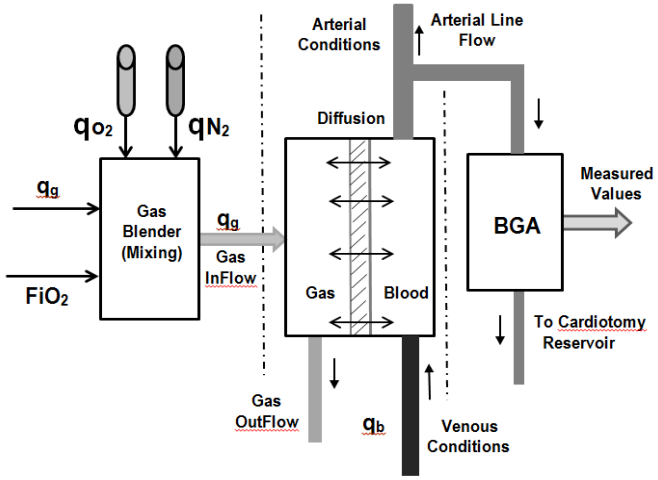

Figure 3. Gas Blender, Oxygenator, and BGA diagram

2.1. Gas Blender Model Design

As the gas concentrations are normally set mechanically with flow tubes, an electronic gas mixer was designed based on electronic flow dosage modules.Figure 4 shows the Gas blender, Oxygenator, and Blood gas analyser model diagram. The static output of the gas mixer is given by

$$Q_G = Q_{O_2} + Q_{N_2} + Q_{CO_2} \quad (1)$$

$$Q_{O_2} = Q_G * Fi_{O_2} \quad (2)$$

$$Q_{CO_2} = Q_G * Fi_{CO_2} \quad (3)$$

$$Q_{N_2} = Q_G\,(1-Fi_{O_2}-Fi_{CO_2}) \quad (4)$$

The combination of the two gases is known as inspired Input Oxygen (FiO_2).Q_G is nothing but total gas flow of the cardiopulmonary Blood Gases. Q_G is measured by blood Gassensor.$Q_{CO_2}$is nothing but total flow of carbon dioxide.Q_{N_2} is nothing but total flow of atmospheric air.$T_{d1}(T_g)$is a transport delay due to the distance from gas blender to oxygenator and applies to the Oxygen fraction in the gas.The time-delay depends on gas flow and the tubing system. Total time delay for the Gas mixer process based on the equation [1] - [2].

$$\mathbf{T_{d1}(T_g)} = \frac{\mathbf{0.7853}}{\mathbf{T_g}} \mathbf{d^2_{t,oxy} l_{t,oxy}} \qquad (\mathbf{5})$$

Where $\mathbf{d^2_{t,oxy}}$ is the square diameter of the Gas Blender,and $\mathbf{l_{t,oxy}}$is the length of the Gas Blender to Membrane Oxygenator,$\mathbf{T_g}$is the Total blood flow in the Membrane Oxygenator.

2.2. Membrane Oxygenator Model

The Membrane Oxygenator replicates the functionality of the Human Lungs in that it exposes the blood to regulate amount of O_2 and CO_2. Gas entering on top inside the fibres while the blood is flowing first down through a Heat Exchanger and then up through the Gas Exchanger. Gas Exchange between Gas and blood phase is a very complicated process involving position dependent quantities and partially nonlinear mechanisms which include Gas transport and diffusion in the Blood or the diffusion across the Membrane. [3]Mass Balance Equation for Gas Stored in the Blood Compartment in membrane Oxygenator.

$$V\frac{d[c]_i}{dt} = T_b\left([C]_{i,in} - [C]_{i,out}\right) + D_i\left(p_{i,ext} - p_i\right) + Ri \qquad (6)$$

Where V is the Volume of the compartment mixing chamber in an Oxygenator, $\mathbf{[C]_i}$is the Concentration of the Components,$\mathbf{[D]_i}$ is the Diffusion Capacity over the Membrane Oxygenator, $\mathbf{[p_i]}$is the internal and external Partial Pressure of the Membrane Oxygenator, **R**i is the external disturbance in membrane Oxygenator cross bonding to chemical reaction .$\mathbf{T_b}$is the Total Blood flow through in the Membrane Oxygenator.

