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CR 114905

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INTRODUCTION

Under Contract NAS9-8185, Quantum Dynamics, Inc. was directed by the National Aeronautics & Space Administration's Manned Spacecraft Center (Dr. John Rummel, Technical Coordinator), to carry out the assigned tasks of supplying advanced Bi-Directional Respiratory Flowmeters and Electronic Instrumentation Systems for the measurement, analysis and computation of metabolic quantities. These systems were delivered in August, 1970. Their detail designs are described in their rather voluminous instruction manuals. The following report therefore, has as its main purpose: A) The review and analysis of significant technological advances achieved at Quantum Dynamics in the area of respiratory and metabolic instrumentation; B) A discussion of the administrative, technical, and other problems we have encountered during the performance of the contract; C) Proposed future effort.

PART I.

TECHNICAL SECTION

1. Review of Areas of Achievement.

1.1. Advanced Bi-directional respiratory flowmetering systems of special turbine type, having unprecedented high resolution and precision, are now available for NASA-MSC programs. Such new systems have the following desirable characteristics:

a) respiratory measurement resolution of 1 milliliter (10^{-3} liter) is now achievable;

b) can cover the entire respiratory flow range of from 5 to 500 liter/min. (0.2 to 17.7 ACFM) with one single and practical instrument;

c) extraordinarily high sensitivity;

d) low pressure drop, and therefore presents no discomfort to the user.

e) calibrations in both (Insp. and Exp.) flow directions can be made virtually identical. Exhibit 1 shows the calibration of such an instrument at 1 atm. pressure. There is no other flowmeter which can even remotely approach such performance superiority.

1.2. For the first time, a series of respiratory flowmeters have achieved dependable full range capability in sub-atmospheric (e.g. 1/3 atm. or 5 PSIA) measurements with equal precision as in higher atmospheres. Exhibit 2 shows the 5 PSIA calibration of a prototype

flowmeter, ranging from 3 to 500 liter/min. with high linearity, sensitivity and accuracy. Moreover, Quantum Dynamics respiratory flowmeters have been thoroughly tested in our sub-atmospheric artificial lung testing system and found to be able to perform at 3.7 PSIA, or even at 1 PSIA. Thus, such flowmeters meet other needs of NASA's programs, such as in metering back-pack flow and in other life-supporting gas metering functions.

In contrast to our demonstrated achievement in sub-atmospheric flow measurement, numerous other makes of gas flow metering instruments have also been tested by us (in our subatmospheric calibrator, or by NASA). These were found to be totally inadequate and, in fact, many of them are considered dangerous under subatmospheric oxygen flow condition.

1.3. Since 1968, a bi-directional respiratory flowmeter has become available in its essential form. Using our Directional Discriminating Electronic Indicating circuits, the change from inspiratory to expiratory flow and vice versa, can be indicated and separated automatically, dependably and safely. Respiratory data can be sorted accordingly for subsequent processing. According to Mr. Paul Schlottman, our former technical coordinator, we have accomplished a feat in which numerous others have failed. Virtually all previous attempts fell short of success: due to unsafe contact switching which did not work; or suffering from chronic mal-function due to moisture, speech sound (acoustics) and a host of respiratory conditions. Our present bi-directional discriminator is totally immune to such effects and, to the best

application of this concept.

1.6. Accurate and dynamic measurements of the absolute pressure and, particularly, absolute temperature, as well as the breath-by-breath dynamic temperature of the respiratory flow have been accomplished. Direct on-line computation of the real-time mass or standard flowrate of respiration can now be implemented in one stage, using the pulse-rate input of flowmeter and the analog inputs of absolute pressure and temperature.

1.7. Method and circuits for the non-simultaneous use of inspiratory and expiratory flow for the determination and storage of uptake and release data were developed, using the techniques described in 1.1 to 1.6. At this stage, the required circuits for the direct digital computation of "uptake and/or release" as divided by T_M have been developed in the "bread-board" form. These will be applied directly to new MIRACLE Systems.

1.8. Effective basic method has been developed for the joint and simultaneous electronic computation of the Bi-directional respiratory flowmeter output and the mass spectrometer outputs. This necessitates the "synchronized delay" of the flowrate signal, and its direction-change signal, by a large amount of (delay) time of up to 600 milliseconds, which must be achieved in order to re-match the belated arrival of the mass spectrometer signals.

There are, however, room for improvements: with the present Data Delay Controller, the amount of delay is fixed by discrete in the amount of delay time. It is possible that the "phase difference" between the flow signals and the mass spectrometer output [illegible]

of our knowledge, there is no device which can even remotely approach its operational reliability.

1.4. Since 1969, the measurement of dynamic respiratory flow phenomena has been proven feasible. Our method is based on the joint-use of: a) turbine respiratory flowmeter (See 1.1; b) the Dynamic Flow Electronic Auto-Corrector; and c) Electronic FPAC (Frequency-to-Period Analog Computer) type Transient Flow Indicator in an electronic computer arrangement. The design of the Auto Corrector is based on the continuous on-line solution of nonlinear higher order differential equation on the real-time basis. The present units are based on the "hybrid" (digital-to-analog-to-digital) type computer approach. It has already demonstrated a ten-fold improvement on the dynamic response (See subsequent section) of the flowmeter itself by reducing the instrument's time constants to satisfactory low figures. As a result, this method provides more truthful waveform response to time-varying respiratory phenomena.
With some additional work based on the proposed improvement to be detailed later, the transient response can be further improved; the noise figure can be cut-down drastically and, the dynamic error can be minimized. (See Exhibits 3A,B,C.)

1.5. A novel concept of metabolic computational analysis was advanced, and its required circuits were developed; namely, in the "multi-channel physiological clock of respiration concept" based on the T_M criteria. This concept made possible economical and effective electronic real-time computation, measure and analysis of metabolic data. NASA-MSC is currently considering extent

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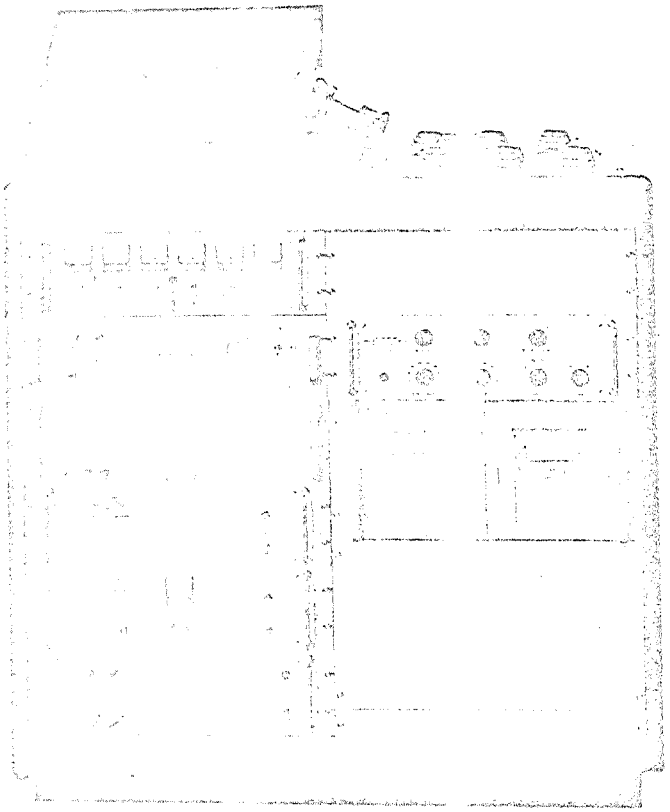
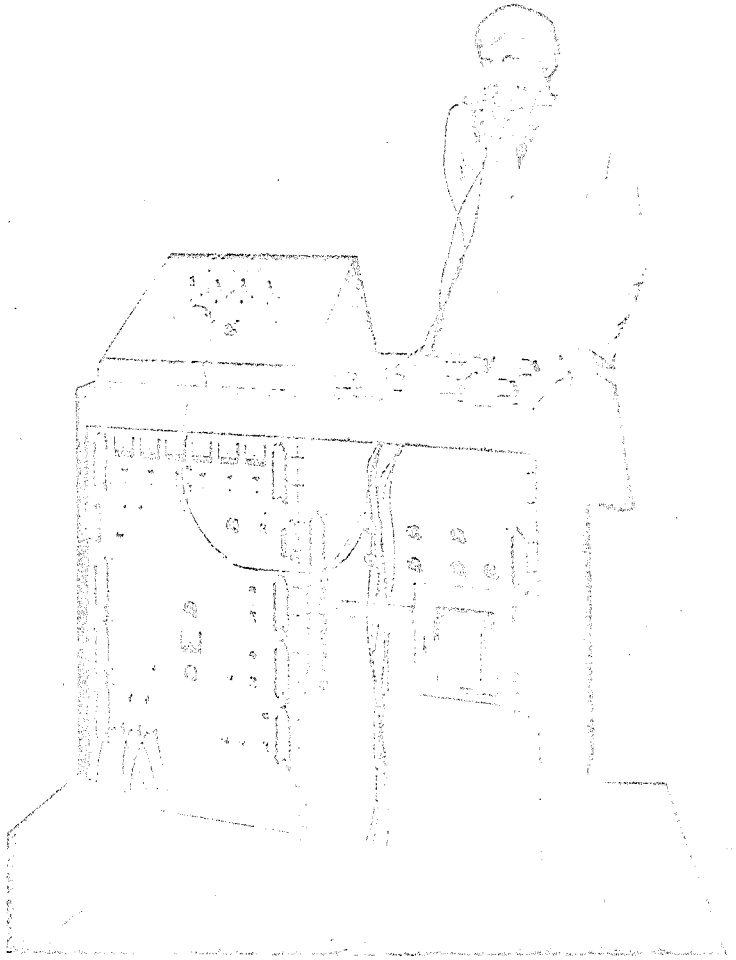
O_2) may not be a constant quantity - due to the asymmetrical nature of the mass spectrometer's input delay. In the next phase of work, an "automatically indicated and adjusted" delay controller can be developed.

1.9. A special data system has been developed specifically for metabolic data based on the T_M control criterion which can provide storage, automatic-recycling without loss of data and print-out command, etc. It operates on the T_M basis of 1 min., 1/2 min., and 1/4 min. The present system is considerably handicapped by the 4-digit counter it employs; for it was designed with the uni-directional system in mind, and had not anticipated the high resolution of the present respiratory flowmeter.

1.10. A sound technical approach has also been definitized with which the existing "dynamic accuracy" of the mass spectrometer can be improved. Several basic advancements can be proposed. Because we are deprived of the use of any mass spectrometer, due to the inability of Teledyne Earth Science Division to deliver one as expected, we were forced to improvise and are at the point of developing an effective one ourselves.

In reporting the aforementioned accomplishments, we cannot say that we have not erred - albeit in minor areas, and often in a "comical" manner. Now that virtually 98.5% of the technological ingredients are at our disposal, it would be constructive for us to make a candid appraisal. It can serve a useful purpose for the cognizant authorities at MSC to see the difficulties and contributions against some of our [illegible] due to the circumstances under which we performed the [illegible]





MODEL II
METABOLIC INSTANTANEOUS RESPIRATORY
ANALYZING- COMPUTING-LOGGING
ELECTRONIC SYSTEM (MIRACLE II)

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3/4" BI-DIRECTION RESPIRATORY FLOWMETER

DYNAMIC ABS. PRESSURE
ABS. TEMP. PLUG-IN

DATA FILE NO. C-102

1 ATM. GAS FLOW CALIBRATION

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Exhibit 1

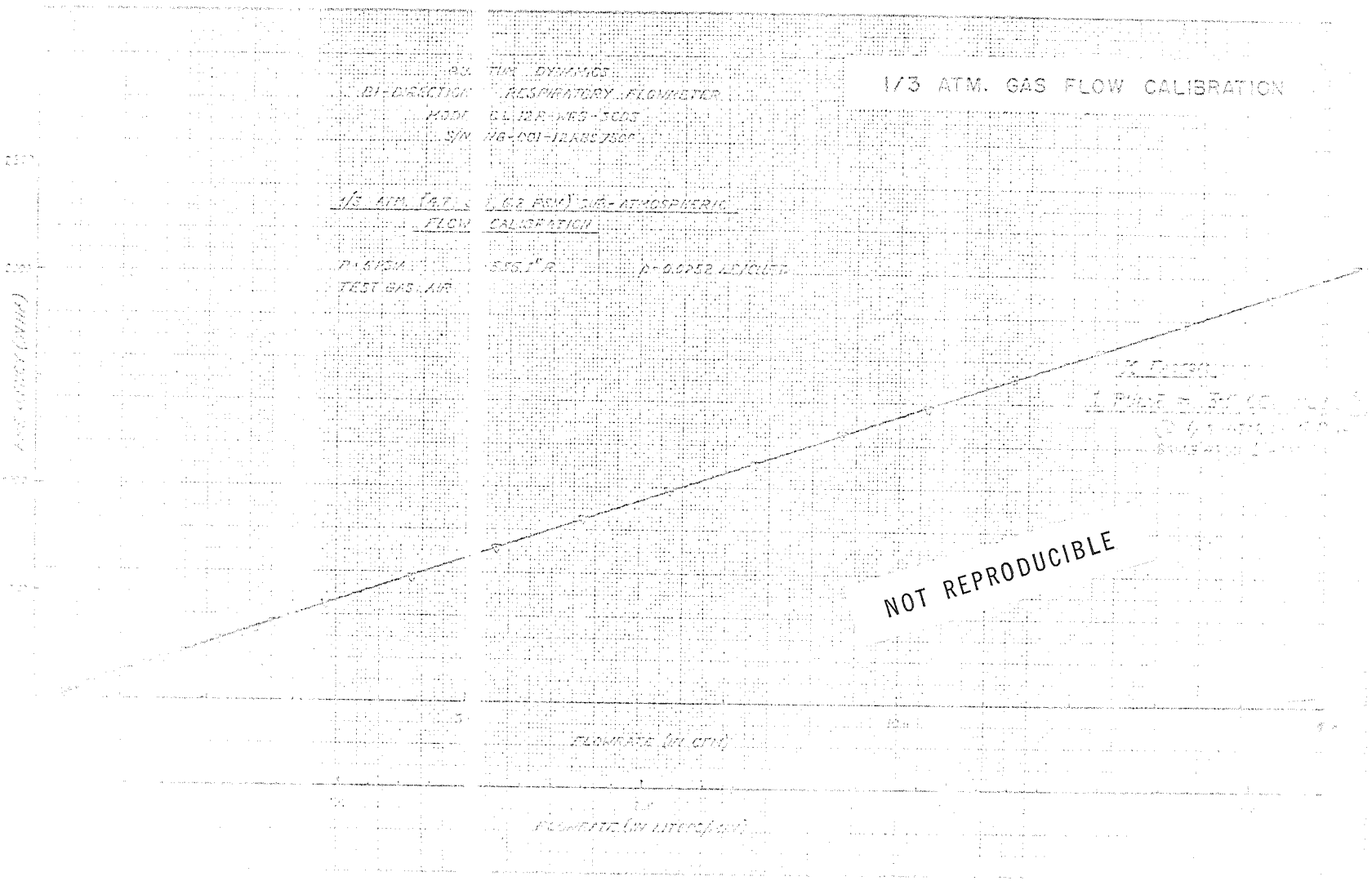

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Exhibit 3A

QUANTUM DYNAMICS TURBINE FLOWMETER
CL128-NRC-3015
NG-001-12RB07500
AUG 10 70 0[illegible]
INLET DIA= 1 THROAT DIA= 1 INCH
TEST GAS: AIR
MTR READ: MPT2(CYCLES/SEC)
K FACTOR: CYCLES/CU-FT
R REY NB: THROAT REYNOLDS NUMBER

L	MTR READ	G CU-FT/MIN	K FACTOR	R REY NB
1	0	.291502	0	129
2	18.3	.355934	2072.41	183
3	37.6	.460207	4200.12	237
4	72.7	.665870	6601.2	345
5	106	.870209	7209.35	453
6	134.8	1.08780	7504.61	561
7	158.7	1.29787	7711.2	669
8	184.9	1.5065	7769.35	778
9	206.4	1.71483	7891.91	886
10	234.3	1.92787	7919.37	994
11	251.8	2.14021	7996.75	1103
12	269.6	2.35833	7996.86	1215
13	288	2.56454	7987.84	1325
14	297.1	2.77689	7921.6	1435
15	314.8	2.9895	7965.07	1545
16	316.2	3.2152.5	7905.39	1216
17	322.9	3.71987	7919.24	898
18	125.9	3.28587	7418.42	560
19	37.2	4.48683	6945.61	237
20	3.6.2	2.40782	7990.62	1206
21	446.9	3.28468	7954.72	1751
22	537.5	4.24542	7949.99	2072
23	117.9	5.35442	7962.38	2373
24	847	6.24727	8004.46	2917
25	953.6	7.2475	8039.12	3206
26	1113.2	8.24594	8034.85	4221
27	1247.9	9.32523	8020.58	4641
28	1377.9	10.3323	8067.04	5340
29	1500.6	11.3036	8060.	5973
30	1631.5	12.3177	8040.45	6395
31	1752.3	13.3175	8036.42	4219
32	1873.6	14.3113	7987.41	7445
33	1991.2	15.2993	8004.32	7977
34	1445.1	16.2812	8034.7	6419
35	1[illegible]	17.2547	8154.11	4140
36	848	6.32545	8043.7	3002
37	445.1	3.37420	7932.5	1760
38	317.9	2.3097	7945.67	1291

[illegible]

QUANTUM DYNAMICS TURBINE FLOWMETER
CL128-NRC-3015
NG-001-12RB07500
AUG 10 70 0[illegible]
INLET DIA= 1 THROAT DIA= 1 INCH
TEST GAS: AIR
MTR READ: MPT2(CYCLES/SEC)
K FACTOR: CYCLES/CU-FT
R REY NB: THROAT REYNOLDS NUMBER

EXCLUDED FROM AVERAGES: 14 POINTS. SCALE: 7.99

L= 1 R= 129	% DEV=-100	
L= 2 R= 183	% DEV=-73.988	
L= 3 R= 237	% DEV=-38.505	
L= 4 R= 345	% DEV=-17.184	
L= 5 R= 453	% DEV=-9.31	
L= 6 R= 561	% DEV=-5.242	
L= 7 R= 669	% DEV=-3.255	
L= 8 R= 778	% DEV=-2.617	
L= 9 R= 886	% DEV=-1.619	
L= 17 R= 898	% DEV=-1.914	
L= 18 R= 560	% DEV=-6.052	
L= 19 R= 237	% DEV=-39.209	
L= 24 R= 2007	% DEV= .446	
L= 26 R= 3001	% DEV= .913	

X POINTS EXCLUDED FROM AVERAGES. K(AVE)= 7970.99 %DEV FROM K(AVE)
L(0) R/1000 MTR READ L(0)-20 -10 0 +10 +20

X 1	.12	0	
X 2	.18	12.3	
X 3	.23	37.6	19
X 4	.34	73.7	
X 5	.45	106	
X 6	.56	134.8	18
X 7	.66	146.7	
X 8	.77	194.9	
X 9	.88	224.4	17
10	.99	254.3	
11	1.1	281.8	
12	1.21	309.6	16
20	1.24	316.2	35
13	1.32	339	
14	1.43	367.1	
15	1.54	396.9	
21	1.75	444.9	37
22	2.26	577.5	
23	2.77	710.9	
X 24	3.38	847	36
25	3.8	953.6	
26	4.32	1117.2	
27	4.84	1247.9	35
28	5.36	1375.9	
29	5.87	1500.9	
30	6.39	1631.5	34
31	6.91	1752.3	
32	7.43	1873.6	
33	7.97	1991.2	

LS 996 <R= 3020 X= 7920.07 K/K(AVE)= .993625
STD DEV= 27.2724 % STD DEV= .344 BASED ON 12 PTS
MAX DEV= 45.8158 % MAX DEV= .578 AT L= 19 CHAN= 1

MD 3320 <R= 5650 X= 8002.14 K/K(AVE)= 1.00338
STD DEV= 37.2579 % STD DEV= .464 BASED ON 5 PTS
MAX DEV= 31.9681 % MAX DEV= .398 AT L= 34 CHAN= [illegible]

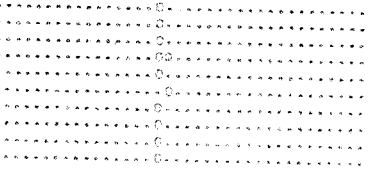
HI 5450 <R= 7977 X= 8009.99 K/K(AVE)= 1.00541
STD DEV= 00.7811 % STD DEV= .096 BASED ON 4 PTS
MAX DEV=-00.2389 % MAX DEV=-.030 AT L= 30 CHAN= 1

R(AVE)= 3004 K(AVE)= 7970.99

R(LO)= 2159.63 R(MD)= 4496.62 R(HI)= 6813.61

X = A + B*R + C*R**2
A= 7763.84 B= .112077 C=-1.0135E-5

K = X + Y*(MTR READ) + Z*(MTR READ)*(MTR READ)
X= 7817.04 Y= .22M407 Z=-1.0454E-4
BASED ON ABOVE TEST DATA (R.7.SAC)
AND ASSUMING K CORRELATES WITH MTR READ



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THIS CALIBRATION IS TRACEABLE TO NBS.
THE FLOW MEASUREMENT ACHIEVES IS [illegible]
GOLF ENGINEERING DIRECTORATE STATION.
PO BOX 3[illegible] GOLF [illegible]

Exhibit [illegible]

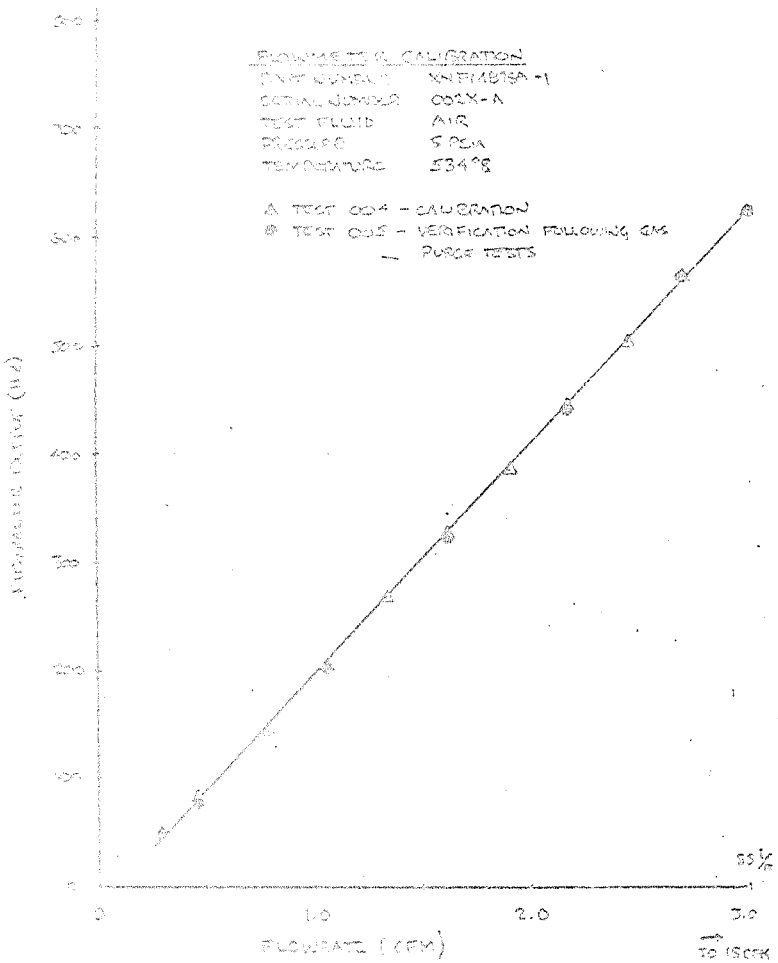


FIGURE 1

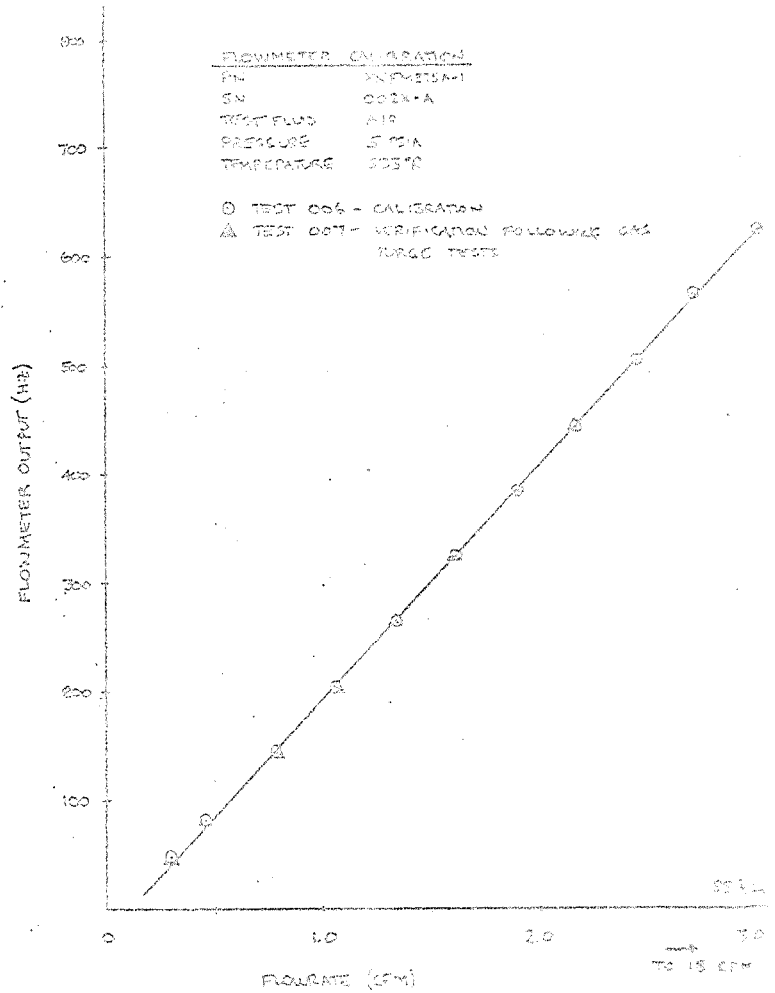


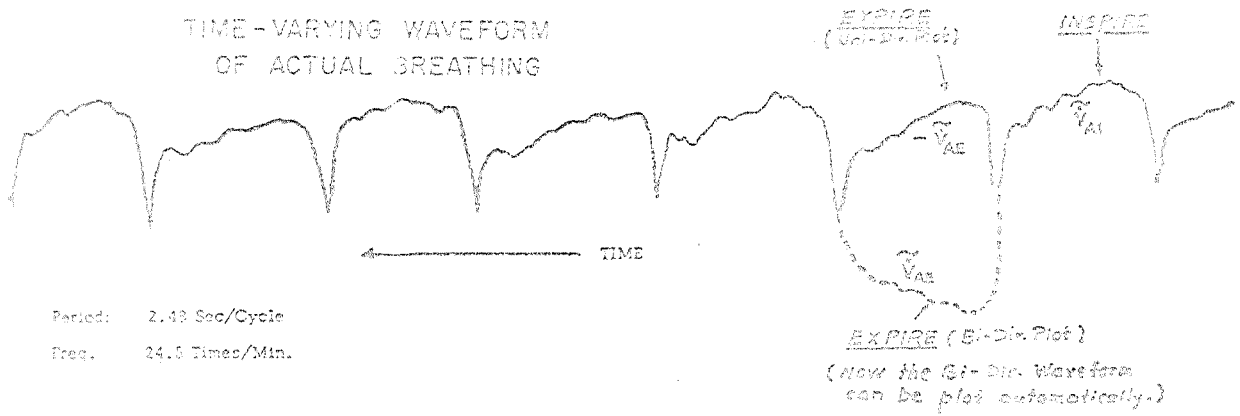
FIGURE 2

Exhibit 2C

[illegible] Respiratory [illegible] Time-Varying Waveform

(Act[illegible] Test Performed on Pa[illegible]k Wallace on August 28, 1969)

TIME-VARYING WAVEFORM
OF ACTUAL BREATHING



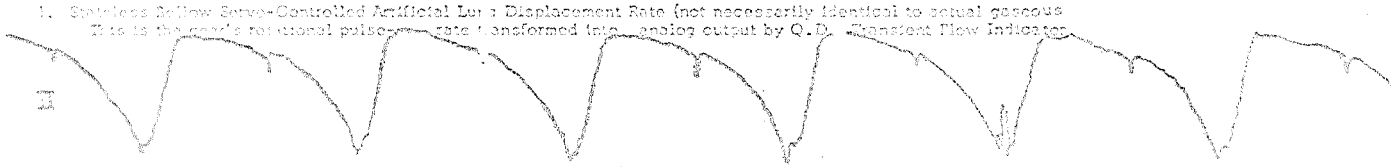
Period: 2.48 Sec/Cycle

Freq. 24.5 Times/Min.

