A SHORT RANGE UNDERWATER BIOTELEMETRY SYSTEM

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Summary The requirement for monitoring the physiological functions of the test subject in weightlessness simulation activities coupled with the advantages of using telemetry for such monitoring led to the development of a biotelemetry system. One valuable technique for simulation of weightlessness uses the neutral buoyancy obtained by having the subject under water, which leads to a requirement that the telemetry system work in this medium. Previous underwater telemetry systems have usually used ultrasonic carriers. The system described in this paper is unique in providing a multiple channel underwater telemetry system using an electromagnetic carrier. The development of transducers used with this system to provide information on the work load Imposed by various simulation tasks is also described.

Introduction In the future, man can be expected to be used extensively to accomplish tasks in space. The zero G environment materially affects the ability of the astronaut to accomplish the task and the metabolic cost of the performance of such work in space. Several simulation techniques have been developed to aid in ground-based experimental studies of man's ability to accomplish work in the weightless condition when operating in the constraint of a pressure suit. The subject in these tests should be monitored for his physiological status, and an ability to measure changes in metabolic work load imposed upon the subject produces more information concerning the effect of the task on the mm. One of the techniques used for simulation of weightlessness requires that the subject be under water. In this condition he is neutrally buoyant and therefore experiences many of the effects of weightlessness as encountered in space. The system used to monitor the man should therefore be designed so that it may be used under water.

Telemetry offers obvious advantages in biological monitoring systems, the most important one being that it removes the necessity for encumbering leads attached to the subject under test. This of course reduces the tendency of measurement itself to have an effect on the test subject. In cases where an umbilical is to be used by the test subject anyway, hard wiring may be used. Even in this case, the use of a mixed composite signal for transmission of all the physiological information,, a concept taken from telemetry systems, is valuable in that it reduces the number of leads required to a single pair.

Several decisions were made early in the development of the system which shaped the final configuration. First, the goals chosen for this system were monitoring of the subject and determination of changes in the imposed work load., as opposed to an attempt to measure total metabolic rate. Second, four physiological parameters were selected for measurement which would reflect changes in work load: (1) heart rate, (2) body temperature, (3) breathing rate, and (4) partial pressure of carbon dioxide in the expired air. The amount of oxygen consumed is a more accurate index of metabolic rate end a more standard measurement of metabolic rate than pCO_2 in expired air, but no technique for measuring O_2 consumption is available which lends itself to a small biotelemetry unit. The various available O_2 sensors are not useable for reasons of power consumption, chemical interference, response time, or susceptibility to shock or vibration damage. Finally, two engineering design constraints were imposed which required that the system respond quickly enough to reflect the changes caused by individual parts of the total simulation exercise, and that the size and weight of the entire unit be kept as low as possible while providing a running time without attention of several hours.

Telemetry Since telemetry would be a desirable feature of the monitoring sysre-mand compatibility with the underwater environment was a necessary requirement,, a system which would provide both of these was the goal. An ultrasonic carrier appeared to be the only proven method of providing both features, but has several inherent disadvantages. First, and most important in this unit to be worn by a test subject, the power and weight penalties imposed by an ultrasonic system are high. Second, telemetry using an ultrasonic carrier is not nearly as well known as standard electromagnetic telemetry, adding considerably to development time and cost. Finally, additional problems were anticipated in this application because the system would be used in relatively small pools., where considerable reflection from the concrete walls and bottom could be expected. Interference from the noise generated by the pumps and other equipment in the pool would undoubtedly be present.

Since a system using an electromagnetic carrier presents significant advantages in cost, power, weight, availability of demodulating equipment, and ease of construction, a program to determine the feasibility of using such a transmitter was undertaken.

It would be expected that the properties of water, being quite different from those of air, would materially affect the ability to transmit effectively an electromagnetic wave in this medium. The relative permeability of water, μ_γ, is approximately 1.00 (approximation good to better than 1%), but the relative permittivity, ϵ_γ, is significantly different from the value for air. The relative permittivity depends on the impurities and on the temperature, but will be approximately 79 in a well filtered pool at 23°C. The equation for the wavelength of a propagated wave is given in equation (1):

$$\lambda = 1/f\sqrt{\mu\varepsilon} \qquad (1)$$

Where: λ = wavelength in meters.
f = frequency in hz.
μ = permeability of the medium in henry/meter.
ε = permittivity of the medium in farad/meter.

Since ϵ may be expressed as the product of ϵ_o, the permittivity of free space and ϵ_γ, the relative permittivity of the medium, it follows easily that the wavelength is inversely proportional to the square root of the relative permittivity. The approximate relative permittivity of water is 79, so the wavelength of the propagated wave is only about one ninth its value in air for the same frequency. This makes the design of an efficient antenna possible at a reasonable frequency for use in a small body-worn telemetry package.

Other properties of water make it a less desirable medium for electromagnetic radiation. The relatively high conductivity of water makes it a very "lossy" medium. What the conductivity will be and therefore how much attenuation to expect depends upon,the water encountered. The value for sea water is approximately 40 x 10^3 micromho/cm., but for distilled water is only 2 micromho/cm. Lonsdale has reported good results using a 100 Mhz. telemetry system for tagging fish in water with a d.c. conductivity of 231 to 366 micromhos/cm., and total dissolved solids count of 236 to 310 ppm (1). Water from a typical pool used for underwater simulation studies was checked and found to have a conductivity of 2.2 x 10^3 micromho/cm. at 60 hz., becoming 2.3 x 10^3 at 1 khz. The total dissolved solids count was 190 ppm.

The relatively high conductivity of the water causes rapid attenuation of the radiated signal. However the reduced wavelength makes efficient coupling possible with a small antenna, and this helps to some extent to offset the effect of the high conductivity. In order to make a real test of the feasibility of using an electromagnetic carrier a simple high frequency oscillator was constructed and tested in a small pool. From the experimental transmitter it was clear that for the short ranges necessary for this application an electromagnetic system could be used.

The use of an electromagnetic carrier allowed the final design to follow that of a typical telemetry system, as illustrated by the block diagram of the system shown in Fig. 1. The voltage controlled subcarrier oscillators are commercially available units operating on standard IRIG channels 5, 7, and 11 for body temperature, percent CO_2, and electrocardiogram respectively. Each VCO accepts a 0 to 5 volt input to produce ± 7 1/2% frequency deviation at the output. The composite signal at the output of the simple resistive mixer is used to frequency modulate the transmitter.

The simple transmitter used in the system consists of only two stages. A shielded and temperature compensated one transistor L-C oscillator is frequency modulated using a varicap diode. This oscillator drives a one transistor tuned amplifier. Both the oscillator and the amplifier are provided with variable capacitors which allow the output frequency to be varied from 105 to 250 Mhz.

Two antenna designs have been used with essentially equal results. The size of the entire package is limited to 8"x 8"x 2 1/2" by the design of current concepts for “chest packs” and “back packs” being considered for EVA. The high permittivity of water results in a wavelength at the lower frequencies used of 0.28 meters or 11 inches, thus permitting the design of antennas on the order of one wavelength even in the limited space available. One antenna used straight wires running at right angles to each other along adjacent sides of the box and fed in the middle. The other used a simple one turn loop with a 7" diameter. In both cases the wire is brought through the removable lid of the waterproof housing and then covered with a thin layer of fiberglass. This procedure preserves the waterproofing but puts the antenna in ‘very close proximity to the water.

The transmitted signal is detected with a communications receiver which provides a tuning range from 85 to 250 Mhz. The composite signal from the discriminator is used for the input to a standard set of filters and discriminators providing ± 2 1/2 volt output for ± 7 1/2% deviation of the subcarrier frequency. The analog output of the discriminators is displayed on an oscilloscope for subject status monitoring and simultaneously recorded on chart paper for later analysis of the data.

Physiological Measurements Transducers and signal conditioners are provided to make four physiological measurements: heart rate, body temperature, breathing rate, and partial pressure of carbon dioxide in the expired air. The heart rate information is obtained by transmitting the subject’s electrocardiogram. The ECG signal is picked up using waterproof electrodes attached to the subject’s chest. Both sternal and trans-thoracic placements of the electrodes have been used with satisfactory results. The signal is amplified and used to modulate one of the subcarrier oscillators. The amplifier used for this channel has a bandpass limited to 1 to 40 hz. This bandpass limiting reduces the fidelity of the received ECG, but in this application the ECG is monitored only for the purpose of providing cardiac rate, and the limited bandpass materially reduces problems of baseline shift, muscle noise, etc. The use of silver-silver chloride electrodes also minimizes the occurance of artifacts in the signal.

Body temperature is measured using a thermistor. The thermistor may be placed in the mouth of the subject or attached to his skin. The surface measurement is generally preferred because it is more likely to change with work load as the body attempts to transfer excess heat produced by the higher metabolic rate. The thermistor is used in a conventional d.c. bridge circuit, with the amplified output of the bridge used to modulate a subcarrier oscillator. The amplifier gain is set to allow a maximum variation of temperature of ± 2 1/2°. Greater temperature variations will produce an output volta7ge outside the acceptable input for the subcarrier oscillator.

A plot of discriminator output voltage vs. thermistor temperature is illustrated in Fig. 2. The temperature response is clearly linear over this limited range of temperatures. The overall system sensitivity and noise allows reading changes in temperature of 0.1°C, and since only changes in work load are being studied, this is the characteristic of major Importance. The accuracy of the system over a period of three days is ± 0.3°C.

The third channel is used for both breathing rate and partial pressure of carbon dioxide in the expired air. The basic concept behind the sensor used for this measurement is the change in thermal properties of a mixture of O_2 and N_2 when CO_2 is added to the gas. The sensor used is a modified thermal conductivity cell of the type used in gas cbromatographs. The cell contains four filaments arranged in a bridge configuration, as shown in the schematic diagram of Fig. 3.

To obtain the fastest possible response time and the smallest size, a thermal conductivity (TC) cell with an internal volume of only 90 microliters per filament is used. The modification consists of further reducing the internal volume at the expense of sensitivity by removing two of the filaments, one on the sample side and one on the reference side, and plugging the entrances to the two unused chambers. This process reduces the internal volume by one half and reduces the sensitivity by one half, but improves the response time. The two removed filaments are replaced in the electrical circuit by a pair of precision resistors.

The bridge filaments are heated by approximately 0.1 watts of power supplied by a small battery. The resistance of each element is a function of its temperature, and the overall temperature of the TC cell will vary uniformly with the ambient temperature due to the mass of the bridge enclosure. The cooling of the filament when there is gas flow will depend on the heat transfer coefficient of the gas, which in turn depends on several properties of the gas as indicated by equation 2:

$$h = C\, k^{0.7}\, c_p^{0.3}\, \mu^{-.25}\, \rho^{0.56} \qquad (2)$$

where: h = heat transfer coefficient
C = constant depending on geometry
k = thermal conductivity of the gas
c_p = specific heat of the gas
μ = absolute viscosity of the gas
ρ = density of the gas

The rate at which a filament dissipates heat is primarily dependent on the thermal conductivity of the surrounding gas, as indicated by the larger exponent.

When air containing no CO_2 is flowing through the system, the cooling of both filaments is equal and the bridge is in balance. If CO_2 is contained in the sampled gas, however, the cooling of the element which is exposed to the mixture is reduced since CO_2 has a smaller thermal conductivity than does air. The CO_2 is removed by the chemical bed containing MM before the gas passes to the other filament, so this filament is still exposed to air. The increased temperature of the filament exposed to the mixture of air and CO_2 increases its resistance, and the bridge will be unbalanced. The chemical bed upstream from the TC bridge contains a desiccant to remove water from the sampled gas before it reached the filaments. The desiccant at the downstream end of the MOR chemical bed removes the water formed as a product of the reaction of CO_2 with LiOH from the air stream before it reaches the filament on the reference side of the bridge. Various other chemicals were considered for water and CO_2 absorbtion but were discarded because of uncontrollable by-products or partitioning action of the sample. The output voltage from the bridge is amplified and used to modulate the subcarrier oscillator.

The system is designed to be used with a subject in a pressure suit, so this pressure is used to provide the flow through the pCO_2 sensor, as illustrated in Fig. 3. Moderate lengths of capillary tubing and a needle valve to control flow rate stabilize the pressure and flow in the detector compared to the pressure fluctuation produced by movement of the subject in the suit. By using a vacuum source to obtain 2 to 31/2 psi pressure drop through the system., the unit may be used with any subject, and this procedure is also used when calibrating the system.

In order to avoid the problems associated with baseline drift which may occur with changes in battery voltage, pressure, or temperature, a method of sampling a standard gas at frequent intervals is desirable. The subject’s breathing air is a convenient source of a standard gas, and by taking the sample directly from the line to his mouthpiece alternate samples of the standard gas and expired air are obtained without the addition of complicated waiving systems. This technique also provides breathing rate without additional sensors or signal conditioning equipment.

Very rapid response time is necessary in order to measure the percent CO_2 in the expired air with each exhalation before the sample of fresh air from the following inhalation reaches the sensor. The response time is minimized by using the small volume TC cell and modifying it as explained previously. In addition the chemical bed containing the desiccant must be carefully designed to minimize mixing of the gas stress and very small diameter tubing must be used throughout the system. A trade-off must be made between flaw rate and response time. Higher flow rate will produce a faster response time, but also causes quicker saturation of the chemical beds and makes the system more sensitive to suit pressure changes. A flow rate of 75 standard cubic centimeters per minute permits a running time of more than four hours without changing the chemical bed and a system response time (to 90% of fall scale) of 550 milliseconds. The chemical bed is mounted so that it can be replaced quickly between simulation tasks.

The calibration of the carbon dioxide sensing system is accomplished by sampling prepared gases containing a known mixture of air and carbon dioxide. Fig 4 is the calibration curve and illustrates the linearity of the system over the limited range of pCO_2 values encountered in expired air.

Results The output of all three channels is illustrated by the recordings in Fig. 5. A portion showing the initial application of the thermistor after removal from a 35°C temperature bath was chosen to illustrate its response time and sensitivity. The recording illustrates the response time of the CO_2 measuring system A close examination shows that the response is adequate to reach its final value for each exhalation and return to the baseline for each inhalation.

The biomonitor has been used to measure changes in cardiac rate, body temperature, respiratory rate, and CO_2 production with various degrees of activity in both the underwater subject and the dry subject in a space suit configuration. It has also been used with a vacuum source when the subject is not in a pressure suit, and in fact has a greater accuracy in this case where ambient pressures are constant.

The system has proven to be a sensitive indicator of changes in work load., although care must be taken in interpreting the results. For example, hyperventilation can reduce the percent CO_2 in the expired air even though the work load increases. This will be accompanied by an increase in respiratory rate and a change in the respiratory cycle, as well as an increased heart rate. A rise in body temperature can also result in a lower CO_2 reading, since in this case the CO_2 tension in the blood may increase. However, in general an increased work load will result in higher pCO_2 readings and an increase in breathing rate. The sensitivity of these measurements is illustrated by the recordings of Fig. 6. The upper recording was made at the beginning of the test with the subject well rested. During the test the subject remained seated with the only change being the requirement that he breathe using the mouthpiece, air line, and valves incorporated in the pressure suit helmet. The second recording was made approximately three minutes later. The carbon dioxide content of the expired air has increased by approximately 12%, and the breathing rate has also increased, indicating the larger volume of carbon dioxide being produced by the body. Of course more strenous tasks result in more dramatic changes in the measured physiological parameters.

Although only short transmitting distances are involved, and the water is not nearly as conductive as sea water, telemetry certainly can be a useful technique for underwater data transmission under certain circumstances. One of the significant advantages of telemetry is the ease with which additional channels may be included. A fourth subcarrier oscillator is already installed in the biomonitoring unit which may be used to measure respiratory minute volume or mass flow when the details of adding this measurement are worked out. This would allow the calculation of total volume of carbon dioxide being removed by the lungs. Even when an umbilical is used., the transmission of a single composite signal is certainly preferable to having several sets of wires attached to the subject.

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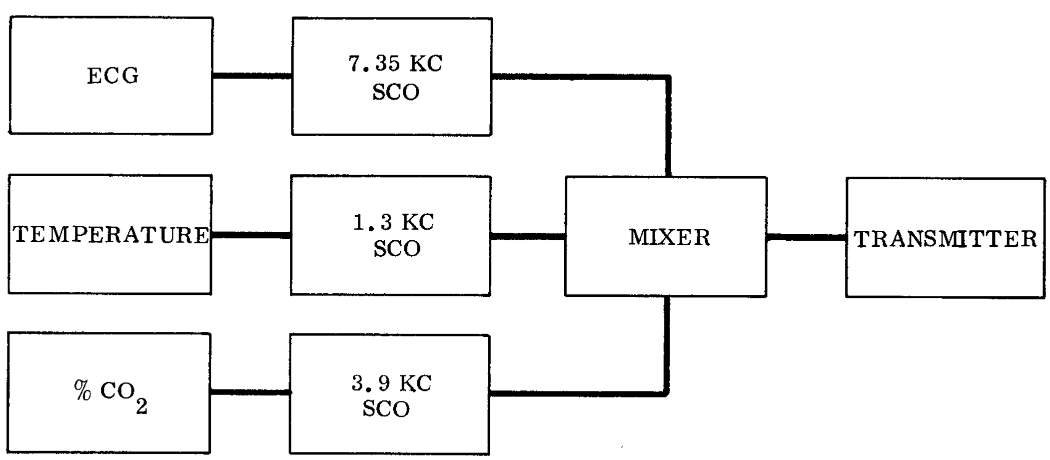

Fig. 1 - Block Diagram of System.

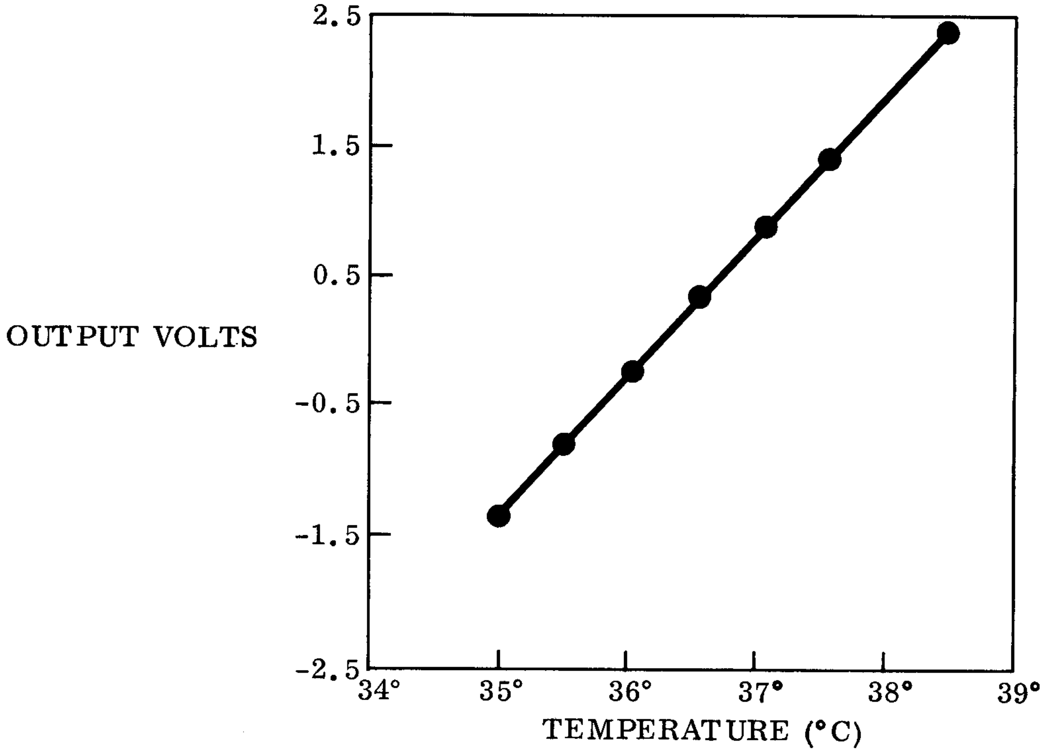

Fig. 2 - Temperature Calibration Curve.

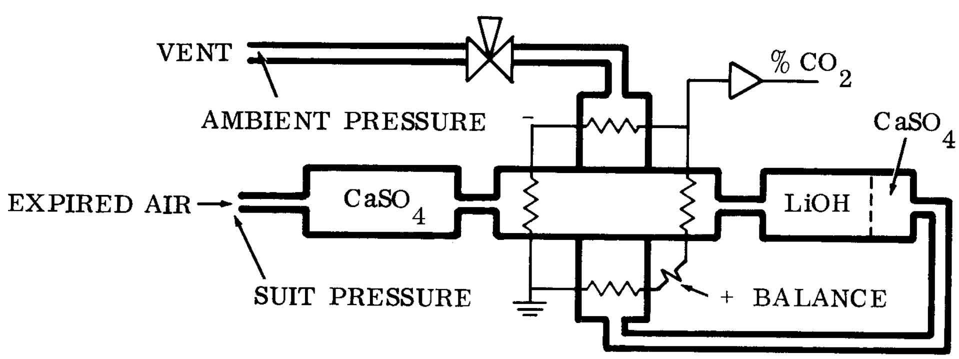

Fig. 3 - Schematic Diagram of CO_2 Sensor.

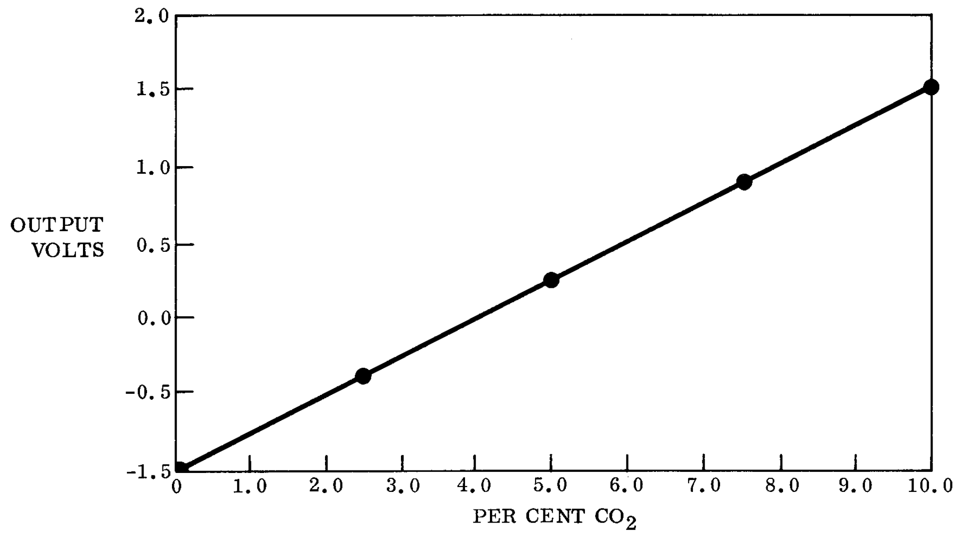

Fig. 4 - CO_2 Calibration Curve.

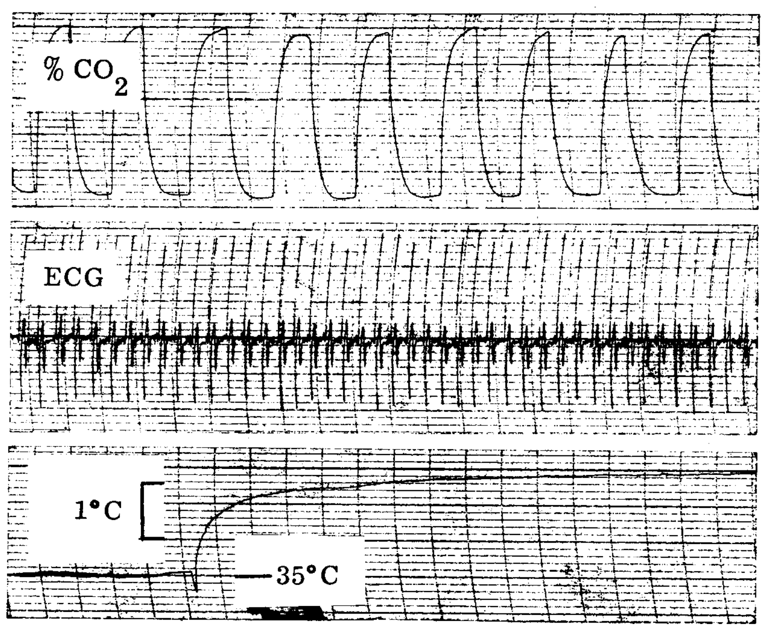

Fig. 5 - Output Recording.

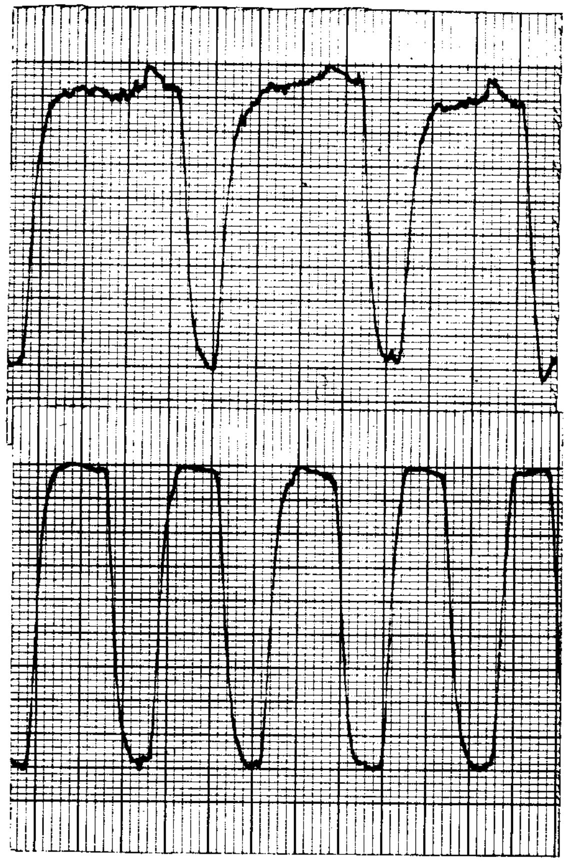

Fig. 6 - Comparison of pCO_2 & Breathing Rate.