2.3.Blood Gas Analysis Model Design

Blood gases are only sampled a few times a day in order to check blood gas status. This clinical blood gas analyser's use up a few millilitre bloods for each sample, so these devices cannot be used for continuous blood monitoring on a frequent basis (e.g. every second). For our setup we chose the CDI 500 blood gas analyser from Terumo which offers online measurement of pH, pCO_2, pO_2, potassium, temperature and etc. The BGA was modelled according to measurements with first order differential equations with its dominant. The modelled BGA time-delay is

$$T_{d_2}(T_g) = \frac{0.7853}{b_{t1}(0.01\,T_b + b_{t0})} d^2_{t,BGA} l_{t,BGA} \qquad (8)$$

Where $d_{t,BGA}$ is the diameter of the Blood Gas analyzer tube , $l_{t,BGA}$ is the length of the of the Blood Gas analyzer tube,$\mathbf{b_{t0}}$ is the time delay for the Blood Gas Analyzer, $\mathbf{b_{t1}}$is the Conversion factor for Blood Gas Analyser, $\mathbf{T_b}$ is the total blood flow through the Blood Gas Analyser .

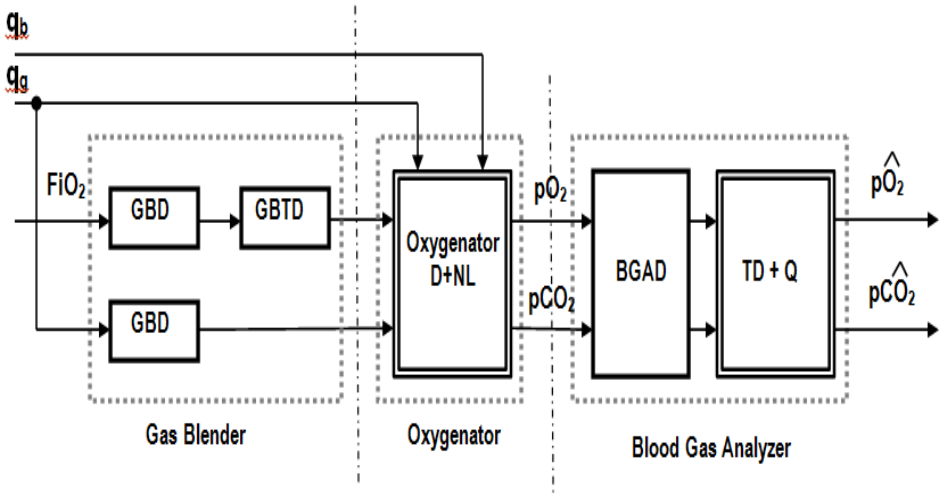

Figure 4. Gas blender, Oxygenator, and Blood Gas Analyser modelDiagram

2.4. Time Delay

Total time delay of the blood flow can be calculated by time delay between the Gas blender to Membrane oxygenator $T_{d_1}(T_g)$ and time delay between membrane oxygenator to Blood Gas analysis $T_{d_2}(T_g)$.

$$T_d(T_g) = T_{d1}(T_g) + T_{d2}(q_g) \quad (9)$$

2.5. Input/OutputLinearization of O_2

The process is linearized with an analytical input/output linearization for the oxygenation process with input FiO_2 and output pO_2. Direct disturbance to the states can be measured with high dynamics in case of the blood-flow which are measured via the blood-gas analyser the disturbance dynamics are expected to be much slower than the dynamics of the blood gas analyser. So the input oxygen from the Gas blender to membrane Oxygenator as follows.

$$u_{oxy} = fiO_2 * p_{bar} \quad (10)$$

The input and output of the oxygen can be linearized with the help of barometric pressure and percentage of inspired input Oxygen depends on the patient respiration.