Oscillographic Recording on the Transient Response and
Dynamic Measurement Capabilities of Quantum Dynamics' Bi-Directional Respiratory Flowmeters

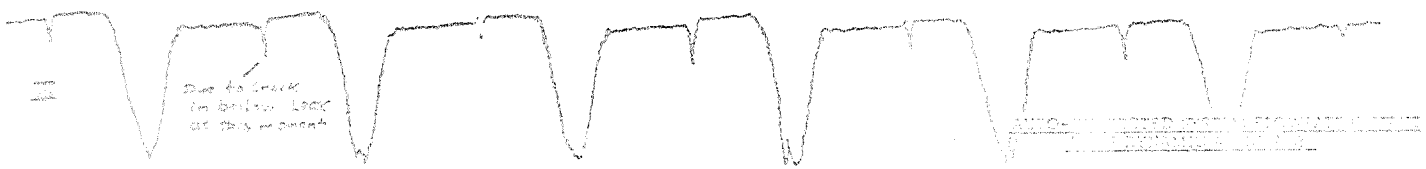
DRIVING FUNCTION

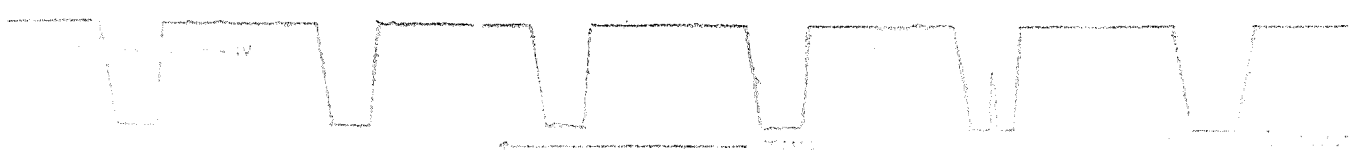
I. Stainless Bellow Servo-Controlled Artificial Lung Displacement Rate (not necessarily identical to actual gaseous
This is the gear's rotational pulse-rate transformed into analog output by Q.D. Transient Flow Indicator



FLOWMETER'S RESPONSE FUNCTION

II. Output Waveform obtained directly from Flowmeter's pulse-rate output after being transformed to analog form
III. Final Electronically auto-corrected Respiratory Flow waveform from the Bi-Directional System for data analysis and computation.





Major Components of a Full-scale MIRACLE System

A complete or full scale MIRACLE II System (abbreviated from Metabolic Instantaneous Respiratory Computing-Logging-Electronic System) is constituted by the following sub-systems or sections:

A. INPUT SECTION

1. Bi-Directional Respiratory Flow Metering Section, consisting of:
   a. QL Bi-Directional Respiratory (RL type) flowmeter;
   b. Bi-Directional Respiratory Flow Direction Discriminator and Electronic Data Gating system.
   c. FPAC (abbrev. for Frequency-from-Period-Analog-Computer) Transient Flow Indicator and Converter.
   d. Bi-Directional Dynamic Flow Auto-Corrector (most rigorous latest design) or Uni-directional Auto-Corrector (existing simpler design).
   e. A voltage-controlled oscillator which receives the analog (or BCD) output from the Auto-Corrector and generates a frequency output which is linearly proportional to the flowrate. This type of output makes possible accurate time integration of flowrate into the desired "volumes" over any long or short totalization period.

   The flowmetering section is self-sufficient to meet the great majority of respiratory and lung measurement requirements.

2. Mass Spectrometer

For the breath-to-breath metabolic measurement on ground, the Environmental Physiology Laboratory at NASA-Manned Spacecraft Center

MAT 3 magnetic disc type mass spectrometer. It provides the partial pressure of O_2, CO_2, N_2, and H_2O as analog outputs. Other suitable mass spectrometers, such as those of the quadruple type, can also be used to provide the partial pressure inputs.

B. INPUT COMPUTING SECTION

1. Mass Flow Computer

This hybrid (digital and analog) multiplier-divider type computer provides an accurate and continuous mass or standard flowrate output in frequency form by precise computation from three simultaneous inputs: a) the pulse-rate (or frequency) output of the flowmetering section; b) the absolute pressure in analog form; and c) the absolute temperature.

2. Partial Pressure Percentage Concentration Computer

Four (4) analog computers receive the analog outputs of the mass spectrometer and transform each partial pressure (O_2, CO_2, N_2 or H_2O) into the percentage concentration (of the total pressure) for subsequent computation.

C. DATA DELAY CONTROLLER

The function of this controller is to correct the time "mismatch" between the flowrate and partial pressure signal. The two outputs of the flowmetering section are delayed by a proper amount so that the flowrate can be computed in correct phase relationship with the partial pressure signals.

D. DATA CONTROL SECTION

This section serves the function of continuously separating, sorting, and classifying the flowmeter and mass spectrometer signals according to whether they are due to inspiratory or expiratory flow; and, to dispatch the resultant signal to the proper stations for subsequent computation. It also automatically determines the T_M period; decides when to begin and when to end each metabolic computational period; and so instructs the various computers to perform their proper computing functions.

It receives its instructions from several sources: the bi-directional discriminator; the flowmeter; the clock pulses, and the pre-set "programming" instruction.

E. REAL-TIME METABOLIC COMPUTER SECTION

1. The O_2 and N_2 Uptake Computers

Each of the uptake computers receives: a) the standard flowrate in pulse-rate form; b) the percentage concentration of O_2 (or N_2); c) the "steering" pulse and other instructions from the Continuous Data Control Section. With these inputs, it computes continuously the volumes due to O_2 (or N_2) and at the same time, computes the uptake by means of "up" and "down" counting process for the prescribed T_M period which can be near 1 minute, or 1/2 minute, or 1/4 minute in accordance to the setting.

2. The CO_2 and H_2O Release Computer

These computers perform in a similar manner as the uptake computers, except that they compute in the reverse direction only for CO_2 and H_2O.

3. The RQ Respiratory Quotient Computer

Two different designs of the RQ Computer are available. The most accurate one is based on all digital design. The continuous indicating R.Q. Computer is based on "hybrid" (digital-analog-digital) design, requiring the use of D-to-A, Sample-and-hold, and A-to-D units.

F. THE COUNTER CONTROL SECTION

The counter control section consists of the following logic units:

1. The Timing and Synchronizing Logic (9-Channel).
2. The 8-channel (16 bits each) Data Storage.
3. The Scanning and Printer Control Logic.
4. The Digital-to-Analog Converter for continuous monitoring of instantaneous metabolic parameters.
5. The Digital Printer or Recorder.
6. The Analog Recorder (optional).

In many cases, it is not necessary to use the entire system for full-scale operation. For instance, the Input Section can be used independently to satisfy a wide range of lung function and respiratory measurements with far more efficiency and adequacy than using a variety of conventional equipment. A full-scale system, however, can provide most, if not all, the required services known to modern research and clinical practices.

I. BI-DIRECTIONAL FLOW-METERING SYSTEM

For The

MEASUREMENT OF RESPIRATORY FLOWRATES AND VOLUMES

General

The metering of human respiratory flow is the first step vital to the measurement of the metabolic parameters. Of the various known types of flowmeters in existence, including those based on thermal, hot-wire anemometer and pressure-drop principles tested thus far, only the advanced turbine type gas flow metering systems developed by Dr. F.F. Liu and his co-workers at Quantum Dynamics have shown excellent accuracy, linearity, as well as safety and simplicity, when used in pure oxygen (and CO) breathing condition. Not only useful for ordinary metabolic measurements at 1 atmosphere, the Liu flowmetering systems proved equally efficient in subatmospheric respiratory measurement well below one atmosphere, for example, 1/3 atmosphere in space flight applications; or, under high pressure breathing conditions such as during diving and deep-submergence applications where helium (98%) and oxygen (2%) are inspired at 40 atmospheres. Under these conditions, other ordinary type of flowmeters become ineffective.

By itself, however, an "elementary" turbine flowmeter cannot meet the dynamic performance requirements. For despite its accuracy in metering steady-state gaseous flow, the "elementary" turbine flowmeter suffers from some performance inadequacy when dealing with certain key respiratory flow parameters in modern research and clinical works.

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1. J. H. Comroe, Jr., R.E. Forster II, A.B. Dubois, W.A. [illegible] L. Carlsen, THE LUNG, Clinical Physiology and [illegible] Tests, [illegible], Chicago, Yearbook Medical [illegible]

As pointed out by Comroe et.al., many such parameters are essentially dynamic in nature (1). Respiratory flow includes both the inspiratory and expiratory flow, with the gas flow direction reversing during each cycle. The frequency (repetition rate RR) of human breathing, with consideration for extended period lung function tests, is between one breath per second and two breaths per minute, while the instantaneous flowrate can vary from .3 to 350 or even 500 liters per minute. Even with the optimal mechanical design of turbine-type flowmeter, the inertial factors of the sensing turbine, when operating in low density gas, will cause dynamic error in breath-by-breath measurement, particularly during the rising and falling portion of the breathing. An accurate flowmetering system, therefore, should have the means to correct such dynamic errors in order to insure that the measurement of breath-by-breath respiratory flowrate is accurate and reliable. Secondly, a turbine flowmeter must provide signal output which can identify itself regarding which portion of the measurement is inspiratory and which portion is expiratory.

In the respiratory flow measuring systems, designed for advanced metabolic studies by Quantum Dynamics, the following features and concepts are included:

a) Measures towards optimizing the mechanical and fluid dynamic performance of the "elementary" turbine meters have been applied methodically to the flowmeter, although limited improvement in dynamic response can be expected. Such measures include the engineering efforts to optimize the moment of inertia, the turbine's geometrical configurations, angle of attack; use of highly sensitive pickup and bearings, and other aero-

dynamic improvements. Most important, absolute safety must be insured when used with pure oxygen; that its performance will not be significantly affected by the presence of water vapor and saliva; and, the flowmeter must also be insensitive to voice conversations. Then, additional effort should be made to achieve, as nearly as possible, the same calibration constant for both inspiring flow and expiring flow.

b) A method and electronic arrangement which enables the flowmeter to identify when it is measuring the inspiratory flowrates or the expiratory flowrate; so that the signals due to the inspiratory flow can be separated from those of the expiratory flow, and vice versa.

c) A new type of electronic correction scheme which, taking into consideration the limited dynamic response capabilities of an "elementary" turbine flowmeter, will automatically perform on-line electronic correction of the dynamic errors due to phenomena such as rise times, over-shooting, coasting, or under-shooting. The same electronic unit also permits simple and accurate analog simulation of the dynamic model of a turbine. Applied in the reverse manner, such a model is used to enable the flowmeter to produce an output waveform which is true to the actual dynamic respiratory flow phenomena.

1.0.0 Overall System Description (see Fig. 1)

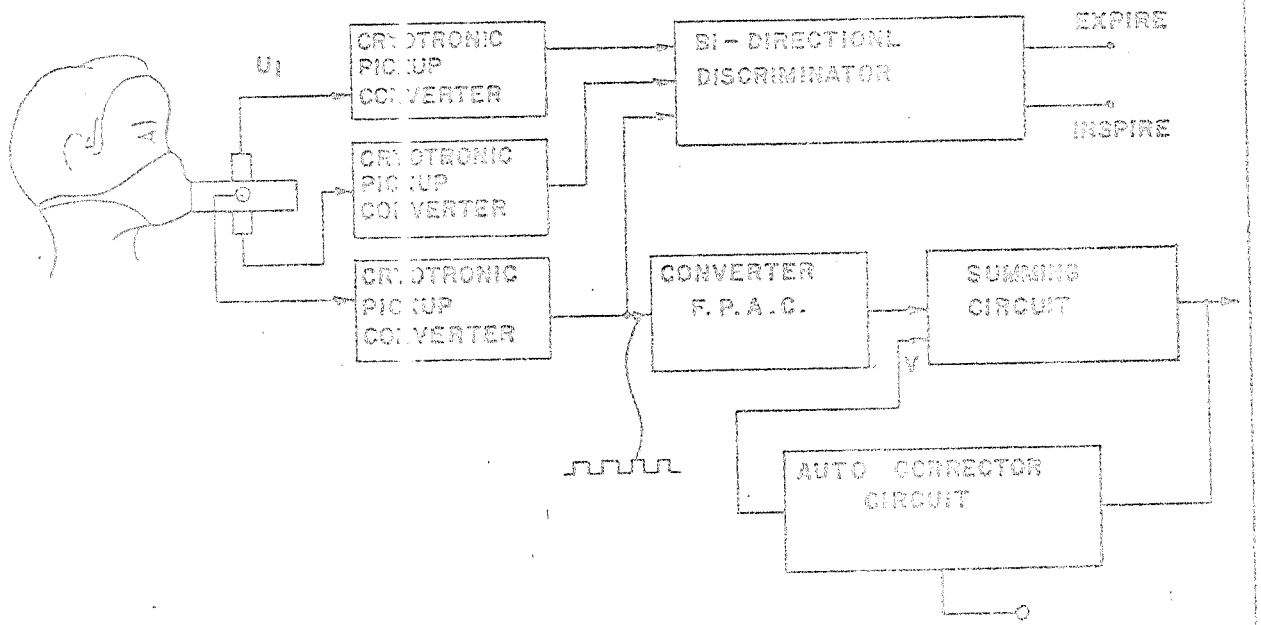
The Bi-directional Respiratory Flowmetering System is composed of several major units, each performing a particular function, be it fluid

dynamical, mechanical or electronic. The high sensitivity respiratory flowmeter (Pat.No.3,135,116) is based on the coaxial indicating-slave turbine design, and presently weighs 6.87 ounces, but this can be reduced by a factor of 2. The turbine weighs less than 0.02 ounces, and is designed to achieve low moment of inertia. Due to the piggie-back-ride bearing arrangement (See Quantum Dynamics Bulletin 5.2) and maximal compliant fluid dynamic design; its rotational motion in response to flow is subject to the lowest possible retarding force.

The turbine's rotational motion is transduced into high amplitude pulse-rate output through the use of the cryotronic pickup which imparts no drag to the freedom of motion. Three such pickups are used in each bi-directional flowmeter for flow measurement as well as for directional discrimination purposes. Since it is of a discrete nature, the pulse-rate signal is transformed into a continuous and instantaneous analog function via the FPAC (Frequency-to-Period-to-Analog Computer) type Transient Flow Indicator - but only for certain necessary functions in the system. The flowmeter is therefore capable of providing both the digital and analog outputs.

In order to correct for the dynamic errors of the turbine's motion, an important instrument, namely, the Dynamic Flow Auto-Corrector, is employed. With this, the output signal can be automatically and continuously corrected in order to reconstruct the actual waveform of the respiratory flow-rate at high resolution and with theoretical soundness.

1. OVER-ALL RESPIRATORY FLOW METERING SYSTEM



1.0.1. BI-DIRECTIONAL RESPIRATORY FLOWMETERS, DESIGN OBJECTIVES

The design of the Bi-directional Respiratory Flowmeter is directed toward achieving the following objectives:

a. Ability to measure the dynamic respiratory flowrate with adequate response, good accuracy and high resolution on a breath-by-breath basis over a range of from 2 to 50 breaths per minute;

b. That the flowrate can be integrated by digital or analog means in order to provide precise measurement of (flow) "volume", such as the "minute volume", "tidal volume", thus making possible such metabolic measurements of uptake, release and R.Q., etc.

It is important to point out that these objectives are to be achieved over a wide range of breathing pressures ranging from one-third (1/3) of an atmosphere to 50 atmospheres. The 1/3 atm. pressure applies to the breathing condition inside space vehicles, which, on account of the low gas density (e.g. 0.025 lb/ft^3) and other factors, renders the measurement of dynamic respiratory more difficult, unless the respiratory flowmeter is provided with an electronic dynamic flow Auto-Corrector; at higher pressures of 1 atmosphere or above, such measurements are technically easier. However, at 40 atm. or above, which corresponds to the breathing pressure of divers and deep submergence vessels, the bi-directional respiratory flowmeter should not introduce discomfort to the user by imparting higher flow resistance than that attributed to the breathing hose; since the pressure drop increases with the density of the breathing gas, which under say 40 atm. pressure, is sufficiently high (e.g. 0.456 lb/ft^3 for 98% He and 2% O_2 mixture). On the other hand, the dynamic response of the respiratory

flowmeter by itself, expressed in terms of "time constant", improves with higher fluid density.

1.0.2. BI-DIRECTIONAL FLOWMETERS, SYSTEM CONCEPT

In the Bi-directional system, such as used in the MIRACLE II System, one respiratory flowmeter (Pat.No.3,135,116) is used for measuring both the inspiratory flow and expiratory flow. The flowmeter generates a pulse-rate or frequency type output when subjected to respiratory flow. To be more specific, during the period of inspiratory flow, the flowmeter generates an output of frequency f_1 which represents the inspiratory volumetric flowrate during this period; when the expiratory flow takes place, the flowmeter generates frequency output f_e. Note, f_1 and f_e are generally time-varying frequencies, thus for each complete breathing cycle, we have:

$$Q_1^i(t) = K_i \int_0^{\tau_i^i} f_i \, dt$$

$$Q_1^e(t) = K_e \int_0^{\tau_1^e} f_e \, dt$$

where Q^i and Q^e are respectively the inspiratory and expiratory "tidal volume"; the superscript i and e denote respectively inspiratory and expiratory, and the subscript 1,2,3..denote Number 1, Number 2, or Number 3 breath cycle, etc. k_i and k_e are the flowmeter's calibration constants in the forward (inspiratory) and reverse (expiratory) flow directions, which, for Quantomics-Liu flowmeters, are generally extremely close to each other, and can be made identical by electronic means; τ^i and τ^e are respectively the inspiratory and expiratory period. The lengths of these periods are determined from the Bi-directional discriminator which generates a pulse whenever the turbine changes the direction of rotation when the the flow is changed from inspiratory to expiratory or vice versa. Since

during the respiration there can be no exhaling during the inhaling period, or vice versa; the circuit is so arranged that, at any time only one frequency f_i or f_e is transmitted for processing. By doing so, two channels of signal are compressed into a single data channel.

This is accomplished by two NAND gates, A and $\bar{A}$. During the inspiratory period, the flow direction discriminator switch sends out a signal of logic state 1 to A, but at the same time applies a signal of logic state 0 to $\bar{A}$: A then allows f_i to pass during the entire period τ^i, while $\bar{A}$ holds off f_e during the same τ^i period.

Conversely, when the flow direction is changed to expiratory flow; the flow direction discriminator sends 0 to A and 1 to $\bar{A}$, allowing f_e to pass through $\bar{A}$ for further processing during the τ^e period.

The operation of A and $\bar{A}$ is therefore analogous to a double-throw switch, allowing the transmittal of one signal at a time through a common processing channel.

1.0.3. Computation of Standard or Mass Flowrate*

Among the advantages of compressing the two data channels into one, is that it greatly facilitates the accuracy and simplicity of standard (or mass) flowrate computation $\dot{Q}^S$ or $(\dot{m})$. The standard flowrate is not only measured by the volumetric flowrate f_i or f_e, but is also dependent on the fluid density and therefore the quantities and the variation of pressure P and temperature T during τ^i or τ^e.

The standard flowrate at any time is given by:

$$\dot{Q}^S = \dot{Q}\left(\frac{P}{P_0}\right)\left(\frac{T_0}{T}\right)$$

NOT REPRODUCIBLE

where P_0 is the reference pressure, generally 1 atm.; and T_0 is the reference temperature, which is either 32^oF (0^oC) or 70^oF.

During computation, the absolute pressure P (in Atm. or PSIA) and the absolute temperature T (in ^oK or ^oR) signals are generated continuously by transducer systems specially designed by Quantum Dynamics for such on-line computation.

In the MINIMIE II System, the auto-corrected frequency output of the flowmetering section provides continuously the quantity of $\dot{Q}$. A hybrid type PT/T Mass Flow Computer receives $\dot{Q}$, P, and T as input and continuously computes the mass flowrate in frequency form for the subsequent computation of metabolic parameters.

* BTPS condition is generally referred to in biomedical practice.

1.0.4. The Measurement of Lung and Ventilation Volumes.

As discussed in previous sections, the breath-by-breath "tidal" volumes of each breath cycle are given by:

$$Q_t^i(t) = K_i \int_0^{\tau_i^i} f_i \, dt \qquad (1)$$

$$Q_t^e(t) = K_e \int_0^{\tau_i^e} f_e \, dt \qquad (2)$$

where Q^i and Q^e are respectively the inspiratory and expiratory "tidal" volume.

The "Minute Volume" inspiratory and expiratory is the totalization of all the (inspiratory and expiratory) volume from $t = 0$ to $t = T_M$ which is the nearest to 1 minute.

$$\text{Minute Volume Inspiratory} = \frac{1}{T_M}\left[\sum_{t=0}^{t=T_M} Q^i(t + T_M)\right] = \frac{K_i}{T_M}\int_0^{T_M} f_i \, dt \qquad (3)$$

$$\text{Minute Volume Expiratory} = \frac{1}{T_M}\left[\sum_{t=\tau_i^i}^{t=T_M} Q^e(t + T_M)\right] = \frac{K_e}{T_M}\int_{\tau_i^i}^{T_M} f_e \, dt \qquad (4)$$

In the MIRACLE II System, the volumes are obtained digitally by totalizing the flowmeter's pulse-rate (or frequency) output over the prescribed period of time. In this form, the volume over any long or short period can be integrated accurately from the flowrate. Explained in a graphic manner: if the time-varying respiratory flowrate is transformed into an analog (voltage) output which is recorded in "voltage vs time" form; then the area under the curve, obtained on a breath-by-breath basis,

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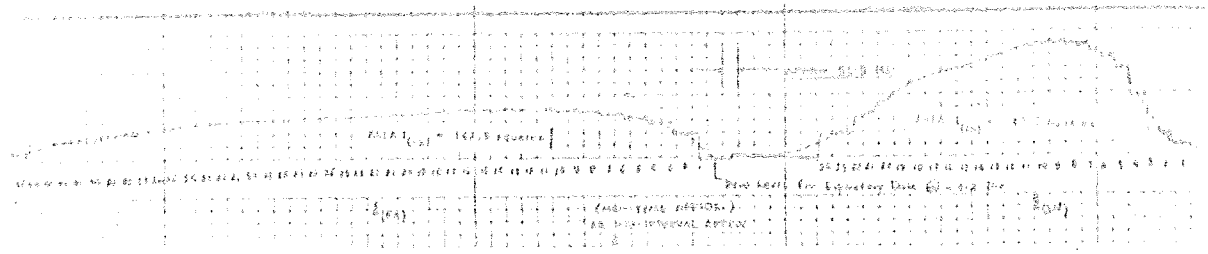

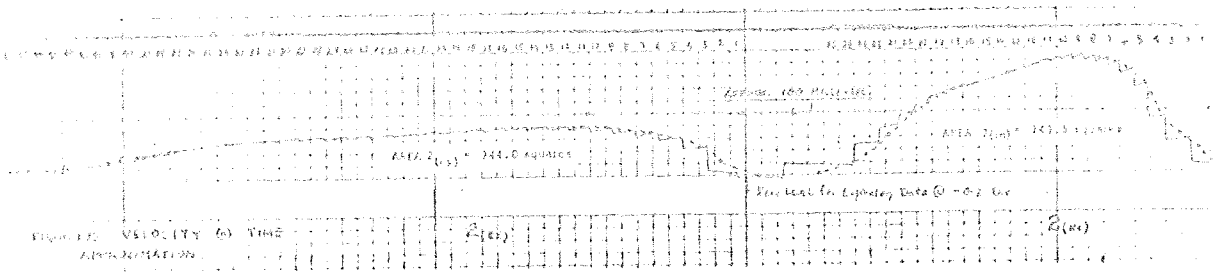

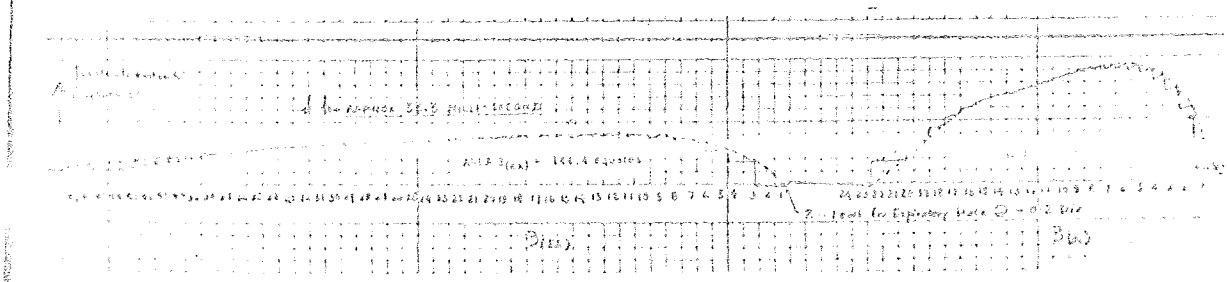

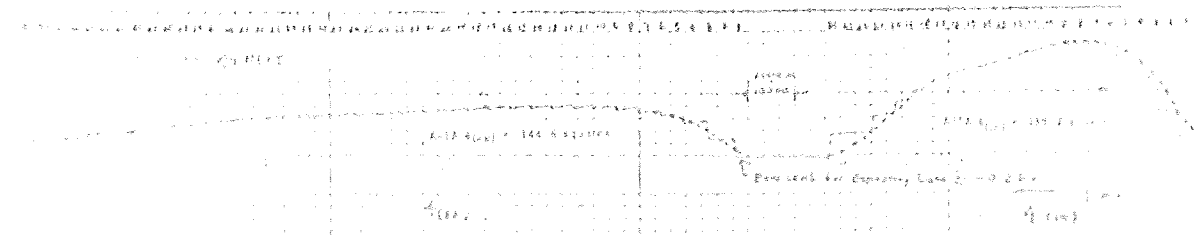

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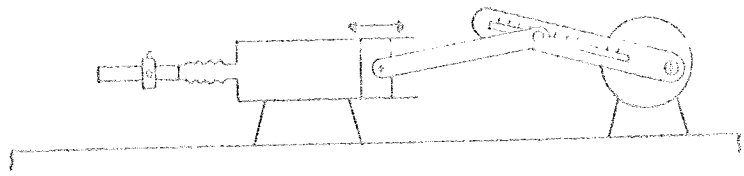
provides the volume per breath, inspiratory or expiratory. Such an area is equivalent to the result of integration of analog flowrate output; or, the totalization of pulse-rate type flowrate output.

