ANALOG-DIGITAL DATA PROCESSING OF RESPIRATORY PARAMETERS

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INTRODUCTION

The data processing application to be described here might be regarded as a somewhat elementary one by those familiar with computer technology, i.e., most of those present. However, it is presented as an example of an application of data processing technology to a field in which most of the data gatherers are unfamiliar with the heights to which the computing art has been raised in its short lifetime.

It is very difficult for the physician scientist to learn how he can use modern computing technology, because he is, due to his medical education, (typically) lacking in mathematical knowledge or engineering knowledge. Another factor militating against his use of computer technology is the language barrier, more bluntly, the jargon. The physician has his own jargon, indeed, the process of medical education is primarily the learning of the medical language. Probably a smaller proportion of the effort in the computer field is directed toward semantics, yet, an individual armed only with a good understanding of the jargon could, with the assistance of an engineer, make significant advances in the application of data processing technology to medical research.

The following study is accordingly presented, not as an earth-shaking advance in computer technology, but rather as an effort to improve communication between the two disciplines of medicine and computer science.

First we present a discussion of the symbols, definitions, and lung model employed in the work.

- \( V(t) \) = flow rate of gas at mouth.
- \( V(t) \) also equals the rate of change of lung volume
- \( V(t) > 0 \) represents \textit{inhalation}.
- \( V(t) < 0 \) represents \textit{exhalation}.

\( F(t) \) represents the CO\(_2\) concentration in the gas at the mouth.

for \( V(t) > 0 \rightarrow F(t) \rightarrow F_i(t) \) i.e., \textit{inhaled concentration}.

\( V(t) < 0 \rightarrow F(t) \rightarrow F_e(t) \) i.e., \textit{exhaled concentration}.

\( F_a(t) \) represents the alveolar concentration of CO\(_2\).

We use a two compartment model of the lung.
Compartment 1, The Dead Space
This is purely a region of gas transport. There is no exchange. Its volume is $V_D$.

Compartment 2, The Alveolar Compartment.
This is a region only of gas exchange. There is no transport.

This compartment is assumed to have no concentration gradients. When gas is exhaled, there is some mixing of gas from the two regions, but the first portion of the exhalate represents dead space gas, the last portion represents alveolar gas. Thus we have the following curve of concentration versus time for carbon dioxide in the exhaled gas:

\[ MV = \int_0^T |V(t)| \, dt \]

The tidal volume $V_T$ is the volume of gas moved out of the lung on a given breath.
The alveolar volume $V_A$ is the volume of gas moved out of the alveolar compartment on a given breath. Obviously,

\[ V_A = V_T - V_D \]

Similarly, we can define the alveolar ventilation rate,

\[ \frac{T}{\sum V_A} \]

Since the CO2 concentration in the dead space is $F_I$ and its volume is $V_D$, the volume of CO2 exhaled from the dead space is $V_D \times F_I$.
And, volume of CO2 exhaled from the alveolar compartment is $V_A \times F_A$.

\[ V_ECO_2 = V_A \cdot F_A + V_D \cdot F_I \]

If $F_I = 0$, a common case, then

\[ V_A = \frac{V_ECO_2}{F_A} \]

If $F_I \neq 0$, then

\[ V_ECO_2 = V_A \cdot F_A + (V_T - V_A) \cdot F_I \]

But $V_T \cdot F_I = V_I CO_2$.

\[ V_A = \frac{V_ECO_2 - V_I CO_2}{F_A - F_I} \]

These two formulae for $V_A$ are known as the Boh formulae. We will now discuss the processing of the above data.

This paper discusses an improved version of two systems previously reported. An attempt has been made here to perform the various operations in the appropriate (analog or digital) section of the equipment instead of doing them all in the “analog” section.

EQUIPMENT

The system used consists of two transducers, special purpose analog computing equipment with digital read out and digital computing facilities.

The transducers are (a) a pneumotachograph...
Fleisch), strain gage (Statham PM 15) and amplifier (Statham CA 9-10), and (b) an infra-red carbon dioxide analyzer (Godart).

The computing equipment consists of 25 operational amplifiers, some with chopper stabilization (G.A. Philbrick Researches, K2PA and K2W) and a multiplier (GAP/R, K5M). Plug-in units for the amplifiers were fabricated by the author from modules (K3). Control circuitry was synthesized from digital modules by Tech-Serv (B.R.S.). Read-out equipment is by Hewlett Packard, and the digital computer is a Control Data Corp. 160 A.

Computation (Fig. 1)

From the pneumotachograph, strain gage and amplifier system a signal arises representing the instantaneous flow rate of the patient's exhalation or inhalation. A small sample (approximately two liters per minute) is taken from this stream and passed through the sampling head of the carbon dioxide analyzer from which is obtained a signal proportional to the carbon dioxide concentration in the gas stream (which is lagged approximately 300 ms). To synchronize the two signals, the "flow" voltage is delayed by an equal amount. This is performed using a (Fig. 2) modification of the Pade approximation devised by Dr. P. D. Hansen. The flow signal is rectified and integrated thereby giving the volume exhaled for that breath. The flow signal and the carbon dioxide signal are multiplied and integrated and this integrand for each breath represents the volume of carbon dioxide exhaled per breath. The peak exhaled or end-tidal carbon dioxide tension is obtained using a peak follower technique and the inhaled carbon dioxide from the inverted curve in the same way. These peak followers are reset after being read out on each breath. A constant voltage is integrated for the period of the breath to give a measure of the time taken for that breath. All five quantities are placed on memory circuits at the end of each breath. The integrators are reset and computation recommences.

Control Circuits (Fig. 3)

These consist of a series of digital modules by Tech Serv (B.R.S.). The input pulse to this system is obtained from a voltage crossing detector and relay on the analog unit. This then initiates a pulse train in the digital system which proceeds through a

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series of one-shots of variable delay time, each one of which is triggered by the trailing edge of the pulse from the preceding one shot. The time delays are adjusted to the appropriate values. The first one shot operates the relay which connects the integrators to the memory circuits. (A) The second one shot is a delay to allow for closure of this relay, a small dead time, and then operation of the second (shorting) (B) relay which is operated from a third one shot. Another one shot is used to provide a suitable delay between readout of the first channel \((F_{A} CO_2)\) and reset of this unit. (Relay C).

**Calibration**

Calibration of this equipment is rather complex due to the large number of functions performed and every attempt is made to cross-check during calibration. The carbon dioxide analyzer is calibrated with gases of known chemical composition. Its response to these is linear to within plus or minus 1 mm pCO\(_2\) at 760 mm barometric pressure. The pneumotachograph, strain gage and amplifier system is calibrated by passing oxygen through a flowmeter which delivers a known amount of gas for any particular position of the rotameter. Stability and linearity of this system are excellent, the only aspect requiring frequent adjustments being the zero level which is sensitive to positional changes of the transducer. The ability of the system to record accurately the volume passing through the pneumotachograph is tested by comparing the computer output with a volumeter and spirometer. Agreement here is excellent. ±3 percent. Stability and accuracy of the integrators is tested with a sine wave of known dimensions and again here reproducibility and accuracy are better than 2 percent.

Finally the system is tested by the simulation of a dead space. Calculation of dead space is the most revealing calibration statistic of the machine because the dead space represents the differences between two fairly large values, namely the tidal volume, and alveolar ventilation, which only differ by about 20 percent. Consequently errors in these quantities are reflected in an extreme fashion in the dead space calculations. Therefore a homogeneous carbon dioxide mixture is flushed through the pneumotachograph to simulate a zero dead space. Results of this typically indicate an average mean dead space determination of the order of 5 cc for a total volume of 500 cc passed through the pneumotachograph. Artificial dead spaces of 40 and 95 cc have been constructed. Average of the mean of 10 determinations for the 40 cc dead space was 41.6 cc in one instance and in another 45. Average of the mean of 10 determinations for the 95 cc dead space was 97 in one instance and 92 cc in another. These results lead us to have some confidence in the ability of the equipment. On the other hand, this confidence can only be maintained if calibration is conscientiously and frequently performed.

**Readout**

A multiplexing device connects the five memory circuits to a digital voltmeter sequentially. The digitized values are then printed or punched out. The first system, the printing system, is a slow speed unit consisting of a multiplexing device (Dymec C 2900 A) a digital voltmeter (Hewlett-Packard 405 CR) and printer (Hewlett-Packard 561 B). This system can read out five parameter in approximately 2.5 seconds. This speed is adequate as long as we do not have to have an observation on every breath. (If the readout sequence is not completed the integrators are merely reset and computation recommences. The integrators are not connected to the memory units in this situation). The other system is faster and consists of a similar stepping switch type of multiplexing device (Dymec C 2901 A) which connects the memory circuits through a 5-space digital voltmeter with a 10 ms sampling time, (Dymec 2401). The output of this is put on a punched paper tape by a teletype unit (BRPE-11). This latter system is of course much faster and will read out 5 parameters within approximately 7/10 of a second. This latter format is also much more convenient as it can be read directly into the digital computer (CDC 160 A). With the printing system data must be transferred onto cards, which is rather tedious.

**Programming**

Several programs are then available to us. The first program simply removes the scale factors used in the analog equipment and punches in the conventional units. Several types of manipulation are performed upon the scaled data of which a few examples are as follows.
We might desire a plot of alveolar ventilation rate against end-tidal carbon dioxide tension. This type of plot is useful in studies of the sensitivity of the respiratory center and the effect of drugs upon it. It is usually necessary to smooth this plot. The technique employed is to average the ventilatory rates over five breaths. Similarly the end-tidal carbon dioxide tension is averaged over five breaths (each tension is weighted by the time of the breath to obtain a meaningful average). This type of plot has been used by us extensively in assessment of the depressant effects of narcotics.

Another typical problem is determination of the relationship between tidal volume and alveolar ventilation. The latter is determined by the use of the Bohr formula above. This is a fairly elementary program and a plotting routine is incorporated here also to avoid the tediousness of plotting the large amount of data.

Using the formula for the case when the inhaled concentration of CO₂ is not zero, we must compute the net output of CO₂ for each breath, as the denominator for the previously derived formula:

\[ V_A = \frac{V_A^\text{CO}_2 - V_A^\text{CO}_2}{F_A^\text{CO}_2 - F_A^\text{CO}_2} \]

For this case the analog equipment is adjusted to compute the product of the concentration signal and all the flow signal, rather than the rectified signal.

By an obvious adaptation of the above program we can plot the net rate of CO₂ production against time for any time interval of interest. Such a plot is of interest because this parameter indicates the overall rate at which blood is returning from tissues in a normal metabolic state. A sharp drop in CO₂ output would indicate that the rate of return blood to the heart was reduced or that there had been a severe metabolic disturbance. Such information could be useful for an anesthesiologist during a difficult procedure.

Comparisons between the partial pressure of CO₂ in the arterial blood and in the lung gases are of interest inasmuch as any great differences reflect inefficiencies in the lung as an exchange device. True comparison is not usually made directly, but rather the "physiological alveolar ventilation" is determined. This somewhat empiric parameter is the result of replacing the \( F_A \) in the Bohr formula by \( F_a \), i.e., the fractional concentration corresponding to the partial pressure of CO₂ in arterial blood. Comparing the volume so obtained with the "alveolar ventilation volume" \( V_A \), defined in the introduction, allows us to express the inefficiency in terms of a volume of the lung (referred to as the alveolar dead space = \( V_A \) (phys.)) − \( V_A \)) which receives an adequate blood supply but an inadequate gas supply.

It would be easy to extend the above techniques to obtain many other parameters of respiration, of interest to the respiratory psychologist and the clinician, such as the timed vital capacity, one second expiration, etc.

**CONCLUSION**

Techniques are outlined for rapid data processing of respiratory parameters. It is suggested that these techniques are much more efficient than the classical techniques of chemical analysis, etc. Much more data is obtained and the maximum number of parameters can be calculated from an individual experiment. It is suggested that we can have a fruitful union of medicine and data-processing technology.