2.6. Nonlinear State Space Process Model

Process model depends on the Gas blender, membrane oxygenator, blood gas analysis and so on. Finally we have to derive the mathematical model for Gas transfer in an oxygenator. So the strong mathematical knowledge should be needed to design the membrane oxygenator model.[3] Initially the model design and control by single input single output (SISO) Processes the may be defined

$$\dot{X}(t) = A\,x(t) + Bu(t) \quad (11)$$
$$Y(t) = C\,x(t) + D\,u(t) \quad (12)$$

Where x(t) is the state vector; A is the system matrix;B is the input matrix; C is the output matrix is zero matrix; y(t) is the output vector ; u(t) is the input control matrix ;[1]

The continuous time invariant state space model for an membrane oxygenator to control the blood gases with the help of controller .The controller approach is based on the mathematical nonlinear model of the gas transfer process in membrane oxygenator. Input control matrix of the single input as followsas

$$B^T u = \left[0\ 0\ 0\ 0\ 0\ 0\ 0\ 0\ 0\ 0\ 0\ 0\ \frac{q_g}{v_g}\right](u)(13)$$

Input vector u = pO_2 = FiO_2 * 760 mm Hg for barometric pressures[3]. The state vector may be defined as

$$X = [x_1 x_2 x_3 x_4 x_5 x_6 x_7 x_8 x_9 x_{10} x_{11} x_{12} x_{13}]^T u \qquad (14)$$

The full process model of the three inputs, two output of the system with input composed of **FiO_2**, Gas flow and Blood flow the output vector is based on pressure of **O_2**and **CO_2**

$$Y^T = [pO_2 pCO_2]^T \qquad (15)$$

3. CONTROLLERDESIGN

The pCO_2 process is much less critical in control terms; compared with the pCO_2 process need not Linearization of the carbon dioxide for pCO_2 controller design.pO_2 process is more complicated in control terms. An input/output state linearization routine with delay-time compensation and an external linear gain scheduled controller had to be developed to handle the strong static and dynamic nonlinearity and the time delay at input and output. Like the pO_2 controller was gain-scheduled in dependence of blood flow through the oxygenator.[1] The PID-pO_2 and the PID-pCO_2 controllers, were discretized at the sampling time of the blood gas analysis, Ts, BGA = 6 s.

3.1. PID Controller

The Proportional-Integral-Derivative (PID) method is a kind of feedback controller which is generally based on the error (e) between desired set point and measured process value. Three parameters must be designed in the PID Controller and each parameter has an effect on the error. The transfer function of the PID controller is writtenas

$$C(S) = \frac{U(S)}{E(s)} = k_p + \frac{K_i}{S} + K_d \qquad (16)$$

4. DISCUSSI ON

By using a detailed model of the gas blender, the Oxygenator, and the Blood gas analysis, a feedback control could be developed for the simultaneous control of Oxygen and Carbon dioxide partial pressures. As the control input for pO_2, the oxygen fraction in the gas flow FiO_2 to the oxygenator was used. As the control input for pCO_2, the total gas flow qg to the Oxygenator was

used. The direct coupling of q_g to the FiO_2 control input is similar to a disturbance at pCO_2 controller gas flow input changes and can be seen as a disadvantage. The advantages of this strategy are that an additional CO_2 gas supply is saved and that this technique is commonly used in most cardiovascular Heart surgery centres. The pO_2 process is more complicated in control terms. An input/output state linearization routine with delay-time compensation and an external linear gain scheduled controller had to be developed to handle the strong static and dynamic nonlinearity and the time delay at input and output. In contrast to that, a PI controller was developed for the O_2 process, as it shows an output delay time only and much less static and dynamic nonlinearities. The controllers were tested in simulations over the whole operating range with the model and incorporated uncertainty and showed a good performance and robust stability