1.0.5. ON THE ABILITY OF QL RESPIRATORY FLOWMETER IN TIDAL VOLUME MEASUREMENT BY THE INTEGRATION OF BREATH-BY-BREATH FLOWRATE

The ability of the QL flowmeter in measuring both the breath-by-breath dynamic flowrate and tidal volume can be evaluated conclusively, and in detail, by means of the test method and apparatus used at NASA-MSC Environmental Physiological Laboratory. The recorded data obtained with the high resolution QL 3/4" Bi-directional Respiratory Flowmetering system are shown in Fig. 2

In these tests, a breathing simulating pump manufactured by Harvard Instrument Company is used to provide the flow in both directions. Although



this pump's driving motor does not have a high degree of speed regulation, not have precisely known waveform for its generated (time-varying) flowrate; its piston moves over the same volume in the cylinder in either direction (simulated inspiration and expiration), and the rate of flow in one direction can be made either larger or smaller than that in the other direction, namely, in unsymmetrical cycle mode of operating. This unsymmetrical

flow feature provides a useful testing method to evaluate the breath-by-breath tidal volume measuring capability of the respiratory flowmeter, in addition to the flowmeter's dynamic response. A valid test, therefore, can be formulated on the following basis:

a. Since the Breathing Simulating Pump creates the same volume change in the inspiratory flow direction as in the expiratory flow direction; these two volumes as measured by the flowmeter should be the same, irrespective of the difference (or, lack of difference), between the inspiratory and expiratory flowrates, so that their period may be different. If the flowmeter's measured volumes for the inspiratory and expiratory periods proves to be identical, or very close to each other in the presence of motor speed variation; then this is a valid proof of the volume measuring capability of the flowmeter. If the measured volumes are substantially different, or far off, then the ability of the flowmeter in measuring breath-by-breath volumes is questionable.
The testing result shown in Fig.2 verifies that the Qi Flowmeter's measured inspiratory and expiratory volumes of the same breathing cycle are practically identical by virtue of nearly equal "integrated area" under the curves;

b. As the (tidal) volume is the result of integration of the time-varying flowrate; the flowmeter must, first of all, have the requisite time constant and dynamic response to breath-by-breath flowrate variation, more specifically, to cover the frequency spectrum of such rapid variation. Otherwise, accurate volume integration results cannot be expected, particularly when the inspiratory and expiratory

flowrate and/or periods are different. If the measured (tidal) volumes for the inspiratory and expiratory periods for the aforementioned tests prove to be the same, so that requirement a. is satisfied; then it is a conclusive proof that the flowmeter must have adequate breath-by-breath dynamic response. It means that the transient response of the flowmeter is adequate to reproduce the rapid flowrate variations contained in the breath-by-breath waveform whether it be under the same or different flowrate and period conditions. In the series of tests shown in Fig. 2 , the expiratory flowrate is made as low as 50% of the inspiratory flowrate, yet the measured tidal volumes have proven to be of good accuracy, and that they agree with the designed test condition by proving that the measured inspiratory and expiratory volumes remain the same.

c. The test criterion described in a. and b. is the identicalness between the inspiratory and expiratory volumes; however, another useful criterion is to compare the measured inspiratory (or expiratory) volume of one breathing cycle to that of the next cycle. If the motor speed of the simulating pump is of highly regulated constancy, then the volumes of consecutive breathing cycles can be assumed as nearly the same. The result shown in Fig. 2 again met this test objective. Moreover, because of its dynamic response, the QL flowmeter was able to depict some speed fluctuations of the Harvard Breathing simulating pump including its period and waveform changes. These results point out the reproducibility

of the flowmeter in breath-by-breath measurement: for without adequate dynamic response to the rapidly varying driving function, and volume integrating capability, such reproducibility would have been inconceivable.

1.0.6. ANALYSIS OF PRESENT TEST DATA

In Fig. 2 the flowrate is shown along the Y axis, and is the analog output of the flowmeter's Transient Flow Indicator and Uni-directional Auto-Corrector. In order to obtain maximum resolution to data analysis, the gain of the recorder was adjusted to provide nearly full-scale deflection at the maximum inspiratory flowrate. This same gain applies to both the inspiratory and expiratory flowrate. The maximum chart speed available to the 4-channel Brush Recorder is used in order to further facilitate the more precise determination and comparison of the integrated areas under the "flowrate vs time" curves.

The analog output shown in Fig. 2 is actually in "quantized" or sampled data form, since the FPAC Transient Flow Indicator provides this analog flowrate output by computing the reciprocal of each time interval due to the passage of two consecutive turbine blades. At higher flowrate, the signal appears to be virtually continuous. In the subsequent differential equation computation process by the Auto-Corrector, the "quantized" output of the FPAC was thus retained in similar form, but not in quantitative value. Although it improves the frequency and transient response by a factor of 10 times; and quantitatively speaking, provides more accurate and truthful waveform for the

breath-by-breath flowrates; the uni-directional model of Auto-Corrector may introduce some inaccuracies near the zero-crossing regions.

In order to determine graphically the enclosed areas, the continuous curve is obtained from the quantized, or step-like, curve with four different approximation methods; but only for the experimental purpose of comparing the resultant inspiratory and expiratory volumes of the same breathing cycle; and to assess the maximum discrepancy which may be caused by the different methods. There is no significant discrepancy noticed in the resultant area measurement as determined by different methods. (See p. A-22,23)

The most rigorous approximation method, in conformance with the operating principle of the TPAC Transient Flow Indicator, is the "Peak and Reverse Peak Method", since the amplitude of each steep-rising line actually represents the averaged flowrate during its preceding interval. As the interval shortens the averaging period is so small that the amplitude distribution actually becomes the continuous and instantaneous value of the flowrate. The other methods use the mid-interval, mid-height, and mid-interval/mid-height approaches to trace the continuous flowrate vs time curve.

For each vertical section constituted by one horizontal time division (equal to 5 mm and equivalent to 33.33 millisecond), a mean (flowrate) amplitude is carefully determined under microscope from the continuous curve for that particular vertical section. The area for each vertical section is then calculated from the best estimate of this mean amplitude and the base length. The totalization of such sector area over an entire inspiratory or expiratory period results into the (tidal) volume.

Based on the high resolution recording shown in Fig. 2 the data analysis is tabulated in the following pages in detail and summary form.

TEST DATA ANALYSIS DIRECTED TOWARD THE
BREATH-BY-BREATH VOLUME MEASUREMENT
AND DYNAMIC RESPONSE CAPABILITIES OF

HIGH RESOLUTION OL BI-DIRECTIONAL RESPIRATORY 3/4-in. FLOWMETER

(Data Obtained with Harvard Instrument Breathing Simulating Pump Operating at 23 to 24 RPM, on 14 October, 1970, Under the Supervision of Dr. J. Rummel of NASA-MSC)

TIME * INTERVAL (Div.) #	MEAN FLOWRATE FOR THE INTERVAL (Division) ** INSPIRATORY					EXPIRATORY			
	#1 Ins.	#2 Ins.	#3 Ins.	#4 Ins.	#5 Ins.	#1 Ex.	#2 Ex.	#3 Ex.	#4 Ex.
0		1.0	0.9	1.0		0.1			0.2
1	1.0	1.6	4.0	2.0	1.0	1.0	0.4	0.2	0.4
2	1.6	2.5	6.6	3.8	2.0	1.8	1.0	1.1	1.0
3	3.0	3.4	8.0	5.7	3.8	2.4	2.0	2.0	1.9
4	5.0	5.0	8.8	7.1	5.8	2.8	2.6	2.5	2.6
5	7.0	7.0	9.3	7.9	7.4	3.3	3.0	3.1	3.1
6	7.9	8.2	9.4	8.6	8.0	3.4	3.4	3.4	3.4
7	8.6	8.8	9.3	8.8	8.6	3.6	3.5	3.6	3.6
8	8.8	9.2	9.1	9.0	8.9	3.8	3.6	3.8	3.8
9	9.0	9.2	8.8	8.9	9.0	3.9	3.7	3.9	3.9
10	8.8	9.2	8.4	8.6	8.8	4.0	3.8	3.9	3.9
11	8.6	9.0	7.9	8.3	8.5	4.0	4.0	4.0	3.9
12	8.4	8.5	7.6	7.8	8.2	4.0	4.0	4.0	3.9
13	8.0	8.2	7.2	7.5	7.8	4.0	4.0	4.0	3.9
14	7.6	7.8	7.0	7.1	7.3	4.0	4.0	4.0	3.9
15	7.2	7.4	6.6	6.7	7.0	3.9	4.0	4.0	3.9
16	6.8	7.0	6.0	6.5	6.7	3.9	4.0	4.0	3.9
17	6.6	6.6	5.0	6.1	6.4	3.8	3.9	3.9	3.8
18	6.1	6.0	4.2	5.4	5.7	3.8	3.9	3.9	3.8
19	5.5	5.4	3.0	4.5	4.8	3.7	3.8	3.8	3.7
20	4.6	4.4	1.8	3.4	4.1	3.6	3.7	3.7	3.7
21	3.7	3.3	1.2	2.5	3.3	3.5	3.6	3.6	3.6
22	2.8	2.3	.2	1.3	2.2	3.5	3.5	3.6	3.5
23	2.0	1.3		0.6	1.6	3.4	3.5	3.6	3.5
24	1.4	0.8		.1	1.0	3.4	3.4	3.5	3.4
25		0.4			0.8	3.4	3.3	3.5	3.3
26			.3		0.5	3.3	3.3	3.4	3.3
27					0.3	3.3	3.3	3.4	3.3
28						3.3	3.3	3.4	3.3
29						3.2	3.3	3.3	3.2
30						3.2	3.2	3.2	3.2
31						3.0	3.1	3.1	3.1
32						2.9	3.0	3.1	3.0
33						2.8	3.0	3.0	3.0
34						2.7	2.8	2.9	2.8
35						2.5	2.7	2.8	2.7
36						2.4	2.5	2.6	2.5
37						2.4	2.4	2.5	2.5
38						2.2	2.2	2.3	2.3
39						1.9	2.1	2.2	2.2
40						1.8	1.8	1.9	1.9

TIME* INTERVAL (Div.)	MEAN FLOWRATE FOR THE INTERVAL (Division) ** INSPIRATORY					EXPIRATORY			
	#1 Ins.	#2 Ins.	#3 Ins.	#4 Ins.	#5 Ins.	#1 Ex.	#2 Ex.	#3 Ex.	#4 Ex.
41						1.7	1.6	1.7	1.8
42						1.6	1.5	1.5	1.7
43						1.2	1.4	1.4	1.5
44						1.0	1.4	1.1	1.2
45						0.7	1.1	1.1	1.2
46						0.6	1.0	1.1	1.2
47							1.0	1.2	1.0
48								1.0	
Area Under The Curve (in Squares)	140.0	143.5	140.3	139.2	139.6	133.7	134.6	134.8	135.4
Area Adjusted for Zero Level-shift at - 0.2 Div.						142.9	144.0	144.4	144.8

SUMMARY OF THE ANALYSIS OF RECORDED DATA

Breathing Cycle	Method of Approximation in Numerical Analysis used	Measured Volume in Terms of the Integrated Area Under the Flowrate vs Time Curve (in Arbitrary Unit of Squares for Comparitive Purposes)	
		INSPIRATORY	EXPIRATORY
# 1	Mid-Interval Profile Approximation	140.0	142.9
# 2	Mid-height Profile Approximation	143.5	144.0
# 3	Rigorous Peak & Reverse Peak Profile Approximation.	140.3	144.4
# 4		139.2	144.8
# 5	Mid-height Profile Approximation	139.6	

The tabulated data and its summary show that: (A) a high degree of agreement exists between each measured inspiratory (and expiratory) volume; and, (B) between the inspiratory (or expiratory) volume of different breathing cycles. These data confirm the ability of the flowmeter to measure accurately the (tidal) volumes on a breath-by-breath basis, and sufficiently short time constant to cover the frequency spectrum and rapid transient encountered in respiratory flowrate measurement.

The test evaluation techniques above are analogous to the auto and cross correlation techniques used in information theories. With the re-activation of the more precise and waveform-controllable "artificial lung" type calibrator at Quantum Dynamics, and the introduction of far more accurate new Bi-directional type Auto-Corrector (in place of the present Uni-directional Auto-Corrector which is, in principle, not strictly compatible with the Bi-directional flowmeter) the ability of the QL Bi-directional flowmeter in breath-by-breath flowrate and dynamic tidal volume measurement can be further improved by a wide margin.

1.0.7. ADDITIONAL PROOF OF THE BREATH-BY-BREATH VOLUME MEASURING CAPABILITIES OF THE FLOW-METERING SYSTEM

One of the tests performed on the QL flowmetering system at NASA-MSC Environmental Physiology Laboratory on 13 October, 1970, was to evaluate the response capability of the flowmetering system by driving the Harvard Breathing Simulating Pump at 60 breaths-per-minute - considerably in excess of the normal breathing frequencies of men. The recorded data (Fig. 2 A) show that the Quantum Dynamics flowmetering system has no difficulty in reproducing the waveform of the 60 breaths-per-minute flowrate. Subsequent calculations indicate that the flowmetering system has the requisite dynamic response to measure breath-by-breath flow even at 85 breaths-per-minute rate.

In the recorded data, the "flowrate vs time" waveforms at 60 breath/min. are reproducible, although at the direction change-over points, the present Uni-directional Auto-Corrector introduced an unwarranted ambiguity and "differentiation" distortion, which would be totally absent when the new Bi-directional Auto-Corrector is used. The recorded data again proved both the volume and flowrate measuring capability of the Bi-directional flowmeters. From the data, which were not recorded at high resolution, the amplitude (peak instantaneous flowrate) of each expiration is shown about 6% less than that of the inspiration, yet the integrated volumes, as represented by the enclosed area under each curve, are the same.

Further proof of the accuracy of volume measurement is the direct measurement of the tidal volume by the High-resolution 3/4-in. Bi-directional flowmeter by means of the digital counter's reading of totalizer (Fig.

pulse count output of the flowmetering system, again using the Harvard Simulating Pump. After a first trial when we mis-read the instruction book and used the wrong calibration constant, a correct direct volume measurement was accomplished and verified as follows:

a) Total Count per Breathing Cycle as registered by counter: 1,200 pulses

b) Setting of the Transient Flow Indicator: # 2

c) Calibration Constant: 1 pulse = 2 cc of setting #2

d) Calculated Volume: $1.2 \times 10^3 \times 2 \times 10^{-3} = 2.4$ Liter

e) At that particular pump setting, the volume change per cycle is given as 2.4 Liter.

The measured volume reading and the given pump volume change thus agree.

One additional test was made by Dr. Rummel using a calibrated hand-pump in a double stroke. The QL Flowmeter registered a total count of 1500 pulses. Again, using the calibration constant of 2 cc per pulse, we obtained a total volume change of 3000 cc or 3 Liters. This corresponds to a cycle of two complete strokes in a 1500 cc calibrated pump.

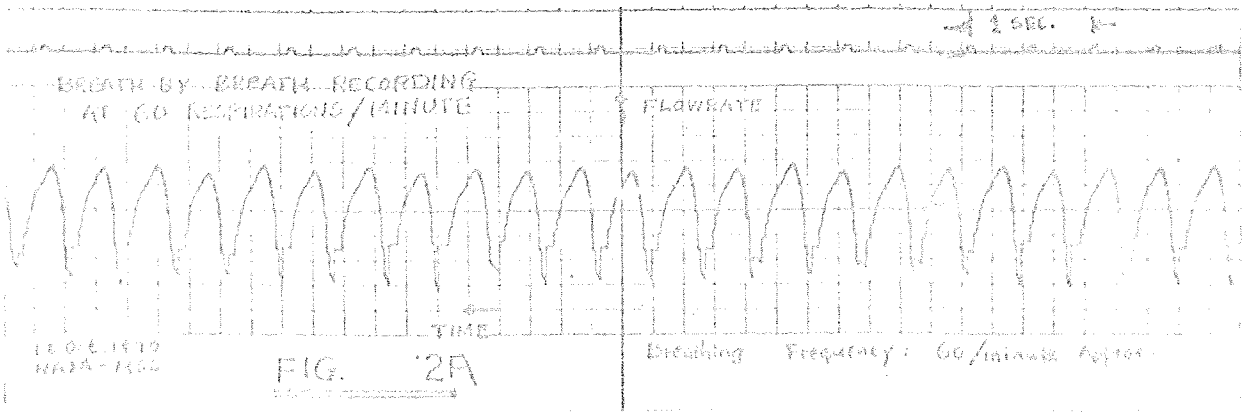


FIG. 2A

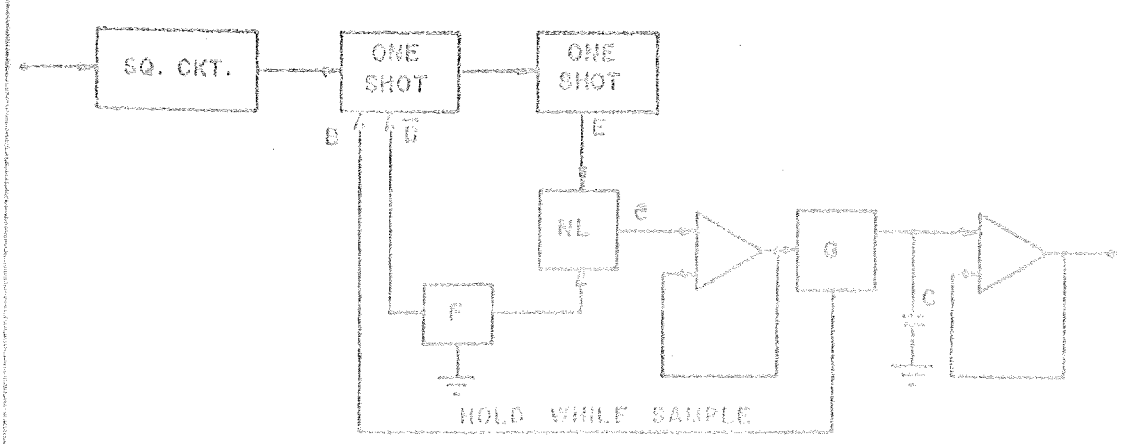
60 Respirations per Minute Flowrate Recording

1.1. NON-LINEAR FREQUENCY FROM PERIOD - ANALOG - COMPUTER and TRANSIENT FLOW INDICATOR

The frequency output of the respiratory flowmeter is converted into a time-varying analog signal by means of the Transient Flow Indicator, or Frequency-from-Period-Analog Computer. This instrument provides a continuous 0 to 5 volt output which is linearly proportional to the flow-rate. Essentially, a non-linear analog computer, the FPAC accepts the flowmeter's pulse-train output; acts on the period T of each cycle; computes the inverse of the time period $e_F = 1/T$; and, holds the information for the period of the next cycle. Thus, the output voltage is in sampled data, or quantized form: each successive level of its time-varying output is linearly proportional to the input frequency F (= 1/T) of the sampled period, and computed once for every cycle. As such, the unit has a much faster conversion speed in comparison with any conventional frequency-to-DC-Converter, which relies on the integration of a large number of cycles for its output.

A block diagram of the FPAC is shown below: (Fig. 3)

FIG. 3 FPAC CONVERTER AND TRANSIENT FLOW INDICATOR



Input F can be a square-wave type frequency output from the flowmeter, or it can be an ac signal shaped into a square-wave by the built-in squaring circuit. The one-shot circuits provide potentials for sampling and resetting: B for set and holding; $\bar{B}$ for reset; E controls the timing and charging the condenser of the nonlinear network. This nonlinear network provides a voltage $e = k_3/t$, where t is time and k_3 is a constant fixed by design at a value of 0.01.

Upon receipt of the first pulse, B is set for sampling. During the sampling period, namely when the push-pull one-shot are operating in B-on-$\bar{B}$-off condition, the potential B is present from the beginning of the period $t = T = 1/F$. Simultaneously, potential E charges the condenser of the nonlinear network, which results in a potential $e = k_3/T = k_3F$. This potential $= K_3F$ is held for the duration of the sampling period during which gates G is on, while F is off, and hold capacitor C is charged to the voltage $e = k_3F$. This voltage is held until the beginning of the next sampling time.

When the sampling is finished, B is turned off and $\bar{B}$ is on: voltage e is reset to a fixed level. When the reset time is over, voltage e begins dropping according to the equation $e = k_3/t$. At time $t = 1/F$, sampling occurs again, and the sequence of operation is repeated.

In the existing flowmetering system, a three-range FPAC converter is used. For example:

Setting	Flow Range	Frequency Range Covered	FPAC Output
#1	0 -108 Liter/m	0 to 500 Hz	0 to 5 volt
#2	0-215 "	0 to 1000 Hz	0 to 5 volt
#3	0-430 "	0 to 2000 Hz	0 to 5 volt

With newer units, auto-ranging type FPAC Converters will be used to replace the two-setting units.

1.1.1. ON THE DIFFERENCE BETWEEN THE INSTANTANEOUS FLOWRATE AND THE SAMPLE AVERAGE FLOWRATE OUTPUT FROM FPAC CONVERTER

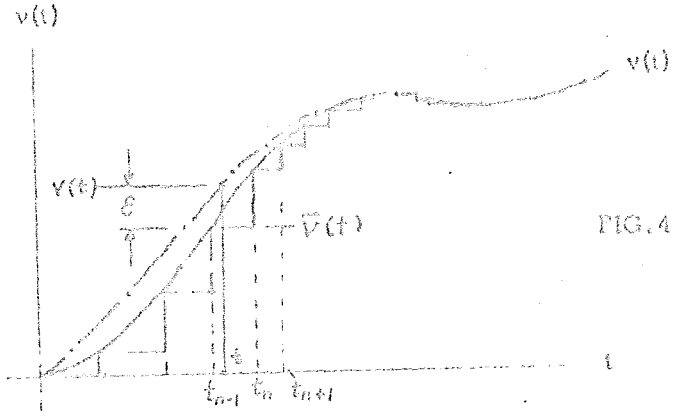


FIG. 4

Using the FPAC (Frequency-from-Period Analog Computer) for transient flow measurement, the average velocity of the turbine is measured once for each sampling period. With each of these sampling periods, a value for the average linear flow velocity $\bar{v}(t)$ can be associated. This is shown by the height of each vertical line trailing its preceding segment of time. The discrepancy between this average velocity $\bar{v}(t)$ and the true velocity $v(t)$ is of concern only when the sampling period is sufficiently long.

$$\tau_n = (t_{n-1}, t_n)$$

If the sampling time intervals $|\tau_n| = (t_n - t_{n-1})$ are chosen to be small enough so that the velocity $v(t)$ can be considered as linear, we have

$$v(t) = v(t_{n-1}) + a\,(t - t_{n-1})$$

for $t_{n-1} \leq t \leq t_n$.

The average velocity $\bar{v}(t)$ for the same time interval is given by:

$$\bar{v}(t) = \frac{1}{2}\left[v(t_{n-1}) + v(t_n)\right]$$

for $t \in \mathcal{T}_n$ ($\mathcal{T}_n$ being the half-open half-closed interval $\mathcal{T}_n = (t_{n-1}, t_n]$)

Thus, the difference between $v(t)$ and $\bar{v}(t)$ is given as:

$$v(t) - \bar{v}(t) = v(t_{n-1}) + a(t - t_{n-1}) - \frac{1}{2}\left[v(t_{n-1}) + v(t_n)\right]$$

$$\boxed{\mathcal{E} = v(t) - \bar{v}(t) = \frac{1}{2}v(t_{n-1}) + a\,t - a\,t_{n-1} - \frac{1}{2}v(t_n)}$$

Thus, the error, when taken at t_n is equal to:

$$\mathcal{E} = v(t) - \bar{v}(t) = v(t_{n-1}) + a(t_n - t_{n-1}) - \frac{1}{2}\left[v(t_{n-1}) + v(t_n)\right]$$

$$= \frac{1}{2}v(t_{n-1}) + a(t_n - t_{n-1}) - \frac{1}{2}v(t_n).$$

This particular value $t = t_n$ is especially significant.

It must also be remarked, that $v(t) - \bar{v}(t) \geq 0$ for $\dot{v}(t) > 0$

$v(t) - \bar{v}(t) \leq 0$ for $\dot{v}(t) \leq 0$.

In summary,

$$\boxed{\begin{array}{l} v(t) - \bar{v}(t) = \frac{1}{2}\left[v(t_{n-1}) - v(t_n)\right] + a(t - t_{n-1}) \\ \text{for any } t \text{ with } t_{n-1} \leq t \leq t_n \end{array}}$$

Although the FPAC Converter provided a "quantized" of sample AVERAGE data for the flowrate, it is significant to point out that this approach is considered highly accurate as far as the volume measurement is concerned, by virtue of the fact that this "sectionalized" integration shows virtually no difference with those obtainable from continuous integration.

1.2. The Dynamic Flow Auto-Corrector

As mentioned earlier, the dynamic behavior of the turbine flowmeter is restricted by certain inherent limitations which cannot be improved upon, even with the best possible mechanical and fluid dynamic design. However, if the mathematical relationship governing the dynamic response of the turbine is known, electronic techniques can be applied effectively to correct for the dynamic errors of the flowmeter.

The dynamic motion of the turbine can be analyzed rigorously by means of complex integro-differential equations wherein all pertinent fluid dynamic and mechanical factors are considered. The complexities, however, can be simplified by setting up a mathematical model of describing function, which is derived with adequate mathematical rigor, yet conducive to electronic computation. Methodologically, this involves continuous on-line computation, continuous comparison and correction of the flowmeter's output according to the mathematical model. Although the model is known in its precise nonlinear differential equation form, it is further subjected to experimental verification and adjustment by means of an artificial lung dynamic calibrator. The latter supplies precisely known dynamic flowrate inputs whose waveforms contain frequency components even higher than those obtainable in human breathing processes. Based on the confirmed model, the Auto-Corrector performs instantaneous computation resulting in a corrected final output which is an accurate representation of the true dynamic flowrate of the respiration; not only as a breath-by-breath basis, but virtually with point-by-point faithfulness.

1.2.1. UNI-DIRECTIONAL AND BI-DIRECTIONAL AUTO-CORRECTOR

The mathematical equations governing the dynamic response of the turbine to respiratory gas flow, have been studied by Liu and other investigators. His work, later confirmed by experiments, has demonstrated that the dynamic error of the flowmeter's measurement can be corrected by electronic methods. Briefly, this is done by means of on-line simulation and correction of the signal emerging directly from the flowmeter's pickup. The essential element of this system is a flowrate simulator model. This is an analog/digital computer which simulates the motion of the turbine flowmeter as described by a rigorous equation. The other major component consists of a non-linear Frequency-to-Period-to-Analog Computer which renders discrete pulse-rate signal of the flowmeter into a continuous analog function.

Theoretically, the Bi-directional Auto-Corrector is the most rigorous means of providing accurate and faithful measurement of respiratory flowrate and volume, although the simpler uni-directional version already in use has already improved the frequency and transient response of the flowmeter by a factor of 10 times.

A block diagram of the uni-directional auto-corrector is shown as follows:

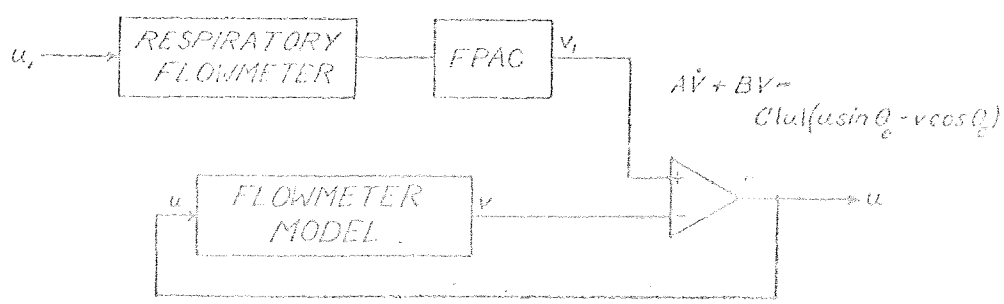
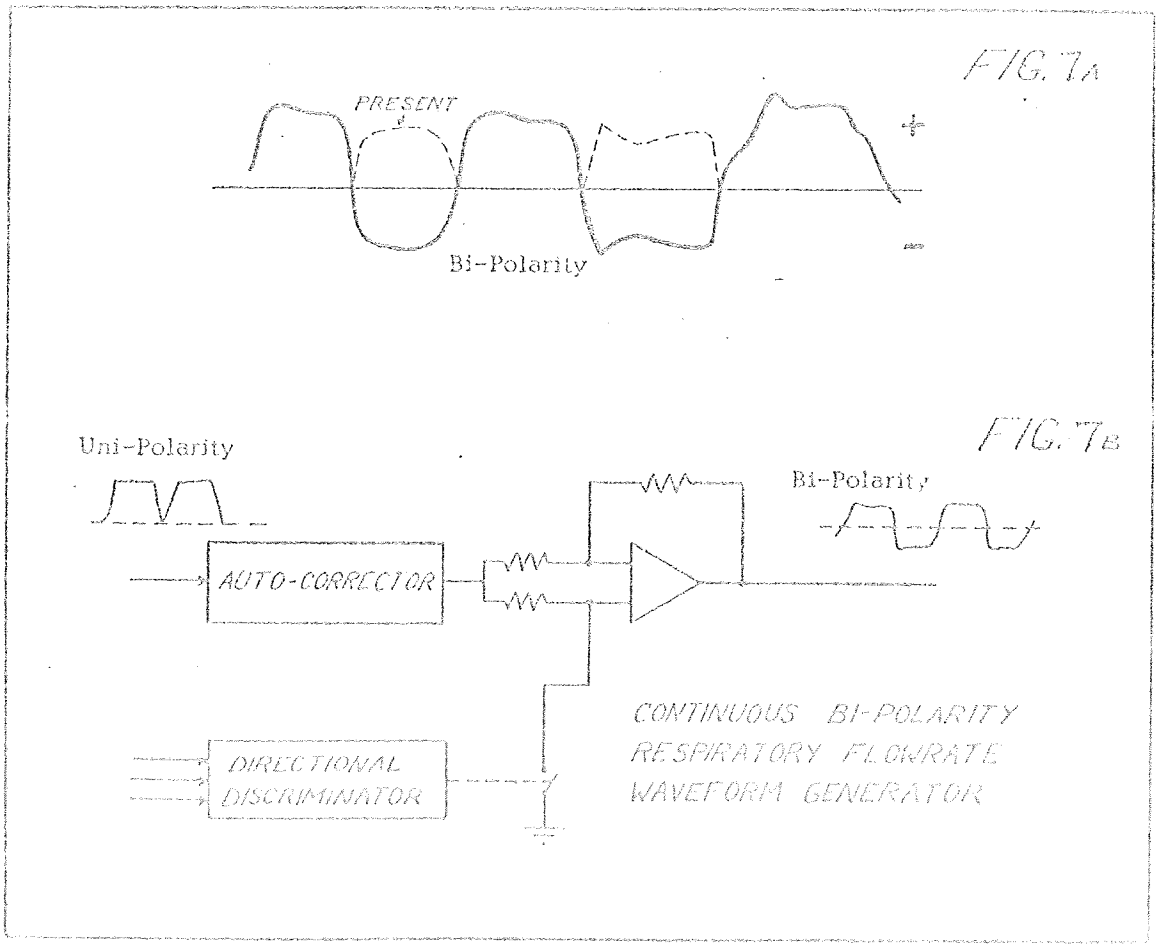
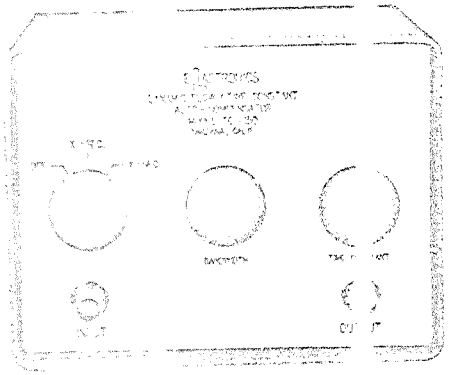


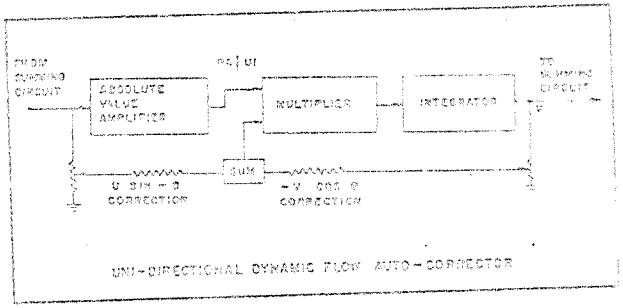
FIG. 5 UNI-DIRECTIONAL AUTO-CORRECTOR (BLOCK DIAGRAM)

As shown in Fig. 7A and 7B. , the Uni-Directional Auto-Corrector deals with signals of a single-polarity nature, not Bi-polarity signals.

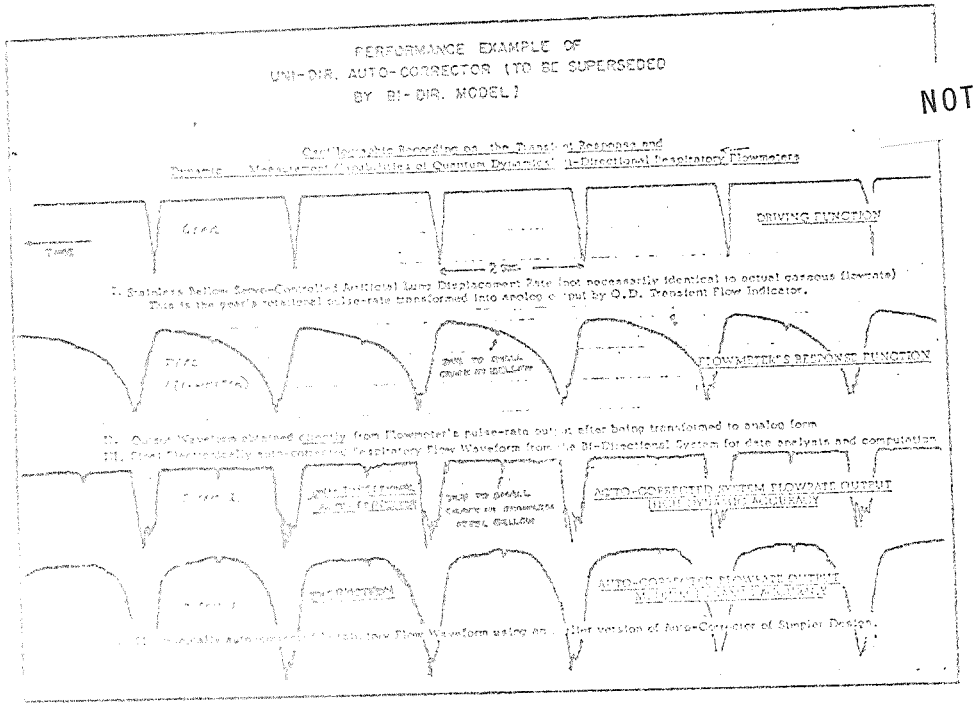
Using this uni-directional simulator model, the actual correction is carried out by the circuit as shown in Fig. 5 . With this system, tests were made at various flowrates. The recordings in Fig. 6 indicate the input signal and the flowmeter signal, both before and after correction. Theoretically, the bi-directional Auto-Corrector will further improve the accuracy and fidelity of the measurement by a very wide margin so that efficient breath-by-breath measurement is no longer a difficult problem.



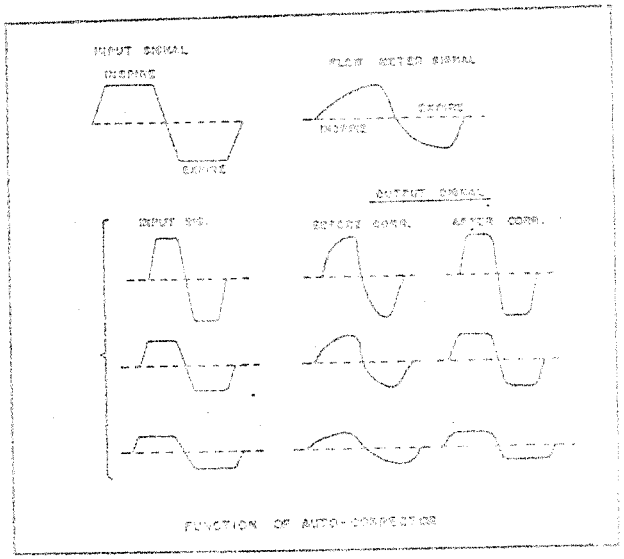




UNI-DIRECTIONAL DYNAMIC FLOW AUTO-CORRECTOR



NOT REPRODUCIBLE



FUNCTION OF AUTO-CORRECTOR

1.2.2. Mathematical Model of Uni-Directional Auto-Conector

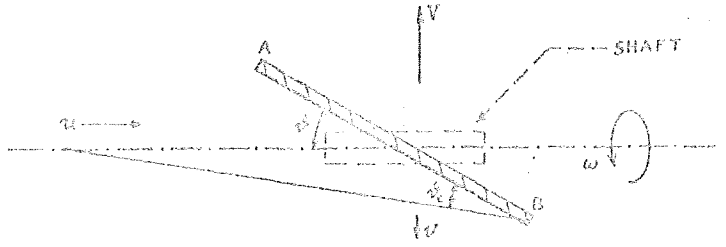


FIG. 8

Fig. 8 is a diagram showing a cross-section of the turbine blade's tip, as viewed along the radial direction from a position atop of the blade's tip. The fluid flow is moving in a direction parallel to the axis at velocity u ; as a result of the turbine's rotation at angular velocity ω , the tip of the blade moves in a plane perpendicular to the axis at a tangential velocity V .

The blade has an angle of attack*, or blade angle, θ . However, during the transient stage, namely during its accelerating and decelerating period, there is an effective blade angle θ_e which is of the same value as θ as soon as ω assumes a constant value. Because of the inertial and frictional factors, v is assumed to be a velocity "lag" component, which actually represents the dynamic error. The following gives the notation used in the analysis:

---

* The so-called "tip angle" is used for initial analytical purpose only. For blades of a spiral-twist configuration, θ is actually the mathematical mean value of the length of blade under question.

AB	=	A cross section of the blade
a	=	Area of cross-section of the flowmeter's duct
ϑ	=	Angle of attack
ϑ_e	=	Effective Angle of Attack
r	=	Effective radius of the turbine blade
ρ	=	Density of fluid.
u	=	Actual fluid velocity
V	=	Indicated velocity in the tangential direction
v	=	The velocity "lag" in a direction opposite to V
ω	=	Angular velocity of the turbine
K	=	Frictional constant
m	=	Mass of the Fluid

Considering the tip velocity V in the direction of motion of the turbine:

$$V = u \tan \vartheta_e - v \tag{1}$$

and the mass flowrate through the meter is:

$$\frac{dm}{dt} = \rho a |u| \tag{2}$$

Thus, the torque applied to the area of the turbine blade is given by:

$$T = F \cdot r = \left[\frac{d}{dt}(mV) \right] \cdot r = \left[\frac{dm}{dt} \cdot V + m \cdot \frac{dV}{dt} \right] \cdot r$$

$$= \left[\frac{dm}{dt} V + \left(m \cdot \frac{du}{dt} \cdot \tan \vartheta_e + m \cdot \frac{dv}{dt} \right) \right] \cdot r \tag{3}$$

Since $m \frac{du}{dt} \tan \vartheta_e$ and $m \frac{dv}{dt}$ are virtually equal and opposite forces, it can be assumed that they really cancel out each other. Consequently,

$$T = V \frac{dm}{dt} \cdot r = \rho a |u| r (u \tan \vartheta_e - v)$$

However, since torque is equal to the rate of change of momentum, we have $T = I\frac{d\omega}{dt}$, and allowing for a small friction $K\omega$, we have

$$T = I\frac{d\omega}{dt} + K\omega \tag{4}$$

That is:

$$T = I\frac{d\omega}{dt} + K\omega = \rho a|u|r(u\tan\vartheta_0 - v) \tag{5}$$

Multiplying all terms by $\cos\vartheta_0$ we obtain:

$$I\cos\vartheta_0\frac{d\omega}{dt} + K\omega\cos\vartheta_0 = \rho a|u|r(u\sin\vartheta_0 - v\cos\vartheta_0)$$

Since $\omega = \frac{V}{r}$ and thus $\frac{d\omega}{dt} = \frac{r\frac{dV}{dt} - V\frac{dr}{dt}}{r^2}$

and as $\frac{dr}{dt} = 0$, we have $\frac{d\omega}{dt} = \frac{dV}{dt}\cdot\frac{r^2}{r}$

Thus:

$$\frac{I\cos\vartheta_0\frac{dV}{dt}}{r} + K\frac{V}{r}\cos\vartheta_0 = \rho a|u|(u\sin\vartheta_0 - v\cos\vartheta_0)r \tag{6}$$

or

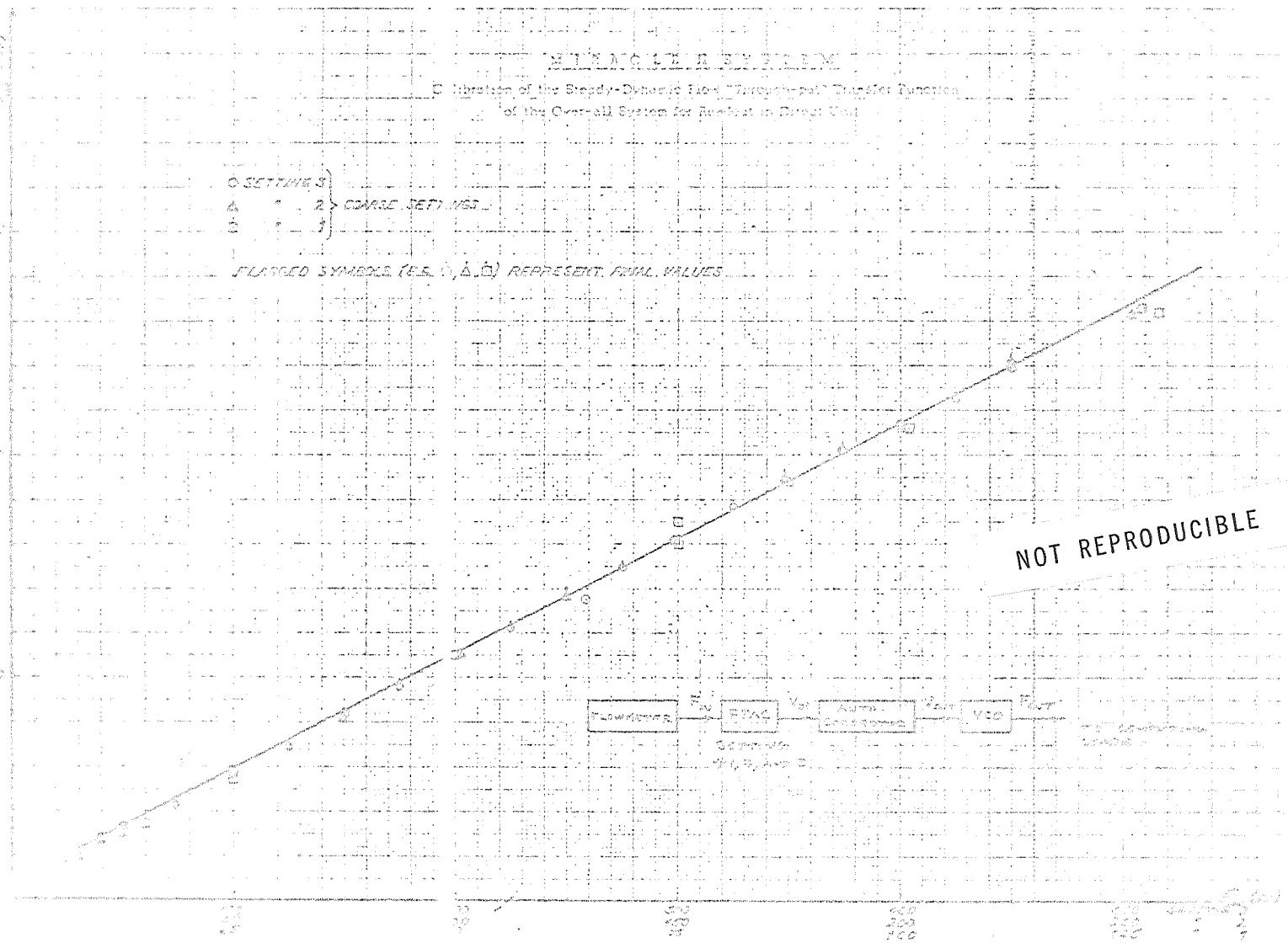
$$\frac{I\cos\vartheta_0}{r^2}\cdot\frac{dV}{dt} + \frac{K\cos\vartheta_0}{r^2}\cdot V = \rho a|u|(u\sin\vartheta_0 - v\cos\vartheta_0) \tag{7}$$

Setting $A = \frac{I\cos\vartheta_0}{r^2}$, $B = \frac{K\cos\vartheta_0}{r^2}$, $C = \rho a$,

we arrive at:

$$A\dot{V} + BV = C|u|(u\sin\vartheta_0 - v\cos\vartheta_0) \tag{8}$$

The above equation is used as an analytical expression of the dynamic motion of the turbine flowmeter. It is expressed in a form more conductive to the practical solution of the problem by means of electronic techniques.


MISSILE SYSTEM
Variation of the Steady-Dynamic Flow Throughput Transfer Function
of the Overall System for Reactant in Direct Use
○ SETTINGS
△ " 2
□ " 1
COARSE SETTINGS
FLAGGED SYMBOLS (E.G. ○, △, □) REPRESENT FINAL VALUES
FLOWMETER
EPMC
AUTO CONTROLS
VCD
NOT REPRODUCIBLE

[illegible] angle α_e approaches α when ω becomes constant; and [illegible] to a small and near stable quantity. For more direct and simpler [illegible] computation, we make the justified assumption that v, being [illegible] induced small quantity, is proportional to the indicated velocity.

The advantage of using the above equation is that, although it is [illegible], a non-linear differential equation, it is more amenable to electronic computer techniques; and, a built-in electronic means can be inserted to [illegible] a linear relationship between the input and output under [illegible] condition, namely, when u is constant. (see: Fig. 8A)

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1.2.3. TRANSIENT RESPONSE OF TURBINE FLOWMETER UNDER SUB-ATMOSPHERIC RESPIRATORY CONDITION

In sub-atmospheric (1/3 atmosphere) respiratory flow measurement, one deals with the astronauts' time-varying bi-directional flowrates in low density gaseous media. The dynamic response of the flowmeter under rarefied flow condition is, therefore, of considerable interest. In contrast, under higher density respiratory flow condition, such as in dealing with the breathing of aquanauts and divers, the equation of motion of the turbine flowmeter can be more simply expressed as follows:

$$\frac{d\omega}{dt} = A\,\Phi(\Phi - \omega)$$

where ω = indicated flowrate

Φ = actual flowrate

A = constant

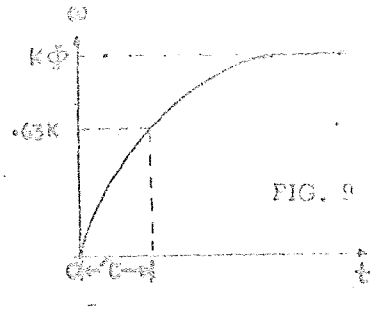


FIG. 9

Alternatively, but more pertinent, the equation can be derived in another form from consideration of the change in angular momentum of the fluid, i.e.

$$\tau\dot{\omega} + \omega = K\Phi$$

where $\tau = C/K\Phi$ the time constant of the sensing element.

and $\omega = K\Phi\,(1 - e^{-t/\tau})$ is the output frequency of the flowmeter.

The time constant τ of the turbine rotor is known to be influenced by several factors: the density of the fluid (temperature, pressure), the flowrate, the moment of inertia of the rotor, and the geometrical design of the flowmeter. In general, τ depends on:

1) Flowrate - the time constant decreases with increasing flowrate.

2) Fluid Density - the time constant decreases with increasing density.

* ω is the instantaneous angular velocity of frequency output f.

3) Rotor blade angle - but only in a relative sense, decreasing the helical angle of the blades does not invariably shorten the time constant as such, since the rotor speed is also reduced.

4) Moment of inertia: decreasing the moment of inertia of the rotor will shorten the time constant.

For small perturbations at moderately high flowrate, the response of the Quantomics-Liu flowmeter has proven experimentally to be sufficiently high. However, for a bi-directional pulsating flow, the situation is more complicated. Not only large perturbations are involved, but the flow changes direction. During the past, very little work has been done in this area owing to the complexity of the theoretical and experimantal problems.

In the following "worst case" study, the problem of transient response is examined in greater detail. In these analyses, the "worst-case" step function or square-wave driving function is used, although in actual respiration, no such abrupt rise or fall is involved, and the QL flowmeters have shown to be able to respond to all ordinary breathing cycles.

Consider the following two diagrams, both using a short inspiratory and expiratory period.

FIG. 11

In (a): As the step type driving function is applied at t=0, the rotor is building up its speed in response to the driving function but will not exceed the maximum. If the pulse is short, at B, the pulse reverses hence exerting a reversing torque on the turbine while the flowrate is at a high point. Because of this, we have a fast drop. The drop starts at C because of time delay or phase shift .

In (b), we assume that there is the characteristic overshootings after the rotor reaches its prescribed speed due to inertial [illegible] on its maximum m; it drops and eventually stabilizes at B'.

In terms of transient response, consider the following diagram:

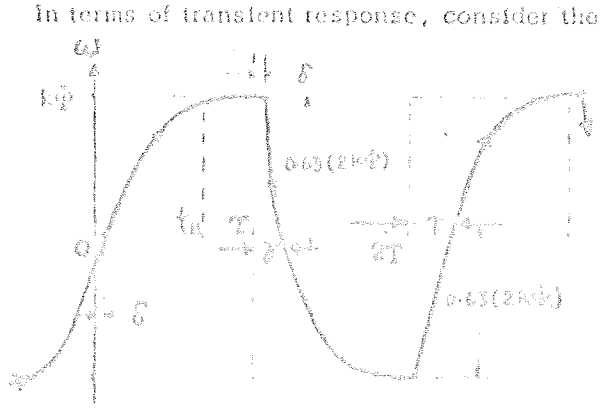


FIG. 12

δ = phase shift
γ = [illegible]
τ = [illegible]
$\tau = 1/2$ [illegible]

From $t = t_1$ to $t = T$, namely, reversal of pulse, the behavior of the turbine is analogous to that in steady state, i.e. $\omega = K\Phi$ or, $\Phi = \omega/K$; but outside this period, the following expressions are applicable:

$$\Phi = \frac{\omega}{K}\left[1 \pm \left(1 + \frac{\Delta\Phi}{\Delta\omega}\right)\right] \quad \text{or,} \quad \Phi = \frac{1}{K}\left(\tau\dot{\omega} + \omega\right)$$

After this point, the pulse reverses its direction, hence imposes a reverse/retarding torque on the turbine, i.e. we have a dropping time constant τ'.

If we assume that only one time constant is involved, then by using computer techniques, we can compensate the error and restore the square wave form. On the other hand, if both a rise-time constant τ and a dropping time constant are involved, the problem becomes more complex.

For the latter case, a simulator model is used; the actual electronic compensation is carried out by the Auto-Corrector. With this system, tests were made at various flowrates. The following recordings indicate both the uncorrected and the corrected signal compared with the original one.

	Input Signal	Uncompensated Signal	Compensated Signal
At 1 Atm.			

From these diagrams, it can be observed that the compensation is adequate, for ordinary breathing rates. Even at very low speed, the system still maintained good performance. The present uni-directional type auto-corrector has already proven to have reduced the flowmeter's time constants by a factor of 10. Breath-by-breath measurement at 1 atm. and

above is no longer a difficult problem.

The Auto-Corrector concept, particularly the bi-directional type auto-corrector, is even more essential for metabolic measurements at sub-atmospheric pressures, e.g. at 1/3 atmosphere or 5 PSIA. Low-pressure respiration experiments at Quantum Dynamics, using artificial lung (enclosed in evacuated bell jar) to provide the driving function, as well as the following comparative analysis, have confirmed such a necessity.

For a given turbine rotor configuration and moment of inertia, the time constant has been analyzed by Grey. The more complex equation is given in the following simplified form for comparative purposes:

$$\tau = \frac{K(1 + 2\eta/A)}{\rho V \eta}$$

where ρ = density

V = the flow velocity

A = the characteristic of a turbine blade

η = streamline coefficient of the blade ($\eta \sim 0.9$)

K = constant

Similarly, Vovchenko has derived an expression of time constant in another form by non-linear treatment of the turbine's behavior. Again, a simplified equation is given as follows:

$$\tau = \frac{C_1\left(\frac{C_2}{\rho} + \xi_1\right)}{\xi_2 \; V} \simeq \frac{C_1\left(\frac{C_2}{\rho} + \xi\right)}{\xi \; V} \simeq \frac{C_1(C_2 + \rho\xi)}{\rho V \xi}$$

where C_1 and C_2 are constants.

ξ_2 and ξ_1 are respectively the streamline coefficients of the flow distribution in the turbine channel, and the coefficient due to the added mass of fluid. In the case of rarefied gas flow, ξ_2 and ξ_1 are virtually

the same.

Both the above equations show that the time constant τ varies inversely with the density and the velocity of the gas flow; or, in other words, varies inversely with the mass flux. If one uses the flow velocity as the criterion: for the same velocity, the time constant at 1/3 atm. should be, theoretically speaking, three times longer. However, the criterion could be the mass flux instead of velocity. In this case, if one assumes that a certain amount of mass flow of oxygen is required physiologically at 1 atm. as well as 1/3 atm., then the relative ratio of the time constant may be different from the 1 to 3 ratio. Nevertheless, one should expect that the time constant would be affected, probably in a systematic manner, under subatmospheric breathing condition. In which case, one of the effects can be the distortion of the 'flowrate vs. time' waveform. Such distortion, however, can be removed by means of an efficient Auto-Corrector, as we have confirmed by means of the following experimental evidence. (Fig. 13)

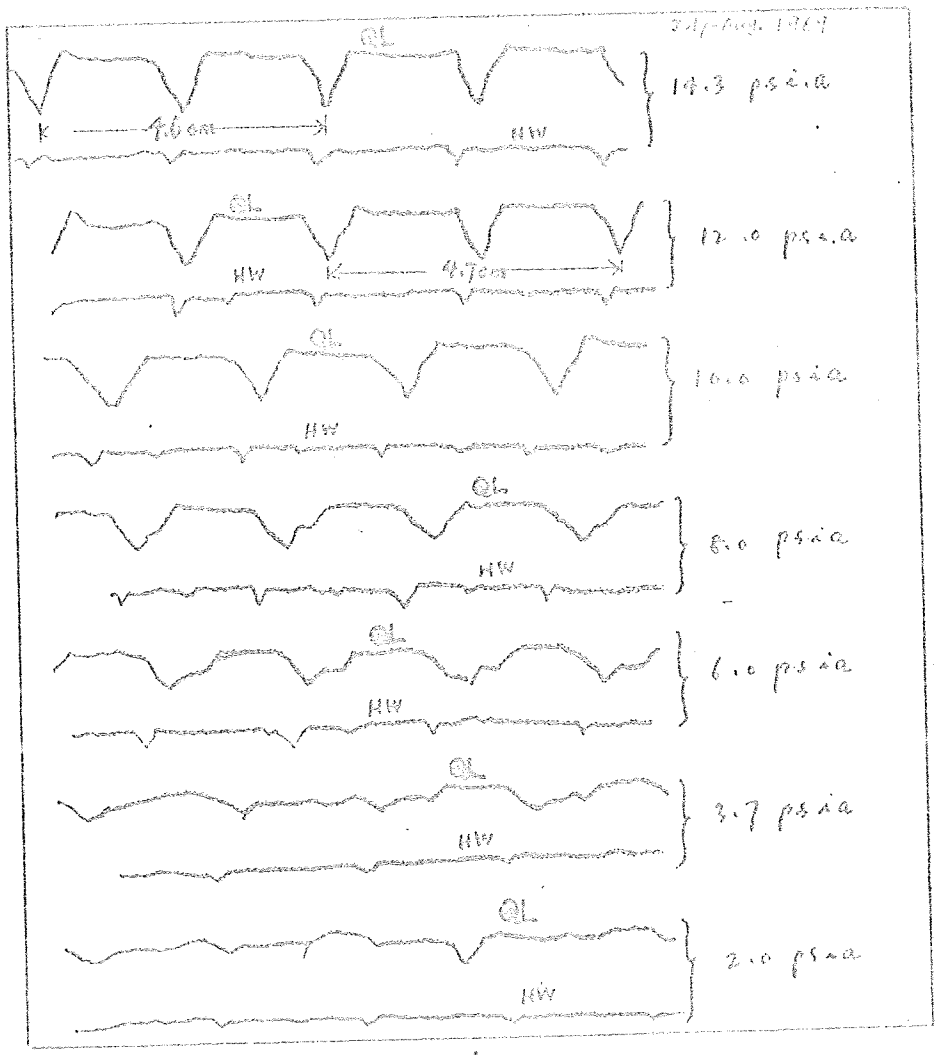
The oscillograms of "flowrate vs time" waveform were obtained earlier by hand tracing the data from memoscope recording, due to the lack of an oscilloscope camera. The Auto-Corrector used for such testing was of the very early version, now already rendered obsolete by our newer designs. This "Thermotron" type electronic corrector lacked a dividing term and was not theoretically rigorous enough for flowmeter use, being used primarily to solve a simpler thermal response equation. It was used in these tests only because it was the nearest available for such purpose, and would provide some proof as to the "correctability" of the subatmospheric density effects. In this regard, it served a useful purpose, but it's ability is far inferior even to the uni-directional type of flowmeter auto-corrector. With the Bi-directional Auto-Corrector of rigorous design, the problem of faithful measurement of the breath-by-breath flowrate and volume can be tackled much more effectively.

FIG. 13

ELECTRONIC CORRECTABILITY STUDY OF THE

SUB-ATMOSPHERIC

DENSITY EFFECTS ON DYNAMIC FLOW MEASUREMENT



Upper Traces (QL)	QD Flowmeter Output from FPAC Converter	5 volt/cm
Lower Traces (HW)	Constant Temperature hot-wire anemometer	1 volt/cm
Time: 1 Sec/cm	R.R. Breath Frequency	12.8 cycle/min
Reference Speed: 300		
Flowrate Generated By:	Q.D. Artificial Lung	

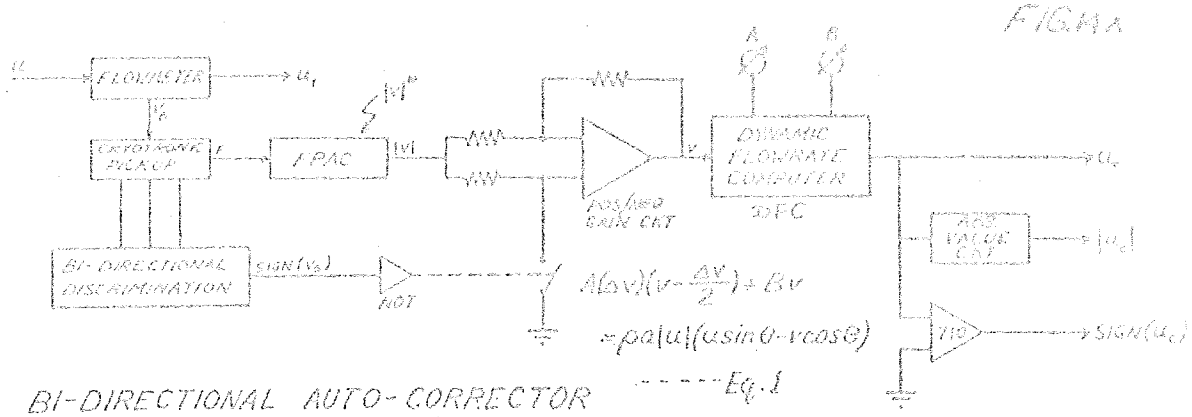
These preliminary experiments conducted on 21 August, 1969, using the artificial lung tester to provide a trapezoidal type flow-rate input, show that: after 6 PSIA pressure, we have a distorted trapezoidal type square-wave which degenerates rather rapidly after p = 6 PSIA. But, above that pressure, the wave-forms appear similar (even in amplitude) to those recorded under 1 atmosphere condition.

After the Thermotron "non-rigorous" Auto-Corrector is applied, there is a major improvement in the wave-form, again down to 6 PSIA, after which the correction remains effective. Even at 2.7 PSIA, the use of Thermotron is still an advantage.

The flowmeter continued to yield large signals even at 2 PSIA. In comparison, however, the hot-wire anemometer and two other gas flow sensors tested under the same condition have ceased to provide any useful signal, after the pressure is below 10 PSIA.

Again, the Thermotron compensator used in these preliminary studies was not regarded as sufficiently rigorous: the use of more rigorous and more efficient Bi-Directional type Auto-Corrector will certainly improve the result by a wide margin.

FIG. 4A



BI-DIRECTIONAL AUTO-CORRECTOR
(BLOCK DIAGRAM)

FIG. 4B

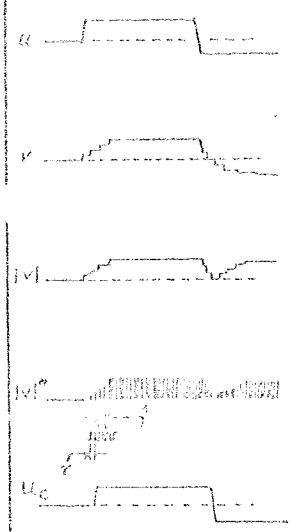
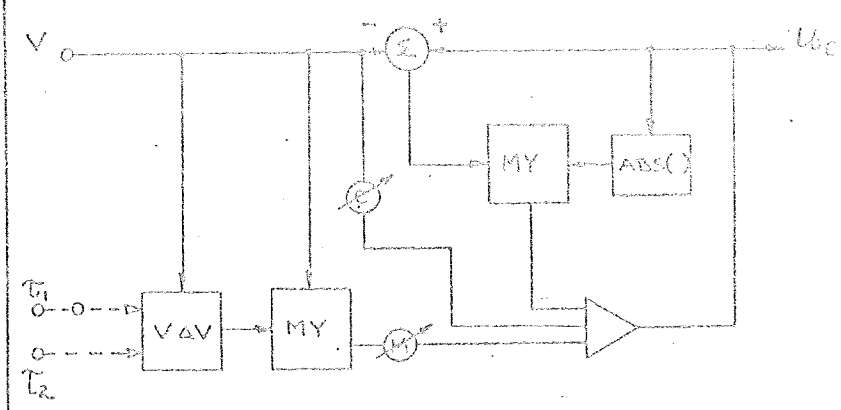


FIG. 4C



ELECTRONIC FLOWMETER SIMULATOR

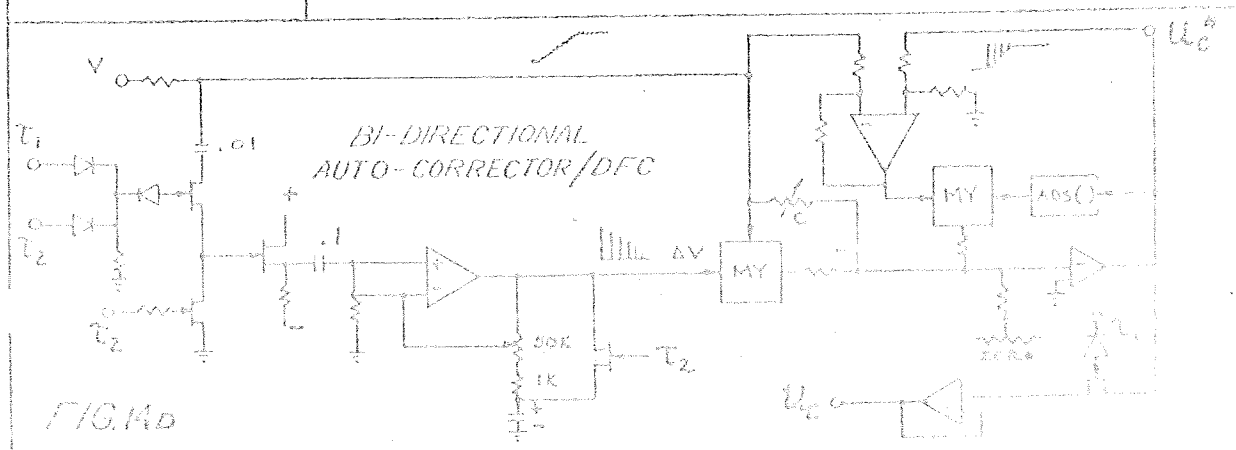


FIG. 4D

1.2.4. BI-DIRECTIONAL AUTO-CORRECTOR (Fig. 14A,B,C,D.)

The true flowrate has been computed using the "uni-directional" model in a feedback loop. The solution is not the most rigorous although mathematically adequate; since the FPAC Transient Flow Indicator's output is not strictly an "all-time" blade velocity in the continuous sense, but a sampled blade velocity. More specifically, with the FPAC output in the form of ascending or descending ladder-step type signal, the differential in the computational process can only be obtained after a number of blade counts. Although the uni-directional system has improved response about 10 times; it has not reached its optimal (or maximum) response. Maximum response is achieved when a calculation is performed for each time the blade passes the sensor. With the uni-directional Auto-Corrector, if higher response is attempted with the uni-directional model, sampling noise is amplified to an objectionable degree. The noise due to the electronic differentiation process is particularly objectionable at the polarity-changing instant; since instead of smoothly ascending or descending, we have to reverse the signal so that both inspiratory and expiratory flow are of the same positive polarity. While at high flowrate and longer breath period, the computing error is shown to be very small; at low flowrate and short period, the accuracy of computation suffers from deterioration which can be eliminated by means of the Bi-directional Auto-Corrector approach.

In the Bi-directional Auto-corrector, the computer solves equation (1) once for each time the blade passes the sensor. Example: if the flowmeter frequency output is 500 Hz, then response is 500 calculations per second, instead of, say 250 or even 125 differential equation calculations per second.

This is particularly important for low respiratory flowrates when the output frequency is lower, and the period is longer. A block diagram of the Bi-directional Auto-Corrector is shown in Fig. 14 . Modification of Equation (1) to more rigorous Equation (2) in discrete form is also presented, from which the new Auto-Corrector is designed.

$$A(\Delta V)(V-\Delta V/2) + BV = a|u|(u \sin\Theta - v \cos\Theta) \qquad \text{Eq. 1}$$

One of the important advantages of the Bi-Directional Auto-Corrector is the elimination of the substantial noises due to differentiation near the zero level when the inspiratory flow change-over into expiratory flow. As in the natural breathing process, the Bi-Directional Auto-Corrector preserves a smooth change of flowrate signal by means of the electronic change of signal polarity. The accuracy of computing true florate by more rigorous and high-density computation process, and the reduction of noises, together with more powerful digital and hybrid computor techniques used in the Bi-Directional approach, will result in considerably more faithful measurement of breath-by-breath respiratory and metabolic quantities.

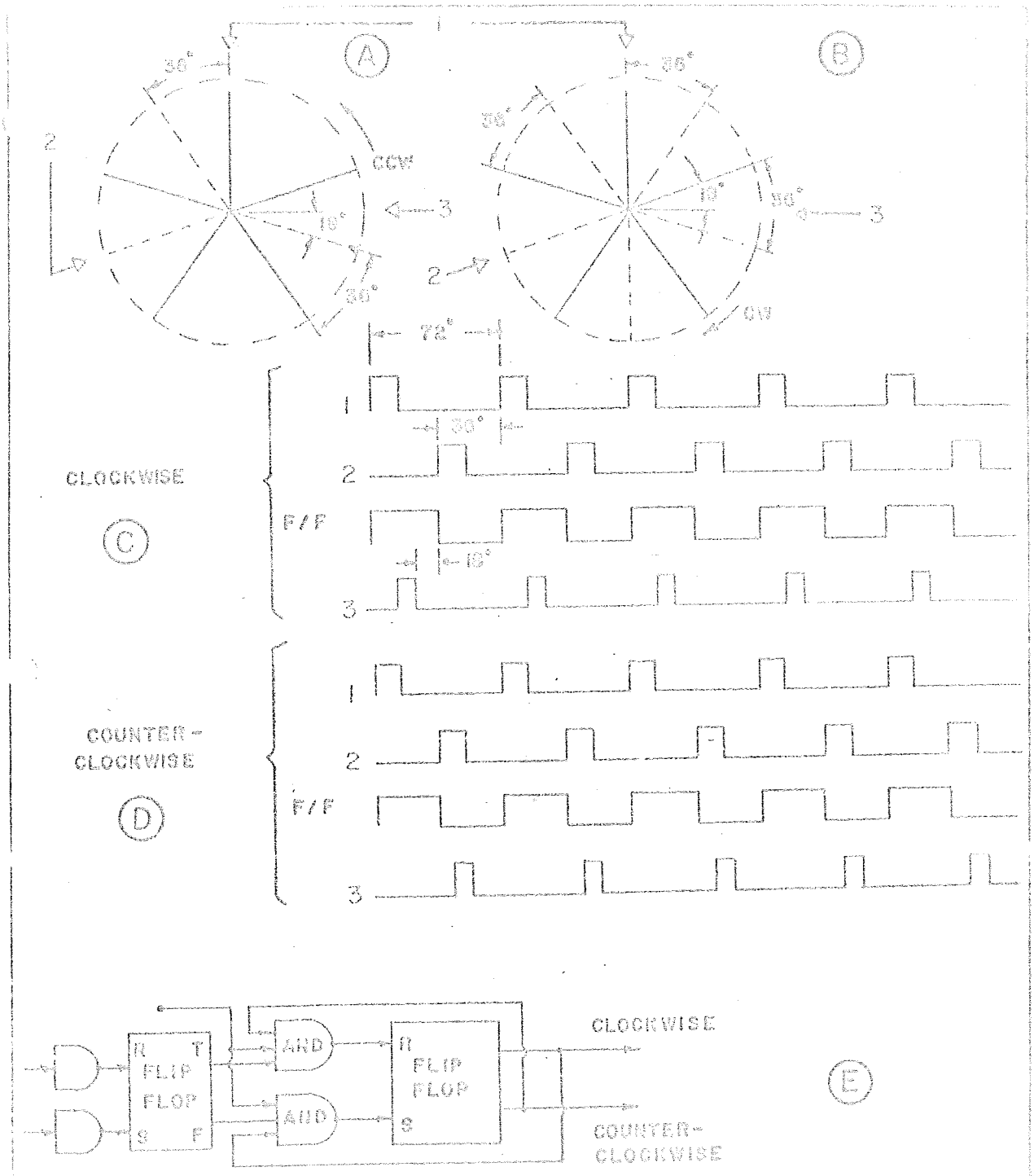


FIG. 15

BI-DIRECTIONAL DISCRIMINATOR

NOT REPRODUCIBLE

1.3. BI-DIRECTIONAL DISCRIMINATOR

The Flow Direction Discriminator has the function of providing positive identification of the direction of the respiratory flow, as to whether the flow is due to inspiration or expiration at any given instant. The following discusses the design regarding to a 5-blade turbine.

Three pickup heads are used with the five blade turbine bi-directional flowmeter. The present flowmeter is a 3/4 or 1-inch unit. Pickup No.1 and No.3 are displaced 90° apart with respect to each other, while Pickup No. 2 is displaced 108° from No. 1. The pickup placement is illustrated in Figs.15A and B. The design for the n-blade unit is similar.

When the turbine is rotating, a pulse is generated at each pickup head each time a blade passes under the pickup. Because of the mechanical skew of each turbine blade, the pulse generated as the blade passes beneath the pickup is too long in duration. To improve this situation, a pulse shaper is used to generate a pulse of 50 to 200 NS in duration each time the leading edge of a turbine blade passes under the pickup. The pulse trains generated by each of the 3 pickups are illustrated in Figs.15 C and D for CW and CCW rotation respectively.

The pulses generated by Pickup No. 1 are used to set a bi-stable RS flip-flop while the pulses generated by Pickup No. 2 are used to reset this RS flip-flop. With this arrangement, the RS flip-flop is enabled for 36° of every 72° of physical rotation for each blade that passes under pickup No. 1. This action occurs regardless of the direction in which the turbine rotates. The pulses generated by Pickup No. 3, however, are developed at different times with respect to the RS flip-flop enabled time

for CW and CCW rotation.

If the turbine blades are moving in a clockwise direction, the pulse generated at Pickup No. 3 will occur after 18° of mechanical movement. At this time, the RS flip-flop is enabled. A three input NAND gate is used to detect the condition. The output of the NAND gate will go low (O VDC) causing the output of the RS flip to go high (+ 4.5 VDC) indicating CW rotation to the remainder of the system. This condition will only occur when the RS flip-flop is enabled, the pulse is present from Pickup No. 3 and the output RS flip-flop was previously in the CCW state. Therefore, once the RS output flip-flop changes to the CW state, it will remain so unless the turbine changes its direction of rotation.

When the turbine changes direction to CCW, the pulses generated by Pickup No.3 will occur when the input RS flip-flop is not enabled. This condition occurs because the turbine blade passing under Pickup No. 3 must travel 54° while the blade passing under Pickup No. 1 has also passed under No. 2 resetting the RS Input flip-flop. This condition is detected by another 3 input NAND gate which then resets the output flip-flop to the CCW state. This condition will only occur when the input RS flip-flop is disenabled, the pulse is present from Pickup No.3 and the output flip-flop is in the CW state.

It is easily seen from Fig.15C and Fig.15D that the pulses from Pickups No. 1 and No.2 are used to generate a time interval during which time Pickup No.3 is monitored. If the pulses from No.3 occur during this interval, the direction must be CW and if they occur after the interval, the rotation must be CCW. This technique will work regardless of the speed of rotation.

The RS flip-flops and gates are all implemented with 2,3 and 4 input DTL NAND gates. The 4 input gates are used as inverters. These integrated circuits are packaged in 14-pin dual in-line plastic substrates. The typical switching times of these devices are 40 NS, which allows the remainder of the system to sense direction changes almost instantaneously with respect to the relatively slow mechanical rotation of the turbine flowmeter.

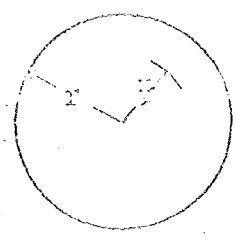
NOT REPRODUCIBLE

"WORST CASE" ATMOSPHERE CALCULATION OF THE TRANSIENT PRESSURE DROP & FLOW RESISTANCE CALCULATION OF 3/4" QUANTUM DYNAMICS BI-DIRECTIONAL RESPIRATORY FLOWMETER

The following analysis and calculations were performed for earlier model of flowmeter, and based on the following assumptions:

1. That the total ΔP is sufficiently small so that the density, or specific weight, of flowing gas, does not change with position along the flow stream;
2. Although the flowmeter uses a 22.5° blade angle in consideration of transient response advantages; 30° straight, but not helical blade angle is used for worst-case analysis: Assume that the pressure drop under locked-turbine condition is the maximum ΔP that could ever be experienced;
3. Assume that the flow process between the stalled rotors is that of a diffuser with no pressure recovery; and that the static ΔP across turbine rotors constitutes the ΔP loss.
4. Assume blade chord is about 5 mm.
5. Number of blades is 5 which applies to the earlier model flowmeters. In the high resolution type flowmeter, the turbine has 9 blades.

The Flowmeter's Analytical Dimensional Configurations



$\bar{r}$ = Area Averaged Radius

In the following analysis, an area average circumference is used, namely, $2\pi\bar{r}$, such that:

$$\text{Flow Area} = A = r^2\pi = \bar{r}\ 2\pi r$$

$$\frac{r^2\pi}{r} = \frac{2\pi r\ \bar{r}}{r}$$

$$r = 2\bar{r} \qquad \bar{r} = \frac{r}{2}$$

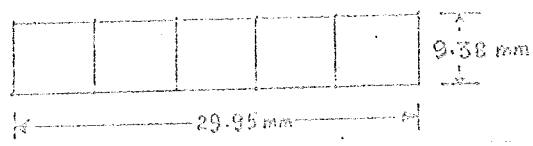
An equivalent linear cascade of blades would have a length dimension of $r(2\pi)(\frac{1}{2})$, where $(2\pi)(\frac{r}{2})$ is the area average circumference.

Area average circumference $= \pi r$

For I.D. = 3/4" = .75"; $r = 3/8 = .375 = \frac{.375}{.040}$ mm = 9.38 mm

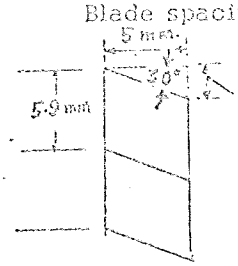
$$\pi r = \pi(9.38) = 29.45 \text{ mm}$$

Equivalent Cascade



For 5-blade turbines:

$$\text{Blade spacing} = \frac{29.45 \text{ mm}}{5} = 5.9 \text{ mm}$$



Blocked circumference/blade = B

$$\frac{B}{5} = \text{Tan } 30^{\circ} = .5775$$

$$B = 2.89$$

The through flow area per blade = (9.38) (5.9 - 2.89)

$$\cong (9.38)(3.0) = 28.15 \text{ mm}^2$$

Total through flow area is given by:

$$5\,(28.15) = 140.6 \text{ mm}^2 \quad \text{at blocked plane.}$$

Inlet Plane area $= \pi r^2 = (9.38)^2\ (\pi) = 276 \text{ mm}^2$

Work Case Calculations Based on 1 Atm. and Liter/min. Air Breathing Consumption

First, an analysis is made using air as the effluent. The composition of air in percent by volume is given by Keenan and Kayes as follows:

N_2	78.03%
O_2	20.97
A	0.98

a) The partial pressures are calculated as follows:

$$p\ O_2 = .2099\ (14.7) = 3.085\ \text{PSI}$$

b) The maximum nominal pressure condition is:

14.700 - Sea Level Pressure

3.085 - Suit Pressure

17.785 PSIA

c) The percentage of $pO_2 = \frac{3.085}{17.785} = .1735$

d) The molecular weight of $O_2 = 32.00$

e) The molecular weight of $N_2 = 28.02$.

f) The following equation is used for the calculation:

$$\text{Mols } O_2 \times \text{Molecular Weight of } O_2 = \text{Weight of } O_2$$

For 1 Mol of Air, Weight of $O_2 = 0.1785\ (32) + 0.8265\ (28.02)$

$$= 5.55 + 23.16 = 28.70$$

The molecular weight of Gas at 17.78 PSIA is given by:

$$M = 28.70$$

with $\bar{R}$ = Universal Gas Constant = 1545.3 Ft. Lb.°R / Lb. Mol°R

$$\frac{\bar{R}}{M} = \frac{1545.3\ \text{Ft. Lb.}^{\circ}\text{R}}{(28.70)\ \text{Lb. Mol}^{\circ}\text{R}} \quad \frac{\text{Lb. Mol}}{\text{Lb. M}} = 53.88\ \frac{\text{ft. Lb}^{\circ}\text{R}}{\text{Lb. M}^{\circ}\text{R}}$$

The Weight Fraction of O_2 is:

$$\frac{5.56}{28.70} = .1938$$

since $P\,v = R\,T$

$$(17.78)\,(144) = 53.88\,(558)$$

$$= \frac{53.88\,(558}{17.78\,(144)}\ \frac{Ft.^3}{LBM}$$

$$= .021\,(558) = 11.7\ \frac{Ft^3}{LBM}$$

The Bernoulli equation applies to the flow across turbine:

$$P_1 v_1 + C_{P_1} T_1 + \frac{V_1^2}{2g} = P_2 v_2 + C_{P_2} T_2 + \frac{V_2^2}{2g}$$

The following thermodynamic properties of gas mixtures are used for turbine ΔP calculation:

At ordinary room temp.:

O_2 Cp = .219 BTU/LBM^{o}F Cv = .156 $\frac{BTU}{LBM^oF}$

N_2 Cp = .248 BTU/LBM^{o}F Cv = .177 $\frac{BTU}{LBM^oF}$

Using the following conversion factors for calculation:

30.5 Centimeter - 1 Foot

3.05 mm = 1 Ft.

2.20 Pounds - 1 Kilogram

1 Litre = 1000 CM^3

The following calculations are based on the maximum 350 liter/min. flowrate.

Given the inlet area (=276 mm^2) , the velocity at inlet area is calculated as:

$$\frac{350\ (1000)\ CM^3\ Min.}{Min.\ (60\ Sec)\ (2.76)} = 21.15\ CM/Sec.$$

$$= 2.115\ m/Sec.$$

$$= 7.06\ Ft/Sec.$$

If $Cp_1 T_1 = Cp_2 T_2$; on a unit mass flow basis,

$$P_1 v_1 + \frac{V_1^2}{2g} = P_2 v_2 + \frac{V_2^2}{2g}$$

The velocity at reduced area is calculated as:

$$\frac{350\ (1000)}{(60)\ (1.400)} = 4.15\ CM/Sec.$$

$$= 4.15\ Meter/Sec.$$

$$= 13.83\ Ft./Sec.$$

$$\frac{(V\ )^2}{2g} = \frac{(7.06)^2}{2\ (32.2)} = 0.775$$

$$\frac{(V_2)^2}{2g} = \frac{(13.83)^2}{2\ (32.2)} = 2.97$$

Assuming at small ΔP, the specific volumes $v_1 = v_2 = 11.71\ Ft^3/LBM$

$$P_1 - P_2 = \left(\frac{V_2^2 - V_1}{2g}\right) \frac{1}{v} = \frac{2.195}{11.71} = .1875\ Lb/Ft^2$$

$$= \left(\frac{0.1875}{62.4}\right)(12) = .0361\ in\ H_2O$$

For the Piggie-Back Two-turbine configuration, the Flow Process occurring between the turbine rotors is analyzed as follows:

Mass Flow $\dot{m}$ Calculation based on 350 Liter/Min. flowrate.

$$\dot{m} = \frac{350\ (1000)\ \cancel{CM^3}\ (60)\ \cancel{Min.}\ (1\ Ft^3)\ LBM}{(Min.\ \cancel{X}\ 1000)\ \cancel{CM^3}\ (1728\ \cancel{in^3})(11.71\ \ Ft^3)}$$

$$= \frac{350\ (60)\ (1)}{(1728)\ (11.71)} = 1.055\ LBM/Min. = 0.0176\ LBM/Sec.$$

The worse case-loss occured under 1 atmosphere air flow condition across two turbine rotors is calculated as:

$$\text{Total} \quad \Delta P = 2X.0361 = .0722 \text{ In } H_2O$$

The total ΔP calculated for 1/3 atm. condition is given by:

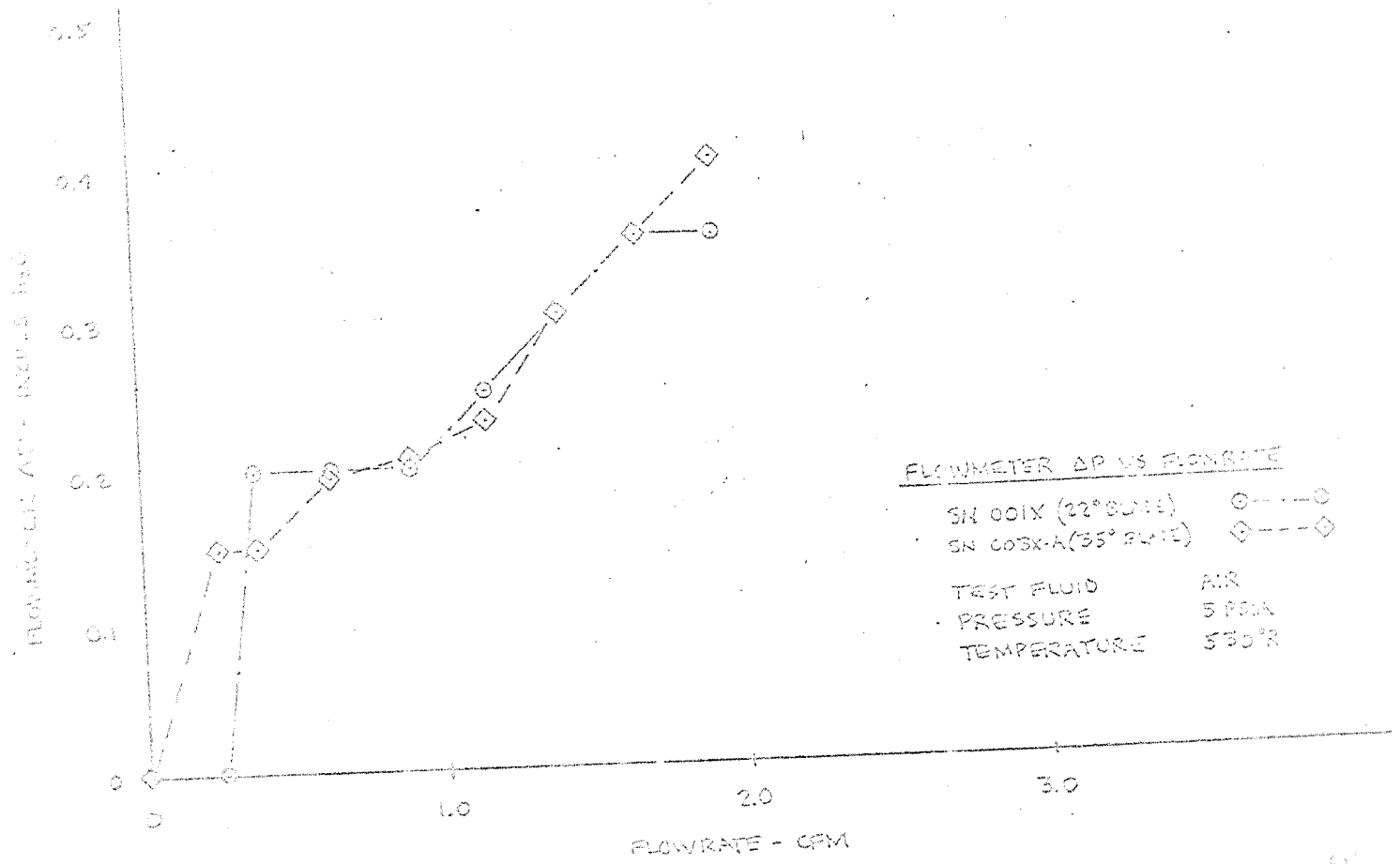
$$\Delta P \text{ (at 1/3 atm)} = .024 \text{ In. of } H_2O$$

It must be pointed out that when the flowmeter is in operation, the turbines are rotating freely in response to the respiratory flow; the actual ΔP is therefore lower than that shown in the illustrative locked-blade or stalled rotor condition. Thus, the Bi-directional respiratory flowmeter is shown to impose an exceptionally low flow resistance and pressure drop to the user during his breathing process.

However, the analysis shown above dealt only with the purely fluid dynamic aspect of the pressure drop, since it is more amenable to analytical treatment. It does not include the losses induced by other factors; such as the bi-directional reacting torques, due to the turbine moment of inertia, bearing friction, etc. For highly sensitive respiratory flowmeters, these factors will increase the presently calculated pressure drop values.

In view of analytical complexity of treating these factors theoretically, we have made some minimal experimental investigations on the magnitude of the over-all pressure drop, particularly under the 1/3 atm. flow condition. Our effort was of a limited scale and covered the 5 to 56.6 liter/min range, due to the fact that: a) the present contract is too small and inadequately funded to permit a more extensive investigation; b) the actual differential pressure is so small that the measuring instrument available to us is far too insensitive for accurate investigation. Nevertheless, the enclosed

lated "Flowmeter ΔP vs volumetric flowrate" at 1/3 atm., and covering the limited range of 5 to 56.6 liter/min., shows the low pressure drop of the flowmeters. Two experimental types of flowmeters were used in the tests. The test data shows that the pressure drop is higher than the calculated pressure drop values due to purely fluid dynamic origin. However, unlike the great precision with which flow measurement can be conducted at Quantum Dynamics; the experimental ΔP measurement using differential gauges far inadequate for the purpose, was subject to fairly large error and could only be relied on to provide an order of magnitude estimate.


FLOWMETER ΔP VS FLOWRATE
SN 001X (22° BUTT)
SN 003X-A (35° BUTT)
TEST FLUID AIR
PRESSURE 5 PSIA
TEMPERATURE 530°R
0.4
0.3
0.2
0.1
0
0
1.0
2.0
3.0
FLOWRATE - CFM

III. METABOLIC COMPUTERS

1.0 Special Purpose Electronic Computers

In the MIRACLE II System, the following Quantum Dynamics/Electronics special purpose computers are used:

Computers	Present Band Width	Accuracy	No. Used	Type
A. PF/T Mass Flow Computer and Multiplier-Divider	0 to 500 Hz for "T" input 0 to 1500 Hz	± 0.1% ± 0.5%	8	Hybrid
B. Model MY Analog Multiplier-Divider	.10 Mega Hz	± 1.0%	2	Analog
C. FPAC Transient Flow Indicator Non-Linear Period-to-Frequency Comp.	0 to 500 0 to 1500	± 0.25% ± 0.5%	1	Hybrid-Analog
D. Non-Simultaneous Uptake/Release Digital Storage Adder-Subtractor Computer	0 - 1 Mz	± 0.1%		Digital
E. BCD Digital Divider for R.Q. Computation & Uptake, Release T_M Division.	0 - 2 KHz	± 0.1%	up to 5	Digital
F. Computers, Counters Function Generator, etc. of other makes.				

Of the above computers, A and B shoulder the most critical functions in real-time metabolic computation. Eight (8) computers of the former types, and one to two of the latter type, are used in each system. The FPAC computer occupies the central role in the precise measurement of dynamic respiratory flowrate.

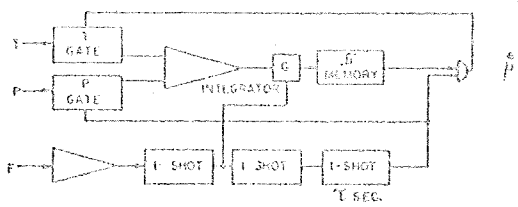
Detail description of the computer designs is not considered necessary here. However, the function of the PF/T computer will be briefly discussed.

1.) PF/T Computer

The PF/T Computer is a small, hybrid computer which accepts an alternating voltage of frequency F and two voltages P and T, and provides output pulses, the average frequency ($\dot{p}$) of which is determined by the equation,

$$\dot{p} = \frac{PF}{T}$$

The actual operation of the system is such that the output pulses occur at the same rate as the input frequency F, except that selected pulses are removed so as to satisfy the above equation. A block diagram of the PF/T computer is shown below:



The P gate, allowing P to pass, is opened for τ seconds once during each cycle of the F input, thus allowing an average voltage $PF\tau$ to appear after the gate. The T gate, allowing T to pass, opens for τ seconds every time a pulse P occurs; thus, the average voltage appearing after the T gate is $T\dot{p}\tau$. These average potentials are applied to a difference-integrator, which then provides a continuous average of the difference of the two gated outputs. The integrator output is gated once during each cycle of F; this output operates a memory amplifier that detects and memorizes whether

the integrator output was greater or less than zero. This memory output controls a coincidence gate, which allows pulses to get through to the T gate and the output. The operation is such that the verage integrator input is zero. Therefore,

$$PF\tau = T\hat{p}\tau$$

and

$$\hat{p} = \frac{PF}{T}$$

2.0 RESPIRATORY DATA DELAY CONTROLLER

2.0.1 General

The Respiratory Data Delay Controller is designed to correct the time "mismatch" situation between the dynamic respiratory flow signal and the partial pressure output signals from the mass spectrometer, which suffer from long time delay. For, if the flowrate signal is not computed cycle-by-cycle with its corresponding partial pressure signals of the same breathing cycle; it is obvious that considerable computational error would result, even if all the signals are accurate by themselves. The Data Delay Controller is designed to correct this situation so that the metabolic "end" data can be obtained more dependably.

The Controller accepts two channels of input data: a) Respiratory flow-meter pulse train output of up to 1.5 KHz on one channel; b) the second channel accepts a true or false level input representing inspiring/expiring breathing cycle. Each input has a corresponding output that is delayed by the same amount of delay time as selected on the front panel.

2.0.2 Operating Principle

The Data Delay Controller utilizes the delay characteristics of a 1536-bit MOSTEK Static Shift Register in series arrangement. Twelve (12) sections of MK 1002 dual 128-bit P-channel MOS static shift register are used to achieve a total delay time of up to 600 ms. Each of the two independent 128-bit sections have a built-in clock generator to generate three internal clock phases from a single-phase external (clock) input. In addition, each section has input logic for loading or recirculating data within the register,

The positive-logic Boolean expression describing this action is shown below (see functional diagram for definition of terms).

$$\text{OUT (delayed)} = (RC)\ (DIN) + (\overline{RC})\ (RIN)$$

2.0.3 System Description

The input data pulses are temporarily stored or held in a master-slave flip-flop where they are synchronized with the internal free-running clock which operates at 2560 Hz.

Once synchronized with the memory clock, the input levels are interfaced with, and entered into, a 1536-bit serial static shift register via a TTL inverter. For each 128 bits there is a convenient output that may be selected by the front panel control switch. The register data is then interfaced and the output is DTL/TTL.

2.0.4 Delay Calculations

Maximum Delay Time = 1536 (Bit Shift Register) x Period of 2560 Hz Clock $= 1536 \times \frac{1}{2560} = 600$ millisecond.

REPORT ON COMPUTING ACCURACY

(In Answer to Letter from Dr. John A. Rummel, 11-6-70)

(REF: EDG-210)

We are grateful that just prior to our submittal of the Final Report, we have received your preliminary testing results on the computing accuracy of the MIRACLE II System's Metabolic Simulator, computing and data section. Your report has again reinforced our conclusion regarding the necessity for the proposed improvement efforts shown in this Final Report. The following points are specifically in answer to your test results:

1. The test approach used in these tests appears to be correct, although for the low flowrate/high RR tests, the sensitivity setting of the FPAC Computer should be at Position #1 for flowrate range between 0 to 100 Liters/Min. At this setting, the FPAC provides larger signal for low flowrate outputs and more accurate reading.
2. The evidence is clear that at low flowrate/high RR testing condition, in particular at RR=60 breaths per minute, the inconsistency was largely attributed to: a) when flowrate frequency is small; the electronic gate operating at RR frequency may chop-off $\pm$ 1 count (of, say a total, 20-counts in the breath) each time. This $\pm$ 1 count is compounded 60 times per minute, resulting in significant discrepancy; b) the larger near zero level differentiating "noise" artificially introduced by the uni-directional Auto-Corrector. This has been explained in detail in several sections of the present report. We believe that such noise can be drastically reduced, or eliminated by the use of the Bi-directional Auto-Corrector. These discrepancies were generated artificially by the uni-polar Auto-Corrector in un-natural polarity reversing process.

3. Regarding the occasionally irratic print-out of ~ 9945 type wrong data; as reported in several sections of this report, we were plagued from the beginning by the 4-digit limitation of the counter. As a desperate measure, we introduced the $\pm$ indicating logic. But this added feature occasionally misbehaves. This misbehavior again convinced us of the correctness of our conclusion: that we would very much like to revamp the entire data system which was handed down from the uni-directional contract. The present system is too jammed in space, and so repeatedly modified that we were frustrated in making-do with it.

4. In the accompanying analyses of computing accuracy, where we compared the print-out results with the carefully calculated results we have proven that, when this unit is operating within its designed regime, the results are extremely accurate - with the exception of the R.Q.. However, our data also corroborate your finding that the accuracy deteriorates under low flowrate/high RR Condition. This situation indicates that the following steps must be taken:

   a) In order to improve the computing accuracy, we must resort to high-density computation and consequently this requires improvement of the FPAC and PF/T computers. At low computing density, the error due to the transient rising and falling stage of each breath becomes more pronounced. However, at present, we are only using approximately 1/5 of our measurement resolution and probably about 1/8 of our computing density.

   b) Within its designed range, the PF/T (hybrid analog-digital) computer is extremely accurate and in fact proven to be less

than 0.1%. However, when 6 or 8 of such computers are used together, each should theoretically require a stabilized power supply. We have used a common power supply for all the computers in the beginning. Later, when we discovered that large error could creep from one computer to the other through the power supply, we resorted to the decoupling of the supply voltage which helped the situation somewhat. There were no funds, or space in the old data system which would permit us to do anything more. We are fully aware that during the transient stage, crept-in transient noises interact one computer against the others and can cause errors in computation - occasionally rendering the computation totally erratic. We know that if we use a stabilized power supply, stabilized analog components, individual shielding and isolation, plus the use of a superior $\pm$ sign indicating logic; or, remove entirely such logic by using a 5-digit counter; then we can control the situation quite well.

c) In the present MIRACLE II System, a voltage-control-oscillator (VCO) designed by ourselves was hurriedly used. Although we have found it is accurate and linear under steady-state condition, this VCO was not of the highly stabilized design; we found that the present Auto-Corrector can be momentarily knocked off its zero level by the aforementioned auto-corrector polarity reversing differentiating negative-going pulses. Moreover, the present VCO may not have high transient response to rapid variation of input frequency. Given time, this problem can be easily solved.

We can use a highly (e.g. chopper) stabilized operating amplifier in the VCO and there are commercially available VCO which we can buy for such tasks. Most important, the use of the bi-directional Auto-Corrector would additionally relieve this problem.

d) The original breadboard version of the low-cost developmental type Data System, originally designed for low frequency uni-directional system application, has been so repeatedly modified that we have lost track with what changes have been made. It was not even made in printed-circuit form or adequately shielded. Since we have previously designed this unit, we now know exactly how to construct a more efficient unit.

5. We have previously reported repeatedly that the continuous R.Q. computer was not to our liking. It was demanded by Mr. Schlotman, partly based on his concept given to us around December, 1968. It is so complex, and the old-fashioned Raytheon D-to-A converter board which we bought, frequently misbehaved. To remedy the situation; we have developed the all-digital R.Q. computer in breadboard form, and are ready to implement in the next phase of work, if needed. We do not believe the use of another computer (such as the PDP-8/I available at MSC) would solve the situation effectively, as simply and economically as we could do it.

In conclusion, there are obviously minor weaknesses in the present computing system, particularly for "extreme case" application beyond its original designed range, such as 60-breaths per minute flowrate. But the system concept is generally sound and we know precisely where the area of improvement should be. These areas have been adequately discussed in many sections

of the present report, and we feel there should be no concern on the part of NASA-MSC. Given a minimal additional support, accurate metabolic computation would not be any serious problem. The technology is ready with us, and only requires the time and support to implement it in an economical manner. We also believe that the overall system concept is sound and efficient. Being a hand-me-down system from the uni-directional system, later patched up with numerous additional features, it should be treated as a breadboard unit, initially designed to prove the feasibility of full scale metabolic computation. With effective design largely available with us, we believe that the electronic unit can be made in more ship-shape and more compact form. In fact, the essential part of the system can be made so compact and reliable that it can certainly satisfy NASA's advanced space-flight requirement more so than any other system we have seen thus far.

We are most grateful to Dr. Rummel for the test data. These data greatly assisted us in directing our effort during the next phase.

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SUMMARY AND CONCLUSION
THE COMPUTING ACCURACY OF
MIRACLE II SYSTEM

The following test and calibration results of the computing accuracy of the Miracle II System were performed in the spring of 1970. A large number of tests have been conducted and it was proven that it is feasible to achieve accurate metabolic computation on a real-time basis through on-line electronic computation technique such as those used with our Miracle II System. However, these points must be brought out:

1. The voluminous data contained herein are typical of the computing efficiency and were randomly selected in order to evaluate the capability of the Miracle II System, when operated within its initially designed range. The data show that with the uptake and release computations, and at normal respiratory frequencies with breathing period of longer than 2.5 seconds, the computing accuracy ranges from ± 0.1 to ± 0.[illegible], with with ± 0.6% maximum. Only in the worst case where the breathing period was one second or shorter, is the probable error shown by the difference between the printed out values and the calculated values, reaching ± 1.5%. These results were considered extraordinary for metabolic measurements. When the ± sign indicating logic misbehaves, the output should be subtracted from 10,000. In these cases, data are not reliable.

2. The present tests were conducted with relatively low-resolution type, and in many cases, largely unstabilized computers of an earlier design. The computers in the Miracle I and II Systems were, according to the early requirements from NASA-MSC, and from its Metabolic Programs, not required to perform computation at accuracies higher than ± 3%, although we have designs at Quantum Dynamics which can insure much higher accuracy using an improved electronic computing unit. The system was basically designed for use with a lower-resolution uni-directional type respiratory flowmeter. There was no intention then to tackle the high repetition frequency breath-by-breath respiration.

3. The source of error may be attributed to the present inadequacy of the PT/T computers to measure the rising and falling portion of the breath-wave. The artificially induced "differentiation noise" generated by the uni-directional auto-corrector also contributed to the error of computation. At normal flowrate and low RR, this error becomes relatively small. We already have a design that can largely eliminate such errors. Another probable source of error can be the low-end non-linearity of the PT/T type Transient Flow Computer. A proposal has already been submitted to improve these computers for more accurate on-line computation at higher pulse density.

4. The data proves that the real-time computing techniques and computers as shown in our design of the Miracle I and II Systems are capable of obtaining precise data for all the presently required metabolic [illegible] on-the-spot, and only modest development is required in order to extend its full efficiency for space flight or ground application.

* Note: In all these tests, both the inspiratory and expiratory values were printed-out for analysis.

MASS SPECTROMETER SIMULATOR OUTPUT

SETTING & CALIBRATION

VOLT INPUT			PERCENTAGE (Volt/)x100	
INSP.	EXP.		INSP.	EXP.
1.000	.299	O_2	19.00%	6.03%
.063	.506	CO_2	1.25%	10.02%
.075	.254	H_2O	1.48%	5.13%
3.905	3.905	N_2	77.50%	78.80%
5.043	4.964		100.04%	99.98%

F (Flowrate) Input: 100 Hz

BREATHING PERIOD TESTED: 1 to 10 Sec.

Test #	O_2 UPTAKE					CO_2 RELEASE					Breathing Period
	V_{A1} x19.8%	V_{AE} x 6.03%	UPTAKE Calc.	UPTAKE Meas.	Diff.	V_{AE} x10.2%	V_{A1} x 1.26%	RELEASE Calc.	MEAS.	DIFF.	
1 V_{A1} = 1452, V_{AE} = 1450	296	87.5	210	183	-27	145	18	127	124	-3	
2 V_{A1} = 1446, V_{AE} = 1450	295	87.5	208	182	-26	145	18	127	116	-1	(Worst Case) 1 Sec.
3 V_{A1} = 1428, V_{AE} = 1435	282	86.0	196	183	-13	143	17.8	125	122	-3	
4 V_{A1} = 1434, V_{AE} = 1435	294	89.5	204	180	-24	149	19	130	124	-6	
5 V_{A1} = 1466, V_{AE} = 1469	290	89.5	200	184	-16	149	19	130	119	-11	1.0 Sec.
6 V_{A1} = 1413, V_{AE} = 1409	280	89.5	190	183	-7	149	17.7	121	122	+1	
7 V_{A1} = 1453, V_{AE} = 1485	290	90	200	188	-12	150	18.4	132	125	-7	

NOT REPRODUCIBLE

F (Flowrate) Input: 100 Hz (Continued)

Test #	O_2 UPTAKE					CO_2 RELEASE					Breathing Period
	V_{AI} x 19.6%	V_{AE} x 6.03%	UPTAKE Calc.	UPTAKE Meas.	Diff.	V_{AE} x10.2%	V_{AI} x1.25%	RELEASE Calc.	RELEASE Meas.	Diff.	
8											2.5 Sec.
V_{AI} = 1473 V_{AE} = 1495	292	90	202	195	-7	150	18.4	132	126	-6	
9											
V_{AI} = 1473 V_{AE} = 1495	292	90	202	193	-9	150	18.4	132	128	-4	
10											
V_{AI} = 1503 V_{AE} = 1512	299	91	208	198	-10	151	19	132	135	+3	
11											3.5 Sec.
V_{AI} = 1497 V_{AE} = 1512	298	91	207	199	-8	151	19	132	144	+12	
12											
V_{AI} = 1497 V_{AE} = 1512	298	90	208	197	-11	151	19	132	130	-2	
13											
V_{AI} = 1602 V_{AE} = 1608	317	97	220	210	-10	161	20	141	138	-3	
14											5 Sec.
V_{AI} = 1599 V_{AE} = 1607	317	97	220	209	-11	161	20	141	141	0	
15											
V_{AI} = 1607 V_{AE} = 1608	319	97	220	208	-12	161	20	141	138	-3	

NOT REPRODUCIBLE

F (Flowrate) Input: 100 Hz (Continuous)

Test #	O_2 UPTAKE					CO_2 RELEASE					Breathing Period
	V_{AI} x 19.8%	V_{AE} x 6.03%	UPTAKE Calc.	UPTAKE Meas.	Diff.	V_{AE} x10.01%	V_{AI} x1.25%	RELEASE Calc.	RELEASE Meas.	Diff.	
16 V_{AI} = 1505 V_{AE} = 1506	298	91	207	195	-12	151	19	132	135	+3	
17 V_{AI} = 1506 V_{AE} = 1506	298	91	207	195	-12	151	19	132	134	+2	10 Sec.
18 V_{AI} = 1505 V_{AE} = 1506	298	91	207	195	-12						

	INSPIRATORY		EXPIRATORY	
	Volts	%	Volts	%
O_2	1.00	20%	.500	10%
CO_2	1.00	20%	.500	10%
H_2O	1.00	20%	1.50	30%
N_2	2.00	40%	2.50	50%
	5.00	100%	5.000	100%

F (Flowrate) = 200 Hz; Period 2.5 Sec.

Test #	O_2 UPTAKE					CO_2 RELEASE					
	V_{AI} x 20%	V_{AE} x 10%	UPTAKE Calc.	UPTAKE Meas.	Diff.	V_{AE} x10%	V_{AI} x 20%	RELEASE Calc.	RELEASE Meas.	Diff.	Breathing Period
19 V_{AI} = 3073 V_{AE} = 3084	615	308	307	298	-9	308	614	-306	-304	+2	
20 V_{AI} = 3070 V_{AE} = 3084	614	308	306	300	-6	307	614	-307	-307	0	2.5 Sec.
21 V_{AI} = 3072 V_{AE} = 3084	614	308	306	300	-6	308	614	-306	-305	+1	
22 V_{AI} = 3077 V_{AE} = 3084	615	308	307	298	-9	308	615	-307	-300	+7	

F = 200 Hz, Period: 2.5 Sec.

	INSPIRATORY		EXPIRATORY	
O_2	1.00 V.	20%	1.25 V.	25%
CO_2	1.00	20%	1.25	25%
H_2O	1.00	20%	1.50	30%
N_2	2.00	40%	1.00	20%
	5.00	100%	5.00	100%

Test No.	O_2					CO_2					H_2O				
	V_{AI} x20%	V_{AE} x25%	Uptake Calc.	Uptake Meas.	Diff.	V_{AE} x25%	V_{AI} x20%	Rel. Calc.	Rel. Meas.	Diff.	V_{AE} x30%	V_{AI} x20%	Rel. Calc.	Rel. Meas.	Diff.
23 $V_{AI} = 3001$ $V_{AE} = 3000$	600	750	-150	-148	+2	750	600	150	148	-2	900	600	300	303	+3
24 $V_{AI} = 2997$ $V_{AE} = 3000$	600	750	-150	-149	+1	750	600	150	149	-1	900	600	300	304	+4
25 $V_{AI} = 3002$ $V_{AE} = 3000$	600	750	-150	-148	+2	750	600	150	151	+1	900	600	300	301	+1

V_{AI} x40%	V_{AE} x20%	N_2 Calc. Uptake	Uptake Meas.	Diff.
1200	600	600	604	+4
1199	600	599	602	+2
1201	600	601	604	+3

	INSPIRATORY		EXPIRATORY	
O_2	.999 V	19.9%	.297 V	6.00%
CO_2	.062	1.23	.505	10.20
H_2O	.074	1.47	.253	5.10
N_2	3.900	77.4	3.900	78.60
	5.035 V	100.0%	4.955	99.90%

Test #	V_{AI} x19.9%	V_{AE} x6%	Uptake Calc.	Uptake Meas.	Diff.	V_{AE} x 10.2%	V_{AI} x1.23%	Release Calc.	Release Meas.	Diff.
26 V_{AI} = 3071 V_{AE} = 3084	611	185	426	419	-7	314	38	276	271	-5
27 V_{AI} = 3072 V_{AE} = 3084	611	185	426	416	-10	314	38	276	275	-1
28 V_{AI} = 3087 V_{AE} = 3084	614	185	429	417	-12	314	38	276	272	-4
29 V_{AI} = 3080 V_{AE} = 3084	613	185	428	416	-12	314	38	276	272	-4

Test #	V_{AI} x19.9%	V_{AE} x6%	Uptake Calc.	Uptake Meas.	Diff.	V_{AE} x10.2%	V_{AI} x1.23%	Release Calc.	Release Meas.	Diff.
30 V_{AI} = 3084 V_{AE} = 3080	614	185	429	415	-14	314	38	276	274	-2
31 V_{AI} = 3084 V_{AE} = 3080	596	180	416	409	-7	314	38	269	272	+3
32 V_{AI} = 3006 V_{AE} = 3006	600	180	420	410	-10	306	37	269	271	+2
33 V_{AI} = 2998 V_{AE} = 3006	596	180	416	409	-7	306	37	269	264	-5

IV. DATA CONTROL, STORAGE, AND RECORDING

1.0 Concept of Synchronizing Data with Time

In defining the quantitative measurement of respiratory flow phenomena, it is necessary first to establish a frame of reference with which the measurement can be described, and with which all comparison can be made according to the same standard of reference.

A logical frame of reference is the "real" clock time, where the period of measurement is defined, for instance, in terms of minutes and seconds. The clock time is represented in modern electronic systems by pulses with regular periodicity. However, human respiratory phenomena do not necessarily abide by the cadence of real time, and natural breathing cannot be expected to follow rigidly the rythems, of, say, one per second, but varies according to the person's physiological and metabolic condition.

To analyse the respiration process quantitatively, the frequency (or periodicity) of breathing wave (e.g. in number of breaths per minute) and the gas flowrate (e.g. in liter/sec) are both involved: the former is in the form of a wave train of alternating inspiration and expiration cycles; the latter is described in the electronic terminology as an analog signal.

In order to place these two dimensions of respiratory quantities within a common frame of reference, and to render them comparable and compatible with "real time", the following approaches are used in the MIRACLE System:

1. Synchronization Scheme

Since there is no way, and indeed no need, to force the human respiratory frequency to get into rhythm with the real "clock" timing frequency and yet it is necessary to express the respiratory phenomena in relation to "time", a synchronization scheme between the regularity of real clock pulse and the non-regularity of the breath-wave is devised.

2. Signal Transformation Scheme

If the flowrate is an analog time-varying type signal, the evaluation and synchronization processes become more complicated, since it involves both the (time) domain and range. Such analog signals should preferrably be transformed into quantities in the domain of "time" only, such as pulse-rate of a constant amplitude pulse-train, in order to facilitate the quantitative evaluation processes.

3. Data Combination Scheme

Instead of using two data channels to study the two aforementioned aspects of respiration, these can both be combined and compressed into a single data channel on the time domain; and, that this data channel can be effectively synchronized with the real time clock pulses. Before this is done, however, the analog signal needs to be electronically normalized, or compensated (e.g. regarding dynamic response characteristics), and then converted into higher frequency pulse-rate form which, in turn, is superimposed onto the alternating inspiration and expiration waveform of a lower frequency.

The two physiological quantities of respiratory flow are thus transformed into a kind of "physiological clock of respiration". By doing so, it becomes possible to synchronize and correlate all the respiratory flow measurements with the real clock time with simplicity, and greater accuracy, yet providing all the necessary information required by respiratory analysis. An additional advantage with the aforementioned approaches is that, with the hybrid electronic computing techniques, the chemical components of the respiratory flow such as O_2, CO_2, N_2 and H_2O, which have been separated quantitatively with the aid of a mass spectrometer (or other analytical means), can likewise be expressed as quantities on the time domain; each of these quantities can be synchronized with the real clock time. The advantages are both numerous and apparent: metabolic measurements can now be accomplished accurately on a one-minute, half-minute, breath-by-breath or any other period basis, while completely satisfying the "minute-volume based on T_M" requirement customarily used in biomedical practice. Because of the wider frequency response of the QL respiratory flowmeters, this arrangement also makes possible the much-needed study of the dynamic breathing phenomena, which has, thus far, been denied to our biomedical research. The importance of studying rapid physiological changes is well known; however, due to the past absence of modern respiratory flow measurement and synchronization methods, such dynamic measurements were conducted haphazardly by means of slow response spirometers - often in a manner quite incompatible with the actual physiological requirements of men-under-test.

1.1. With these clear advantages, a number of innovations are included in the MIRACLE System:

1. The respiratory flowrate is provided in frequency form, which is much higher than the frequency of the alternately inspiring and expiring breath wave.
2. Specially designed Electronic Bi-directional Discriminator, or, Inspiring/Expiring electronic switches (INEX Switch) is used to provide the principal control in the MIRACLE System. Its function is manifold: they generate, in a single data channel, superimposed wave-train containing precise information on both the respiratory flowrate and on the frequency (or periodicity) of the breath-wave, resulting in the so-called "Physiological Clock of Respiration" which is then amenable to the aforementioned quantitative analysis. They also sort and separate which part of the respiratory flow is due to inspiration, and which part is due to expiration, thereby channeling the data for separate computation and analytical processing.
3. In order to fulfill the requirement for the Minute Volume/T_M mode of respiratory measurement, which is customarily used in the biomedical field, the "physiological clock" must be synchronized with the "real clock" without any appreciable loss of respiratory data. This is accomplished electronically by the use of control logic.

To describe more fully, the synchronization between the real clock and the "physiological clock of respiration", one must further elaborate that the electronic control logic of MIRACLE II itself uses at least three different kinds of clocks: a) the one-minute clock, which is used mainly to provide a minute-by-minute standard, and for alerting or synchronizing the human breathing function to real time; b) the free running clock which

is used in a variety of control logic functions in order to make possible the synchronization process; c) the printer clock which is used to regulate the scanning sequencing and data printing functions; otherwise valuable data cannot be effectively gathered and recorded. Functionally, the division of labor can be described as follows:

1.2 Metabolic Data Acquisition

The physiological clock of respiration is provided jointly by two separate sources: a) the timing provided by the frequency output of the respiratory flowmeter and its electronic auto-corrector which is used to correct any possible dynamic response error due to the flowmeters' time constants; b) the timing of the flow direction discriminator regarding the breathing period; c) the % partial pressure outputs from the mass spectrometer (e.g. O_2, CO_2, N_2, H_2O) in analog form.

Both a) and b) provide pulse type signal which when superimposed together into a single pulse train, forms the physiological clock of respiration. This pulse-train is then "DELAYED" by a proper amount of delay-time so that a, b, and c can be computed together accurately without time-mismatch.

1.3 Sorting of Respiratory Data

The pulses of the physiological clock of respiration are sorted and identified according to whether they are due to inspiration or expiration. This function is controlled electronically by the BI-DIRECTIONAL DISCRIMINATOR.

In order to compute the uptake rate, (for instance O_2), the inspired flow quantity must be subtracted by the expired flow during the same breathing cycle. This is accomplished through the up-down counter.

The function of "up" or "forward" can be regarded as the repeated addition or accumulation of the inspiring flow counts X^i at the timing rate (frequency) as provided by the inspiring flow. This flowrate, provided in frequency form, is applied to the Up Counter during only the gated-in inspiration period, but was "gated-out" during the expiration period when the up-counter receives no signal.

Similarly, the counting of the "down" or "reverse" counter operates downward when it receives output pulses X^e from the expiring flow during the gated-in expiration period. It serves the function of successive subtraction of the expiring flow quantity from the already accumulated inspiration flow quantity. No flow signal is applied to the "down" counter during the gated-out inspiration period.

The result of the consecutive "up" and "down" counting over a given period, say near one minute, gives the desired uptake quantity (or release quantity), or minute volume of the respiratory gas, such as O_2 (or CO_2) which is stored and printed out at command when the last breathing cycle within the prescribed period (e.g. 1 minute or 30 sec.) is completed. Thus, the up-down counter is used essentially in totalization and algebraic subtraction modes.

It is important to point out that during the aforementioned operation of the up-down counters; each transition (or change-over) from "down" to "up" counting following the synchronization with the "one-minute" pulse, provides the timing of the T_M period.

1.4 Synchronization for T_M Mode Measurement of "Minute Volume"

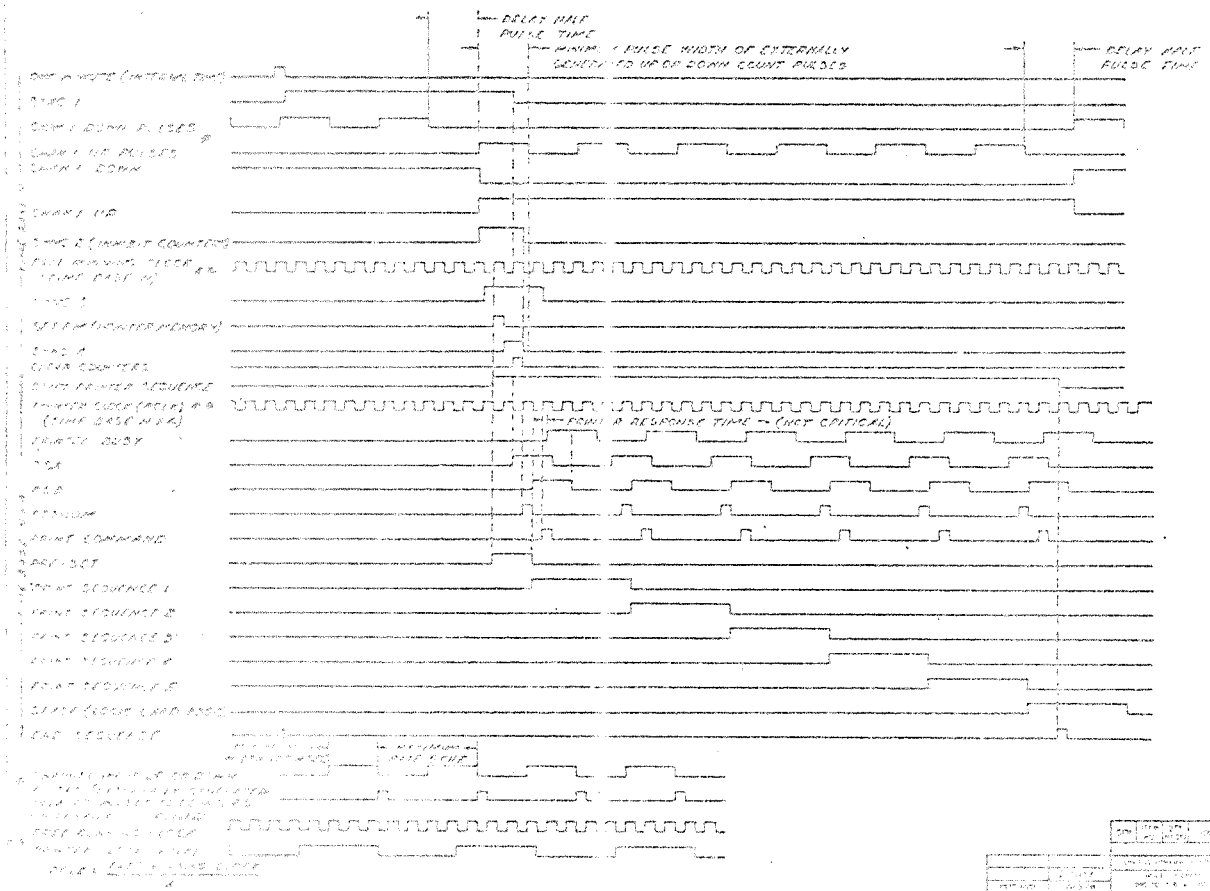
The T_M is defined as the time period to encompass a series of complete breath functions, measured from a starting point on the breath wave form to a corresponding point after a time of approximately 1 minute. It changes from time to time and is not a fixed period. In order to describe the synchronization of this non-constant T_M period with say, one-minute period, the function of the control logic is illustrated by the accompanying timing diagram, (FIG. D-1)

As used in the MIRACLE System: each T_M counting period is preceeded by a one-minute pulse which serves merely to alert the entire system. The T_M period does not start, and no counting takes place until the circuit is "hit" by the pulse front of the first inspiratory breath of the period. This pulse-front marks the transition from down to up counting at the command of the directional discriminator. Following this, the up and down counting proceeds continuously without stop even at the instant of receiving the second one-minute pulse whose function is again to ready the system to stop the counting. The counting process is only terminated at the occurrance of the first starting pulse-front of the inspiring breath wave of the next T_M period. Thus, although the counting is "regulated" by the independent one-minute pulses, (or, in other modes, e.g. by 30-second pulses), its period is precisely T_M, governed only by the arrival of the first and last breath wave nearest the one-minute (OM) pulser. As soon as each counting period is completed, the resultant count is transferred to the storage memory; the counter is immediately cleared, and ready to count again within a period of microseconds. Thus, virtually no breath information is lost [illegible]

1.4 Synchronization for T_M Mode Measurement of "Minute Volume"

The T_M is defined as the time period to encompass a series of complete breath functions, measured from a starting point on the breath wave form to a corresponding point after a time of approximately 1 minute. It changes from time to time and is not a fixed period. In order to describe the synchronization of this non-constant T_M period with say, one-minute period, the function of the control logic is illustrated by the accompanying timing diagram, (FIG. D-1)

As used in the MIRACLE System: each T_M counting period is preceeded by a one-minute pulse which serves merely to alert the entire system. The T_M period does not start, and no counting takes place until the circuit is "hit" by the pulse front of the first inspiratory breath of the period. This pulse-front marks the transition from down to up counting at the command of the directional discriminator. Following this, the up and down counting proceeds continuously without stop even at the instant of receiving the second one-minute pulse whose function is again to ready the system to stop the counting. The counting process is only terminated at the occurrance of the first starting pulse-front of the inspiring breath wave of the next T_M period. Thus, although the counting is "regulated" by the independent one-minute pulses, (or, in other modes, e.g. by 30-second pulses), its period is precisely T_M, governed only by the arrival of the first and last breath wave nearest the one-minute (OM) pulses. As soon as each counting period is completed, the resultant count is transferred to the storage memory; the counter is immediately cleared, and ready to count again within a period of microseconds. Thus, virtually no breath information is lost durin



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the change-over of measurement cycles.

1.5 Measurement of the T_M Period

To time the beginning and ending of each T_M period, a 16.667 Hz clock pulse train is gated through during this period to an up counter. With this clock frequency, the resultant counting of T_M is expressed directly in units of thousandths of a minute instead of the conventional seconds.

1.6 The Functions of the Control Logic in the Synchronization Process

Prior to describing the successive timing actions of the various elements in the control logic, it is necessary to specify the purpose and function of these elements. The following provides a summary of their functions:

1. The entire system is alerted, or synchronized with, but not directly controlled by the real time clock which provides timing pulses only as "one-minute (OM)" reference (also 1/2-minute and shorter periods). In other words, while the entire operation (counting and print-out etc.) is based on the actual physiological timing or tempo of human respiration, it is only synchronized with and regulated by the one minute clock.
2. The logic system instructs the system when to start and when to stop counting.
3. It also instructs the counters of the system, when to count upward (e.g. for inspiring flow of O_2), and when to count downward (e.g. for expiring flow of O_2) so that arithmatically the uptake and release

rate can be computed from non-simultaneous inputs.

4. The logic also commands the storage memories to:

   a) Clear the previously stored (BCD) data in order to ready the memory to receive fresh data;

   b) to inhibit the counter from further counting when ready to print;

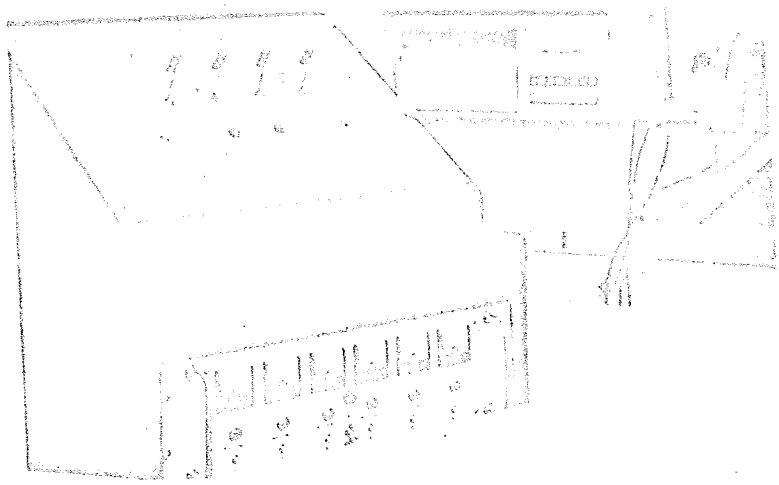
   c) to transfer the BCD data from the counter to the memory for storage.

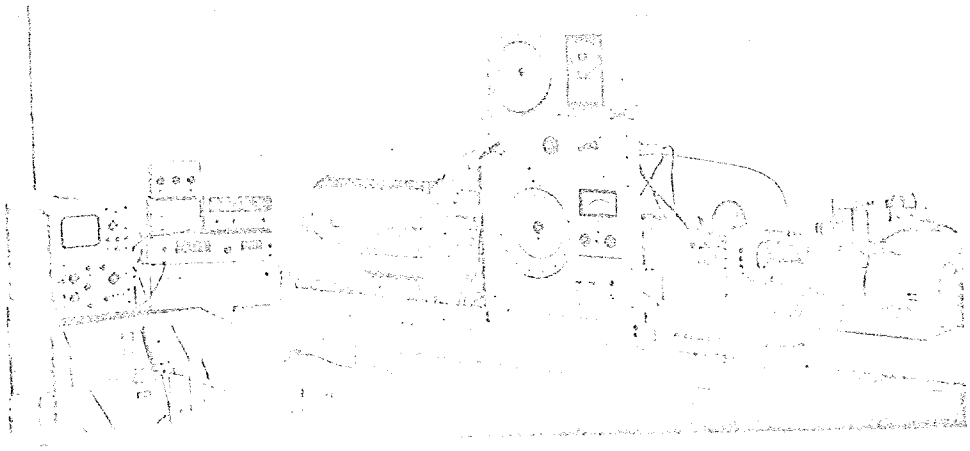
5. It "interrogates" the digital printer if it is "busy" or "ready" to receive signal from the memory for print-out.

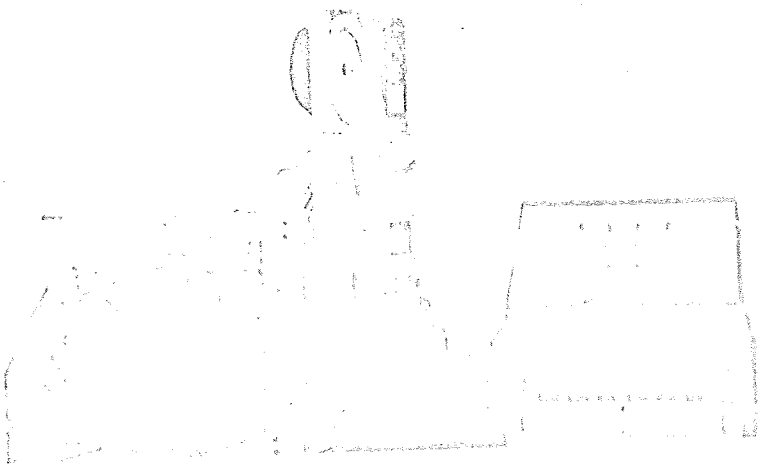
6. The logic instructs the recorder to print-out the data according to the proper sequential order of the counter numbering.

At present, three different versions of the data system have been completely developed. The simplest is a 2-channel version using Nixie digital readout and is intended only for the measurement of metabolic phenomena due to O_2 and CO_2 quantities. The second version now in use with the MIRACLE II System is an 8-channel unit. The same design is modified to gear with large scale digital data acquisition system for inter-facing with digital tape recorders and large-scale computers.

These versions are shown in photographs of previous section.







ARTIFICIAL LUNG TESTING

Servo-Controlled Artificial Lung Bi-directional Calibrator.

The Respiratory Simulator or Artificial Lung is a system designed to simulate respiratory conditions of various modes. Constructed from a stainless steel bellow, its volume change is closed-loop controlled by a servo motor system, which can be set at various precisely known rates. It is now used primarily to perform dynamic calibration and testing of respiratory flowmeters. However, when operating in conjunction with a bi-directional respiratory flowmeter, the system can be made to function as an automatic resuscitator which supplies breathing gas in synchronization with the user's needs, and at the desired quantity and rhythm.

The present system can simulate bi-directional respiration under atmospheric or sub-atmospheric pressure conditions; operating at cyclic rate of from 1 breathing (inspiration and expiration) cycle per second to 10 cycles per minute, and flow rates ranging from 0.5 to 150 liter/min.

The simulated flowrate can be of various wave-forms: flat-top type square waves, triangular waves, ramp functions - with adjustable rising and falling slopes. The reason for choosing the flat-top type waveform is to make possible highly precise steady-flowrate calibration, while the adjustable (rising and falling) slopes of the waves are designed to facilitate the determination of the flowmeter's time constant.

The system's capability is not limited to normal atmospheric pressures: it simulates breathing under sub-atmospheric pressure conditions to below 1 psia (or 50 mm Hg); or at pressure above 1 atmosphere. The artificial lung

generates and simulates respiratory volumetric flowrate by controlled time rate of change of volume of the lung. A wide range of different mass flowrates can also be simulated by varying the pressure and the temperature at any particular volumetric flowrate.

Description

The system consists of essentially 3 units:

1. The Functional Unit - consists of a) a glass bell jar (15-inch diameter, 24-in. length); or b) a stainless steel dewar (11-in. diameter 18-in. deep) cylindrical evacuable chamber in which is housed a stainless steel bellows type artificial lung. For higher pressure and gas density experiments, the chamber is of sealed stainless construction. In order to facilitate visibility of the motion of the "lung", a large glass bell-jar with sealed base is used as the evacuated chamber for sub-atmospheric and low-gas density testings. The motion of the lung is provided by a closed-loop electronically speed-controlled motor (PMI-U12M4 motor: torque 55 oz-in. at 3650 RPM) through gears which transform rotational motion into linear motion by means of a ball-screw. The motion of the gear is continuously measured by a cryotronic pickup which produces a pulse-rate output. The mode of the movement can vary both in length of travel, period, rate, and in waveform by means of electronic and mechanical adjustments. The opening of the lung is connected to a tube or adaptor, in which is installed a turbine flowmeter and, when it is safe to use, a hot-wire anemometer for wave form comparison. The waveform of breathing can therefore be measured and compared accurately by

two or three independent methods. The waveform of the lung's motion is obtained from the pulse-rate generated by the teeth of the gear, using the cryotronic pickup and FPAC Transient Flow Indicator. This pickup-electronic arrangement is identical to that used by the respiratory flowmeter, so that more objective comparison between the flow-rate and the rate of change of the "lung's volume" can be made.

A linear potentiometer, or a LTVT, is also provided for the supplementary indication of the position and motion of the "lung", although the afore-mentioned gear teeth method is by far more preferable.

Inside the chamber are also housed a temperature control device. A thermocouple and a platinum resistance thermometer are provided so that the temperature inside can be measured. The absolute pressure inside the chamber can be precisely controlled and is measured by a Texas Instrument Quartz Bourdon gage, or a W.T. absolute pressure gauge. The chamber is connected to a vacuum pump via a Cartesran manostat, or vacuum regulator, so that a wide range of subatmospheric pressures can be maintained for simulating the respiratory condition of high altitude or inside space vehicle.

2. The Control Panel - The control of the entire system is carried out at the control panel. The panel is equipped with an absolute pressure gage, an iron-constantin thermocouple, and a platinum resistance thermometer readout instrument for indicating both the absolute pressure and temperature inside the chamber. The modes of respiration, namely, increase of the level of the flowrate and the "ramp" slopes and the chart of respiratory flowmeter, are also controlled at the panel.

THE LOW PRESSURE CALIBRATION OF ORDINARY TURBINE FLOWMETER USING ARTIFICIAL LUNG

A standard OL gaseous flowmeter of ordinary design was used in the worst case study of the subatmospheric pressure effects on calibration under simulated bi-directional breathing condition with the artificial lung. The objective is primarily: a) to pinpoint the role of dynamic error in bi-directional flow measurement at subatmospheric low pressure for "uncorrected" flowmeter; and, b) to study the degree of reproducibility of integrated output for example, over period of 1 minute under various low pressure condition from 1 to 14.7 PSIA. The pressure range of interest is from 5 PSIA upward.

The testing procedure is as follows:

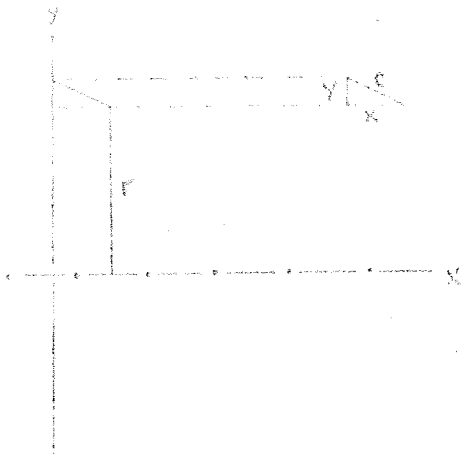
1. Since the volumetric flowrate is provided by the bi-directional linear motion speed of the artificial lung; this speed as well as the frequency(of pulsation or breathing) are maintained throughout the entire sequence of testings covering various pressures.

2. For each test, the vacuum regulator and manostat are set to maintain precisely at a particular absolute pressure throughout that particular test. The temperature inside the chamber is maintained at a const. 70°F value.

3. Each series of tests begins with the data taking at 1 atm.; the counts are taken bi-directionally, namely, the summation of counts for both inspired and expired output are totalled for a 1 minute-period. The resultant data of several runs at the same setting are averaged to prove the mean calibration value.

The tests are repeated for another absolute pressure. In order to study the trend of data, these tests were carried out to pressure as low as 1 PSIA.

ANALYSIS OF THE ARTIFICIAL LUNG'S BELLOW MOTION



$$x^2(t) + y^2(t) = c^2$$

$$y^2(t) = c^2 - x^2(t)$$

$$y(t) = \sqrt{c^2 - x^2(t)}$$

Adding a constant r to the rightside of the equation raises the curve (line) by r.

$$y(t) = \sqrt{c^2 - x^2(t)} + r$$

The volume V of a bellows segment is equal to two times the volume obtained by rotating the above curve around the x-axis.

$$V = 2\pi \int_{x(t)=0}^{x(t)} \left(\sqrt{c^2 - x^2(t)} + r\right)^2 dx$$

$$= 2\pi \int_0^{x(t)} \left[c^2 - x^2 + 2r\sqrt{c^2 - x^2} + r^2 \right] dx$$

$$= 2\pi \left\{ \int (c^2 + r^2)\, dx - \int x^2\, dx + 2r \int \sqrt{c^2 - x^2}\, dx \right\}$$

$$= 2\pi \left\{ (c^2 + r^2)\, x(t) - \frac{x^3(t)}{3} + 2r\left(\frac{x(t)}{2} \sqrt{c^2 - x^2(t)} + \frac{c^2}{2} \sin^{-1} \frac{x(t)}{c} \right) \right\}_0^{x(t)}$$

$$\boxed{V = 2\pi \left\{ (c^2 + r^2)\, x(t) - \frac{x^3(t)}{3} + r\left(x(t)\sqrt{c^2 - x^2(t)} + c^2 \sin^{-1} \frac{x(t)}{c} \right) \right\}}$$

The above equation gives the volume of 1 bellows segment as a function of x, which is also a function of time.

Now, if a bellows of n segments is being linearly expanded with the speed v (vec) = const., then for each segment we have $x(t) = \frac{vt}{2n}$

Thus, the total volume of the bellows of segments is given as:

$$V_{\text{Bellows}}(t) = 2\pi n \left\{ (c^2 + r^2)\frac{vt}{2n} - \frac{1}{3}\left(\frac{vt}{2n}\right)^3 + r\left(\frac{vt}{2n}\sqrt{c^2 - \left(\frac{vt}{2n}\right)^2} + c^2 \sin^{-1}\frac{vt}{2nc}\right)\right\}$$

where by t = 0 is the time of absolute compression of the bellows. (This holds for bellows with infinitely thin walls. If the walls are of thickness τ, then a constant volume of <u>approximately</u> $(n-1)\, 2\pi r^2 \tau$ must be added to the above Volume).

The change of volume of the bellows with time is therefore given as:

$$\dot{V}_{\text{Bellows}}(t) = 2\pi n \left\{ (c^2 + r^2)\frac{v}{2n} - \left(\frac{v}{2n}\right)^3 t^2 + r\left(\frac{vt}{2n}\sqrt{c^2 - \left(\frac{t}{2n}\right)^2} + \frac{vt}{2n}\,\frac{1}{2}\left[c^2 - \left(\frac{t}{2n}\right)^2\right]^{-\frac{1}{2}}\left(\frac{v}{2n}\right)^2 2t\right) + c^2 \frac{1}{\sqrt{1 - \left(\frac{vt}{2nc}\right)^2}}\,\frac{v}{2nc}\right\}$$

i.e. $\dot{V}_B(t) \neq$ Const.

PART II.

ADMINISTRATIVE SECTION

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The following critical review of our previous progress and set-back [illegible] prepared in order to facilitate an objective evaluation by the cognizant authorities on our past contract performance.

2. A Critical Review of the Past Pattern of Contract Work

At this point of our contract work, when the efficient and accurate measurement analysis and computation of respiratory and metabolic phenomena have already become a reality; that most of the difficult problems are behind us, and, that the remaining issue is only a matter of degree of perfection, we feel it is useful for Quantum Dynamics to submit the following background report:

In 1968, MSC-NASA awarded simultaneously two contracts to Quantum Dynamics: for the procurement of an uni-directional respiratory flow system complete with a data system, and for the further product development of a bi-directional respiratory system. At that time, we had already demonstrated to MSC-NASA and to Perkin-Elmer (then a NAAS contractor) that the bi-directional type flowmetering unit was available in its essential form, at Quantum Dynamics, Inc.

2.1. During the earliest part of our work, there was this unavoidable vacillation between the uni-directional and bi-directional approach; we were directed by the then technical coordinator to place emphasis on the first approach, aiming to create and to demonstrate the feasibility of a "total" metabolic data system, leading to the required computed metabolic data such as minute volume, uptake and release rates, T_M and RQ. Such "total" system had never been attempted previously, even by much larger contractors after a large amount of expenditure. Because of the extremely limited budget, we threw in our know-how and many proprietary computer and instru-

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From hindsight, the addition of one more digit involves very little additional cost when compared with the extensive engineering effort to "make-do" with limited data capacity; or, making sacrifice with a sound approach by building a low resolution flowmeter to accommodate a lower capacity data system.

2.2.2. One other decision which may or may not be considered as error in judgment is that, perhaps, typewriter type digital readout could better identify each data on a clearer, and more readily understood basis by means of words. For example, the typewriter can print-out:

O_2 Uptake	1445 cc or mL
CO_2 Release	1720 cc or mL
Min. Volume-Inspire	18.7 Liter
R.Q.	0.893

whereas with the present series entry type HP 5050 Digital Recorder (14 columns), it can only printout the numerical number: the identification of what is what can only be based on the arrangement. The HP printer is of a much higher speed, but in metabolic recording, we only require print-out once in 1 minute, or 1/2 minute, or 1/4 minute - a typewriter is fast enough. However, one basic reason behind our early decision was that the HP digital printer-recorder costs only $2,000 whereas a typewriter requires probably twice the expenditure.

2.3. In terms of our Work and Fund losses due to Interfacing Problems With Other Contractors.

2.3.1. During the uni-directional phase of the contract, we were, supposedly, to receive from another MSC contractor, the needed uni-directional valves with built-in switches for flow-direction indication; and, these were to be mounted in the face mask. When these were not forthcoming, we were compelled to divert considerable effort from our main contracted tasks to develop a substitute "uni-directional valve and switch" although we already had solved these problems in our bi-directional indicator. We designed and fabricated two types of "uni-directional valves" - one passive and one electrically driven which operated only tolerably well, but was still not up to our desired standard. By this time, our fellow NASA contractor belatedly provided us with several valves through the technical coordinator - but with no switch. Quantum Dynamics was then requested to build switches on these valves. Three successive designs were evolved, but none entirely successful, until the fourth photo-electric one. At that time, we were struggling with such unfamiliar problems as plating gold and attaching terminals on silicon rubber - no specialty of our own." By this time, the superior efficiency of the bi-directional flowmeter and directional discriminator was amply demonstrated to NASA and another NASA contractor, S.W.I. Only then were we instructed to switch-over to the bi-directional approach. While these happenings caused substantial loss on our part

in terms of time and funds which could otherwise be devoted much more beneficially to our main contracted task; it afforded us experience on the inter-dependence between ourselves and other MSC contractors: that other contractors' failure to live-up to their commitments can seriously affect our progress and indirectly force us to suffer losses.

2.3.2. The Funds originally earmarked for our own engineering effort were lost to us. Similar situations happened in the case of the mass spectrometer, although to a lesser extent: thus far, our work has not the chance of teaming up with a mass spectrometer (hereafter abbreviated), although our MIRACLE II System is designed for, and must interface with an MS in order to effect metabolic computation. Teledyne's Earth Science Division could not realize such a system under NASA contract. Thus, we had to design an entire system with a "hypothetical" MS in mind, by assuming what the MS signals would look like.. In desperation, we created the "Mass Spectrometer Simulator-Calibrator" which proved extremely helpful to us and, in fact, is bound to be useful to NASA metabolic programs. But, this again is a case of diverting our effort and funds to meet an exigency by filling another's missing gap. Fortunately, this effort is rewarding and was in the interest of NASA's work. Nevertheless, a simulator is not a real MS. To team it with the MIRACLE II System, [illegible], an imperfect analogy, is like a shot-gun marriage imposed between a real girl and a "dummy". We still do not know the real problems of interfacing with an actual mass spectrometer.

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*For the first time, a loan collection agency, threatened to [illegible] [illegible] client for further payment of [illegible], after we [illegible] [illegible] totally useless metal coating. The contract [illegible] [illegible] we paid from our own packet to avoid [illegible]

2.4. [illegible]

Much improvement in the performance, quality and fabrication efficiency of the bi-directional respiratory flowmeter would [illegible] possible, if we had more advanced equipment; for instance, a small Electric Discharge Machine (EDM) of even the low-cost type. In the absence of which we relied solely on [illegible] and time-consuming conventional type machining process. Yet, even the most careful machining process has its limitations. It would be difficult, for example, to achieve completely identical turbine blade (topographical) contour from face-to-face and from piece-to-piece, even with our high precision Deckel spiral Mill, and operated by our most experienced master machinist. In the case of the bi-directional respiratory flowmeter we had succeeded, after repeated trials, in making the calibration in one direction very close to that flowing in other direction; but, we still cannot make these two calibration characteristics exactly identical. The contours of the two opposite sides of the blade were cut as precisely identical as we knew how, but even though we spent days of patient machining time, a single pass with an EDM (less than 10 minutes) would have achieved much better results. Another example can be seen from our often frustrating effort of further optimizing the dynamic response of the turbine by cutting the blade as thin as possible, aiming to achieve the lowest figure of moment of inertia commensurate with the strength. In milling the 9-blade turbine, we have now reduced the mass of the finished part to approximately 0.03 ounce mass; but our calculation [illegible]

if we succeed to reduce it to 0.02 ounce, we could expect a significant improvement. But when we attempted to cut it still thinner, the blade, being unsupported on one side, may bend under the slightest of machining drag. It was heartbreaking for us that, after painfully finishing the 8th of the 9 blades with such great care, the last cut bent the ninth blade and the whole piece must be rejected. Sometimes it took days to obtain one good blade out of many blades made, yet we were already using a most precise and costly machine which our small company could ill afford.

Had we obtained the use of an EDM, for example, through leasing (the lowest cost EDM was $18,000 which was too steep for us), we could have fabricated parts of improved quality and exactness with one-twentieth amount of time - and far less anxiety. We partly regret our error of not having the foresight of leasing one, but the main inhibition was due to the high monthly rental of $685 which our small contract could not afford, and after other expenditures we had nothing left for such leasing fees.

In summary, the problems of a small contractor must be appreciated by cognizant authorities at MSC-NASA. Economy-minded small contractor, struggling under inadequate fixed-price contract to achieve beyond the state-of-the-art, must either forfeit the badly needed equipment which this contract would not afford, or request the necessary funding for the technical objective.

We were attempting to fabricate space-age components with older conventional machinery at our disposal. Now that we have learned our

lesson and nearly mastered the techniques; we recognize that the short-sighted economizing measure on our part was actually very wasteful. In the next phase, where the most accurate and completely inter-changeable flowmeter components must be made, we must avoid our past mistakes and request NASA's support to enable us to lease the needed modern production equipment.

We are of the firm belief that despite working under the aforementioned circumstances, our technical successes and the rapid pace of our progress are highly significant. The greatest strides were undoubtedly made under the program supervision of Dr. Rummel, who has directed our energies toward the most essential and rewarding technological areas, and enabled us to bring a full system into fruition. During this period, we are privileged to state, that compared with any other large or small NASA contractor, Quantum Dynamics has used every one of NASA's dollars to do two or three dollars worth of work, and that in terms of contributions of breakthrough significance, one cannot actually assign a dollar value.

3. Proposed Technical Effort for the Next Phase

3.1. Objectives

The following efforts are proposed and were selected on the basis of essential need, and the consideration that these renewed circuits and techniques have already been fully or partially developed by Quantum Dynamics, but have not been implemented in previously constructed systems. Therefore, no unforeseen technological difficulty or excessive cost are anticipated.

3.1.1. Implementation of Deferred but Developed Improvements.

From the beginning, we were aware that due to cost considerations, certain circuit and fabrication techniques must be deferred to a later stage. For example, the uptake and release should be reduced directly in its final form after having been divided by the T_M. This final divisional process was not implemented in MIRACLE II System; instead we used one data channel (which we badly needed to print-out clock-time) to print out the value of T_M. Now that digital division process has been developed, we are of the belief that they should be implemented into the system. Others include the new RQ circuits, the phase difference indicator, etc.

3.1.2. Consolidation of Design Gain

While implementing the needed improvements, it also appears necessary to consolidate the tremendous amount of technological gain which we learned from the "trial-and-

ing" or "hand-made" stage of the metabolic systems. Through making post-development units of improved design and superior workmanship (at least for the more crucial sub-systems) and by arranging the circuits in a more rational modular form, the resultant hardwares can be put into active use in NASA's programs. Equally important, such technology would not be lost, or diluted, through total lack of production experience and negligence. Now that the most difficult and costly developmental period is behind us, it is relatively economical - except for the more costly purchased items - to make new units, than to take the older hand-made units apart for inclusion of new improvements. Reworking an old unit would still end in one improved old unit, and probably worsen the construction. Many parts of the present system have been reworked tens of times for successive improvement or corrective measures. We consider it far more economical to build well planned newer units.

Thus, to standardize on the proven design while implementing improved design is an imperative task for the next phase of work.

3.1.3. Completion of Unfinished Development

Complete the needed but unfinished developmental improvements which will be outlined more specifically in a later section of this proposal - which have already been initiated but require further contract work to bring them to full fruition.

3.1.4. Team Work

Under the direction of NASA, to team or coordinate our work with: a) NASA planned Mass Spectrometer; b) NASA designated special computers. These works would enable us to locate the problem areas, and to solve any remaining operational and interfacing problems. For instance, to construct the necessary hardwares, or make needed system modifications as required by effective integration.

3.2. Specific Technical Tasks

3.2.1. Data System

The data system used in MIRACLE II System is already highly developed. The achievement is an extremely valuable one, although it is fabricated in hand-wired form. Since it is an integral and vital part of the metabolic data system, we respectfully propose that we should consolidate our technological gain by transforming it from its present rough, bulky hand-wired form into reasonably compact, preferrably miniaturized hybrid circuit form, with the minimal of expenditure - but at least in printed circuit board form.

In the present system, the power of the data system can be enhanced manifold with the following improvements:

a) Add "timing" identification, showing [illegible]

0019 (Data) _____

0015 (Data _____

For long periods metabolic data-taking, the time identification of what transpired at what time is of measurable importance to the processing and correlation of data. This was not done because there were no funds for digital clock, and the 8th channel was diverted for T_M.

This proposed addition is technically simple: a digital clock (which we can build ourselves, if needed) with BCD output and the addition of 4 digital columns to the digital printer (parallel entry) or print-out typewriter.

b) It is virtually imperative that in any future data system, the 4-digit bit counters should be changed into 5 digit counters.

c) Transform the present design so that the uptake and release data become the result of "divided by T_M" process using digital computation. The technique has already been developed and is ready for implementation.

d) The same newly developed technique, namely direct mathematical division process from BCD coded data, can also be used to calculate the RQ in a simplified and more accurate manner. During the past, we used Digital-to-Analog (D to A)Converter; then, sample-and-hold circuit; then, a special hybrid PT/T Computer using a fixed T; then an up-counter and finally the resultant count is converted to BCD. Using the new technique, greater accuracy and standardization can be expected.

e) In the present data system, we have the dynamic flowrate (in time-varying analog waveform) of the inspiratory and expiratory flow available from the Auto-Corrector. Such data will prove to be valuable for physiological investigations; but we were unable to make use of them, since there was no analog recorder to preserve such data. (See Exhibit 3A)

During the past, both the inspiratory and expiratory signals are of + sign due to the need of the subsequent MIRACLE II data processing. Now, we can easily make the inspiratory flow signal of positive polarity and the expiratory signal negative polarity, for clearer and more logical continuous recording purposes. If NASA-MSC deems it advisable, a multi-channel tape recorder can be used for simultaneous recording of this analog data trace, and all the other seven or eight channels of computed metabolic data,(either in BCD form or D-to-A converted analog form). Such data would be invaluable for future work. (See Fig. 2) Our present task, however, is limited only to providing an output which can be recorded in both positive and negative signals.

3.2.2. Bi-Directional Respiratory Flowmeter

As we have reported previously, the performance of the bi-directional respiratory flowmeter can still be improved significantly if we can lease an EDM equipment for more precise and more efficient fabrication of the flowmeter parts, replacing our present time-consuming hand-machining processes. With such equipment, parts can be made more speedily and uniformly from piece-to-piece.

Experimentation can also be made on the use of new metallic materials. Moreover, the unit cost can be greatly reduced due to the far less man-hours required for the fabricating process. We propose that this be considered a much-needed task for the next phase of work.

3.2.3 Auto-Corrector and Computer Electronics

In addition to the proposed improvement of flowmeter; a) the improvement of the present uni-directional/uni-polar auto-corrector; b) the range extension and voltage stabilization improvement of the hybrid PF/T Computer; and c) the range extension of the FPAC Transient Flow Indicator are the three vital necessary achievable tasks for the next phase of work. All of these are working. It is only necessary to improve them in order to achieve optimal performance. The reasons for proposing these tasks are valid; and, we are particularly grateful that this necessity was reinforced by the results of the recent preliminary checking of the computing accuracy on the border-line case with the MIRACLE II System by Dr. Rummel's staff. The following paragraph will present the technical reasoning for these proposed tasks.

3.2.4 Further Improvement of the Dynamic Flow Auto-Counter

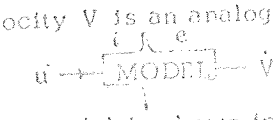
a) Introduction:

As shown in the sections of this report, in particular, the analyses and data contained therein, Quantum Dynamics' Bi-directional respiratory flowmeters are dependable and efficient instruments which are accurate, sensitive, light-weight and reliable in atmosphere and sub-atmospheric measurement of respiratory flowrates. In the case of liquid, the time constant of our flowmeter is less than 2 millisecond. However, when used to measure low density and low flowrate, the meter's response time regarding the rising, falling and coasting should, and can be further optimized in order to achieve optimal transient response adequate for the most accurate bi-directional dynamic flow measurement, particularly, for space-flight 1/3 atmospher applications. As we have demonstrated, if the equation of motion of the flowmeter is established, it is possible to build a computer (analog, digital or hybrid) that will compute true flowrate given the output of the turbine flowmeter. This feasibility has already been proven conclusively, and it remains for the present proposal to request some further necessary improvements.

b) Flowmeter Model

$$A\dot{V} + BV = pa\,|u|\,(u\sin\theta - v\cos\theta) \qquad (1)$$

The above equation (1) was implemented as a model for the uni-directional flowmeter, where input flowrate u is replaced by a voltage $\dot{V}$, and the blade velocity V is an analog output voltage $\dot{V}$.



A block diagram of the model is shown in Figure 1. Several special models have been constructed and tested with good results. With

this assurance, it is proper for us to propose the bi-directional Auto-Corrector which is much more powerful and appropriate for the tasks ahead of us.

The need for transforming the uni-directional to bi-directional auto-corrector is a logical step, since we are no longer dealing with the less efficient uni-directional and uni-polar data approach but with the bi-directional and bi-polar data which, by their very nature, are in much better conformance with natural breathing phenomena. By forcefully reversing the signal to achieve uni-polar output, we in fact create large discontinuity. As such, the uni-directional uni-polar auto-corrector introduces an "artificially" generated "transitional (from inspiratory to expiratory flow and vice versa) error" near the zero crossing region due to differentiation of the discontinuity(which is actually non-existent, but was introduced by the unipolar conversion), which is larger when the flowrate is low; therefore, particularly annoying in low flowrate-high RR breathing condition. The recent tests by Dr. Rummel's staff at NASA-MSC Environmental Physiology Laboratory have shown that at low-flowrate/high RR the effect of the transitional error becomes more pronounced, while at normal and higher respiratory flowrate/normal and low RR condition, we have proven that the metabolic computation of uptake and release can become very accurate. The reason is: under the former condition the transitional error occupies a high percentage in each breath because of less number of counts, and that this higher percentage error is compounded many more times in the case of RR=60 breaths/min., than in the case of 12 breaths/min. But this transitional error actually should not be there and can be nearly completely removed with the Bi-directional Auto-Corrector. Thus, the main source of error can be reduced or eliminated.

Equally important, the Bi-directional Auto-Corrector is designed on the basis of high-density differential equation computation. Therefore, there is less noise and a minimum of signal discrepancy. A QL flowmeter has three pickups each providing high output frequencies. We are using only one of the 3 pickups, operating at low frequency output, and are tolerating the unnecessary loss of computing accuracy and resolution; since it is well known that differentiation is, in general, more accurate for short than long time intervals; at higher rather than lower repetition rate.

It is our recommendation that the Bi-Directional Auto-Corrector, in combination with new PF/T and FPAC Computers operating in high computation density, will improve the metabolic measurement to a high order of dynamic, or, breath-to-breath, accuracy - totally adequate to meet the requirement of the NASA program.

3.3.5 Improvement of the Special Computers

There are eight (8) PF/T Computers, two (2) FPAC Computers, and two (2) MY Analog Multiplier-Divider in the present MIRACLE II System. These are high accuracy special purpose computers of our own design, none of which can be obtained from the commercial market for such special functions; or having such good accuracies.

The PF/T and FPAC Computers shoulder the most critical functions in real-time metabolic computation; and the FPAC Transient Flow Indicator occupies the central role in the measurement of and recording of the dynamic breath-by-breath respiratory flowrate.

These two computers were designed some ten years ago, although successive improvements were later added.* Their solid-state [illegible]

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* