ADAPTIVE CONTROL OF RESPIRATORY MECHANICS

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INTRODUCTION

Hierarchy and adaptivity are typical characteristics of physiological control systems. In order to gain a deeper insight into the function of the regulators in man, the systemic approach and mathematical modelling have to be used. Large scale human systems are often divided into seemingly independent, in reality highly interconnected, blocks. When examining these blocks it is impossible to control all the relevant inputs and outputs simultaneously under experimental conditions. The only way to try to cope with the increasing number of variables in dynamical subsystems is to build a simulation model. Simulation with black box models cannot, however, explain the ability of the system to learn that is to adapt itself to new circumstances. To account for this adaptivity the model must include some kind of a performance criterion, a natural choise is a minimization criterion. The system learns the new circumstances through a continuously active identification or self-optimizing process correcting the operation of the system so that it would minimize the objective-functional in question under the possible constraints on the control variables.

The mechanical breathing process is a typically adaptive system, whose main purpose is to meet with the ventilatory demand determined by the chemical state of the blood. The desired ventilation level can, however, be produced in many ways,thus a decision is needed to choose between the different alternative control strategies for the mechanical system. Our work is concentrated in the study of the adaptive nature of the system. Its performance principles are being cleared up with the aid of mathematical modelling. The models we have developed will be dealt with in this paper.

MODELLING THE ADAPTIVITY

The early work of Rohrer(6) and Otis et al.(5) were based on the minimum respiratory work rate criterion. They used a sinusoidal air flow shape and found that there existed an optimal breathing frequency. Mead(4) showed that a minimum force amplitude criterion would give better results. Further development was made by Widdicombe and Nadel(8), who included control of airway dimensions in the former models and obtained an optimal dead space volume with respect to both criteria. There has been only one attempt(7) to find criteria, which could explain the shape of the respiratory air-flow curve. The authors ended up with minimum mean squared acceleration as a suggested performance function for humans. In a closed form their solution is a parabolic curve for both inspiration and expiration and it does not deviate much from a sine-curve. None of the previous models have been able to predict any changes in the individual respiratory variables of test subjects. In these papers the aim has merely been to prove that breathing is optimal with regard to minimum effort or maximum gas exchange. Our main hypothesis is on the contrary that the system is adaptive, and so it is also supposed that it must have some criterion for decision making in adaption. When such subprocesses as breathing are concerned the criterion is probably not so clear and general as the principle of minimum energy expenditure, although the energy cost seems to be an essential component in it. It should be noted that in large scale systems partial optima need not necessarily give the total optimum. Further it is very difficult to define precisely all the tasks that different subsystems may perform and yet these tasks should be reflected in the corresponding performance criteria. At the first stage we have tried to find a physiologically meaningful performance criterion expressed in mechanical entities, which could successfully explain the spontaneous breathing of test subjects. To reach this it has been necessary to extend the number of variables studied and to make the assumptions less restrictive and more detailed. In the earlier studies for instance the differences in the duration of inspiration and expiration and the active role of muscles in the expiratory phase were neglected. Moreover as far as we know this is the first time that a model is used to study the control of rib cage and abdomen with respect to adaptivity.

A HIERARCHICAL MODEL

In the model the control of respiratory mechanics is thought to be built up of two levels. The lower level criterion determines the flow pattern for the cycle with the durations and tidal volume fixed, while the higher level criterion gives the overall pattern of breathing satisfying the alveolar ventilation demand ($\dot{V}_A$). The overall pattern is described in the model by duration of inspiration (t_1), expiration (t_2) and pause (t_3), the operating level (V_0)(the change of FRC-level from the equilibrium value) and tidal volume (V_T) and also the size of the airways (V_D)(see figure 1). The concept of preprogramming of each cycle during normal breathing is an inherent assumption in the model. It does not matter yet nothing is known about the existence of the corresponding neural identification networks, since the primary aim is first to explain and understand the observed control principles.

The system is assumed to have only one degree of freedom and a first order linear differential equation is used to describe the system dynamics. It is thought that the 'real' decision criterion in the respiratory control system is associated with the effectivity of oxygen transport and with the minimization of oxygen consumption. The $P\dot{V}$ and P^2 terms in the objective functions correspond directly to the oxygen cost of breathing, whereas the $\ddot{V}^2$ terms correspond to the gas exchange and to the efficiency of muscle shortening. The third and the fourth terms in the higher level objective-function account for the positive effects of resting, which may be related to the efficiency of the muscles or to the general comfortability. The weighting coefficients α_i and β_i that appear in the cost functions are constant individual parameters. It is clear that structural and other personal characteristic besides the system equation parameters must be taken into account in the model when we try to describe a universal biological functioning principle with the aid of mechanical entities. The lower level objective function to be minimized is

$$J^{II}=\begin{cases} J_1^{II}=\int_0^{t_1}(\ddot{V}^2(t)+\alpha_1 P(t)\dot{V}(t))dt, & \text{for inspiration} \\ J_2^{II}=\int_{t_1}^{t_1+t_2}(\ddot{V}^2(t)+\alpha_2 P^2(t))dt, & \text{for expiration} \end{cases}$$

It is assumed that the system equation

$$P(t) = K\cdot V(t) + R(V_D)\cdot\dot{V}(t)$$

does not change during the cycle. The size of the airways V_D is kept at a constant value, which is determined on the higher level. The end expiration level, where the inspiration starts is V_0 and the corresponding boundary condition to the problem

$$V(0)=V(t_1+t_2)=V_0 \;,\; V(t_1)=V_0+V_T \;,\; \dot{V}(0)=\dot{V}(t_1)=\dot{V}(t_1+t_2)=0$$

are also dependent on the higher level minimum. Moreover there is no flow during the pause period, $\dot{V}(t)=0$ for $t\varepsilon[t_1+t_2,t_1+t_2+t_3]$. In the following the first level solution will be denoted by $V^*(t) = =V^*(t;t_1,t_2,V_0,V_D,V_T)$.

The higher level objective function is

$$J^I(t_1,t_2,t_3,V_D,V(t)) =$$

$$=\frac{1}{t_1+t_2+t_3}[\beta_1\int_0^{t_1+t_2+t_3}P_i^2(t)dt+\beta_2\int_0^{t_1+t_2+t_3}P_e^2(t)dt+$$

$$+\beta_4\int_0^{t_2}dt+\beta_5\int_0^{t_1}\ddot{V}^2(t)dt+\beta_6\int_{t_1}^{t_1+t_2}\ddot{V}^2(t)dt],$$

where P_i= inspiratory muscle pressure and P_e= expiratory muscle pressure. Constraints to this problem are the systems equation, the optimality of the airflow $V(t)=V^*(t)$ and the alveolar ventilation equation $\dot{V}_A=(V_T-V_D)/(t_1+t_2+t_3)$, where $\dot{V}_A$ is a given constant for each breath. The total non-elastic resistance $R(V_D)$ has been divided into airway resistance R_A, lung tissue resistance R_{LT} and thoracic wall resistance R_{TW} so that $R(V_D)=R_A(V_D)+R_{LT}+R_{TW}$. The relationship between anatomical deadspace and airway resistance is derived in the manner presented by Widdicombe and Nadel[8] giving

$$R_A(V_D) = \frac{0.082 \text{ cm } H_2Os}{V_D - 0.0825 \text{ l}} + 0.4 \text{ cm } H_2Os/l$$

The model is presented as a two level hierarchical system in figure 1, where the coordination variables are t_1,t_2,V_0,V_D,V_T and $V^*(t)$. The lower level problem is solved analytically with the classical variational techiques. At first sight the upper level also looks like a dynamical optimization problem yet it is a static problem because of the trajectory constraint $V(t)=V^*(t)$.

Testing of the overall results of the model is being carried out at the moment. Only preliminary comparison can be made between the model and the observed values for t_1,t_2 and t_3, which shows good agreement (table 1). The resemblance of the theoretical and individual experimental flow curves is also great and the typical flattening effect of resistive loading is demonstrated in the flow pattern solutions[9].