5. SIMULATION RESULTS

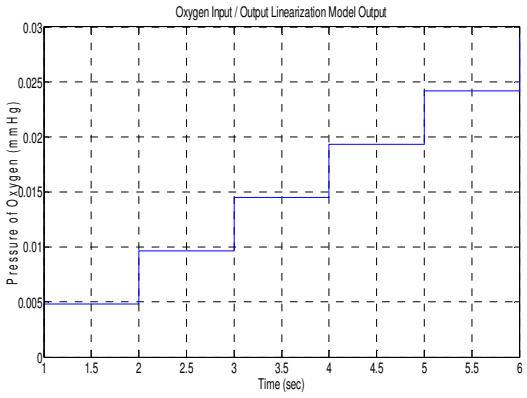

Figure 5. A. Oxygen linearisation I/O Response

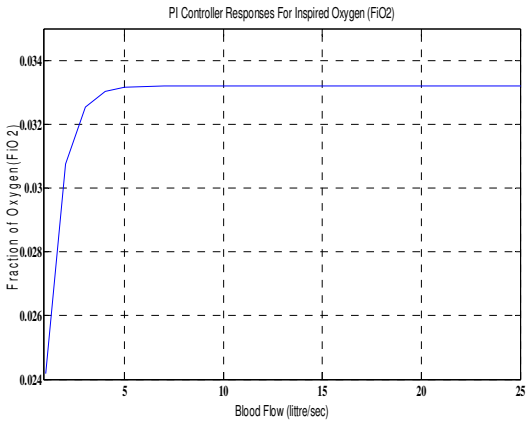

Figure 6. Fraction of Oxygen (FiO2) Control Response

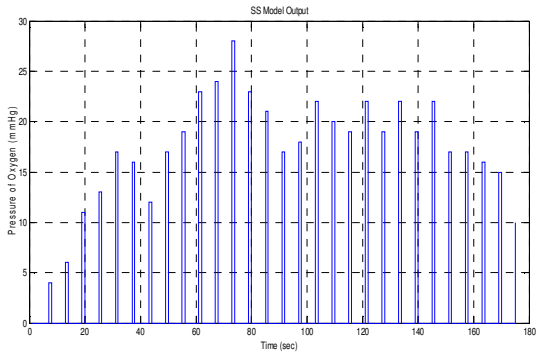

Figure 7. State Space output for pressure of oxygen

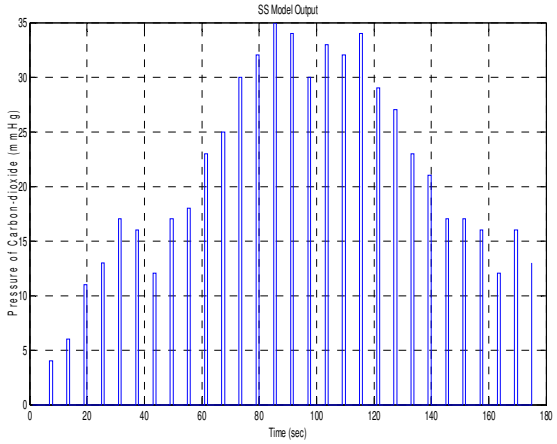

Figure 8. State Space output for pressure of carbon dioxide

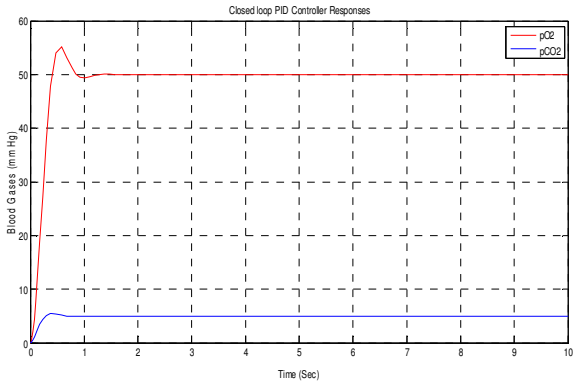

Figure 9. PID Controller Responses

6. CONCLUSION

The proposed model of the blood-gas controller for Oxygen and Carbon dioxide exchange during ECMO support shows fast reference tracking and good disturbance rejection results, while process coupling is kept low. This is possible because of the Oxygen process I/O linearisation and time-delay compensation. Linear process models for the two-input- two output plant were

obtained for the linearised system without time-delay over the complete operating range during surgeries. Uncertainties in the resulting transfer functions, as well as time delay uncertainties were modeled in a multiplicative output uncertainty bound. Only one linear controller was applied in nonlinear simulation with a validated Oxygenator model and showed very good results However, lumping uncertainties of various sources at the system output in a multiplicative uncertainty bound is suggested to yield conservative controller result.

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