First notes on the model and some experimental results can be found in ref.(9), a more detailed report[3] is still under preparation.

THE LOWER LEVEL MODEL WITH TWO DEGREES OF FREEDOM

Although movements of the respiratory apparatus are generally described only with one degree of freedom, in fact a change in the lung volume is a result of the independent movements of rib cage and abdomen so the system actually has two degrees of freedom (at least)[1]. The lower level performance criterion, which determined the flow pattern of breathing, can also be applied with this extended system model to give optimal control strategies separately for the chest wall and the abdominal muscles. The system consists now of the two parallel components, rib cage and abdomen, connected in series with the lung component and it is described by a set of linear first order differential equations.

$$P_{rc}(t)-P_{pl}(t) = K_{rc}V_{rc}(t)+R_{rc}\dot{V}_{rc}(t)$$

$$P_{ab}(t)-P_{pl}(t) = K_{ab}V_{ab}(t)+R_{ab}\dot{V}_{ab}(t)$$

$$P_{pl}(t) = K_l V_l(t)+R_l\dot{V}_l(t),$$

where P_{pl} is the change from the resting pleural pressure difference with respect to atmosphere, P_{rc} and P_{ab} are the pressures produced by the chest wall muscles and by the diaphragm and the abdominal muscles. The elastances K_{rc},K_{ab},K_l and the damping factors R_{rc},R_{ab},R_l are assumed constant. Reference values for the relative volume changes V_{rc},V_{ab},V_l are zero at the equilibrium (FRC) level. The volume changes of the lungs are connected to those produced by the rib cage and the abdominal movements by the constraint equation $V_l(t)=V_{rc}(t)+V_{ab}(t)$. It is assumed that at the beginning of inspiration and at the end of expiration the system is in a stationary position at level V_0 so that neither rib cage nor abdominal volume is changing. This gives us the following boundary conditions

$$V_{rc}(0)=V_{rc}(t_1+t_2)=K_{ab}V_0/(K_{rc}+K_{ab}) \;,$$

$$V_{ab}(0)=V_{ab}(t_1+t_2)=K_{rc}V_0/(K_{rc}+K_{ab}) \;,$$

$$\dot{V}_{rc}(0)=\dot{V}_{rc}(t_1+t_2)=\dot{V}_{ab}(0)=\dot{V}_{ab}(t_1+t_2)=0$$

Further at the end of inspiration the tidal volume V_T is reached and the total flow is zero yielding the transversality conditions

$$V_{rc}(t_1)+V_{ab}(t_1)-V_0=V_T \text{ and } \dot{V}_{rc}(t_1)+\dot{V}_{ab}(t_1)=0 \;.$$

The lower level objective function will in this case be of the form

$$J^{II}=\int_0^{t_1+t_2}L(V_{rc},\dot{V}_{rc},\ddot{V}_{rc},V_{ab},\dot{V}_{ab},\ddot{V}_{ab})dt \text{ where}$$

$$L=\begin{cases} L_1=P_{rc}\dot{V}_{rc}+P_{ab}\dot{V}_{ab}+\gamma_{11}\ddot{V}^2_{rc}+\gamma_{12}\ddot{V}^2_{ab}+\gamma_{13}(\ddot{V}_{rc}+\ddot{V}_{ab})^2 , \\ \text{for inspiration } t\varepsilon[0,t_1] \\ L_2=P^2_{rc}+P^2_{ab}+\gamma_{21}\ddot{V}^2_{rc}+\gamma_{22}\ddot{V}^2_{ab}+\gamma_{23}(\ddot{V}_{rc}+\ddot{V}_{ab})^2 , \\ \text{for expiration } t\varepsilon[t_1,t_1+t_2] \end{cases}$$

Here it has been more convenient to place the personal weightings γ_{ij} in a different position than the corresponding weightings α_i. It can be shown that the formulated model has a unique solution when not more than one of the coefficients γ_{1j} and γ_{2j} is zero.

Results from the model, for example in figure 2, correspond quite well to the flow and pressure curves given in ref.(1). The effect, that lowering the compliance of either part results in a smaller contribution to the tidal volume due to that part, that is observed in experiments (1), is also found in the model (2).

SUMMARY

Previous models of the adaptive control of respiratory mechanics have been inadequate to predict the respiratory pattern in detail, as the main variables that have been treated are respiratory frequency and dead space volume. A new hierarchical approach has been reported, which leads to significantly better understanding of the control principles of respiratory mechanics. The independent control variables studied are inspiratory time, expiratory time, pause period, change of FRC-level, airway size and the flow pattern of the cycle. Movements of the rib cage and the abdomen and their contribution to the tidal volume are also treated.

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REFERENCES

(1) CAMPBELL,E.J.M.,AGOSTONI,E. and NEWSOM DAVIS,J.: The Respiratory Muscles Mechanics and Neural Control second edit.,Lloyd-Luke Ltd.,London, 1970.

(2) HÄMÄLÄINEN,R.P. and RANTA,J.:(paper under preparation).

(3) HÄMÄLÄINEN,R.P. and VILJANEN,A.A.:(paper under preparation).

(4) MEAD,J.: Control of respiratory frequency. J.Appl. Physiol. 15:3, 1960, pp. 325-336.

(5) OTIS,A.B., FENN,W.O. and RAHN,H.: Mechanics of breathing in man. J.Appl.Physiol. 2, 1950,pp.592-607.

(6) ROHRER,F.: Physiologie der Atembewegung. in: Handbuch der Normalen und Patologischen Physiologie, vol.2, ed. BETHE,A. et al., Springer Berlin, 1925 pp. 70-127.

(7) YAMASHIRO,S.M. and GRODINS,F.S.: Optimal regulation of respiratory airflow. J.Appl.Physiol. 30:5, 1971, pp. 597-602.

(8) WIDDICOMBE,J.G. and NADEL, J.A.: Airway volume, airway resistance and work and force of breathing: theory. J.Appl.Physiol. 18:5, 1963, pp. 863-868p

(9) VILJANEN,A.A.: Coordination of neuromuscular effector systems in ventilation. Proceedings of the International Symposium on Respiratory and Laryngeal Control Systems, London, January 1972 (arranged by the Royal College of Surgeons of England).

	t_1[s]		t_2[s]		t_3[s]		V_0[l]		V_D[l]		V_T[l]	
	obs.	model	obs.	model	obs.	model	obs.	model	obs.	model	obs.	model
SUBJECT AP. $K = 9.7$ cm H_2O/l $\alpha = (2.9, 0.07)$ $\beta = (1,30,7.4,5.2,0.9)$												
normal breathing	1.70	1.65	2.50	2.54	0.14	0.13	-	0.002	-	0.148	0.70	0.699
increased ventilation I	1.65	1.64	2.50	2.42	0.10	0.00	-	0.004	-	0.154	0.85	0.823
increased ventilation II	1.60	1.61	2.40	2.39	0.04	0.00	-	0.006	-	0.158	0.90	0.898
SUBJECT JE. $K = 5.8$ cm H_2O/l $\alpha = (3.0, 0.35)$ $\beta = (1,30,7.0,1.9,0.01)$												
normal breathing	1.50	1.48	1.90	2.17	0.30	0.34	-	0.017	-	0.173	0.77	0.842
increased ventilation I	1.40	1.43	1.90	2.04	0.00	0.00	-	0.030	-	0.184	0.95	1.024
increased ventilation II	1.30	1.39	1.70	1.93	0.00	00.00	-	0.046	-	0.196	1.10	1.249

TABLE 1. Comparison of model and observed values for two subjects. The increase in ventilation was produced by adding dead space loads of 150 ml and 300 ml. The sum of lung tissue and thoracic wall resistances was taken to be 2.0 cm H_2Os/l. Alveolar ventilation demand is in each case estimated from the minute ventilation.

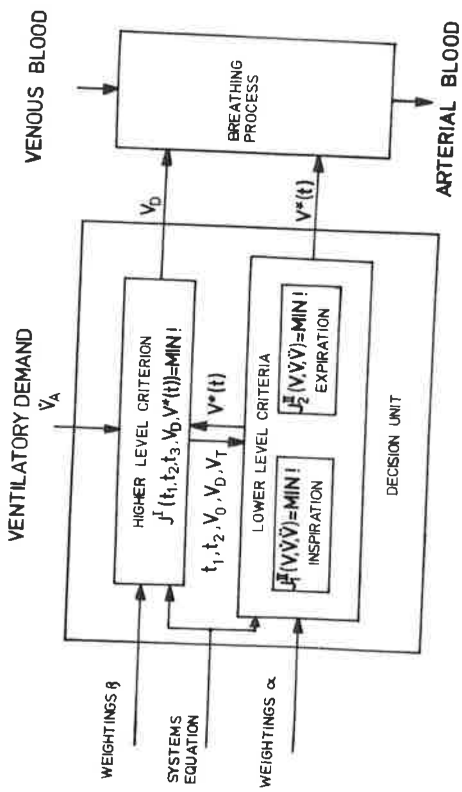

FIGURE 1. A schematic diagram of the two level model of control of respiratory mechanics.

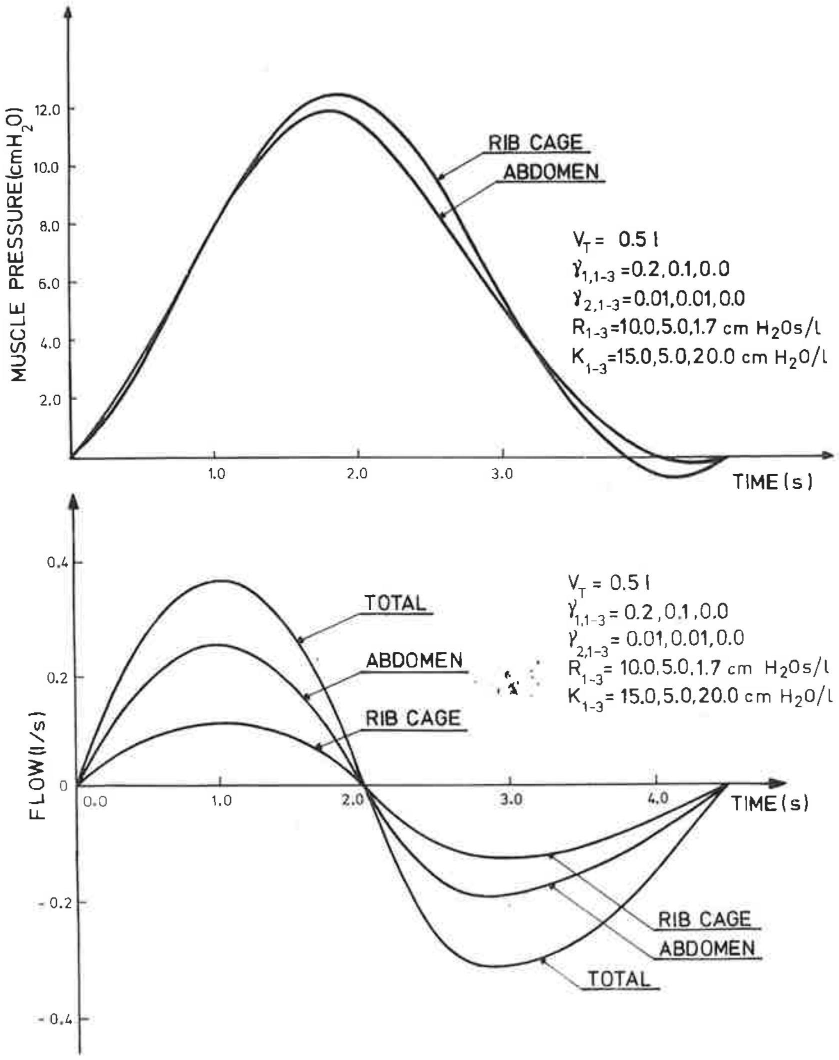

FIGURE 2. Flow and muscle pressure curves from the model with two degrees of freedom. Parameter values obtained from ref.(1).

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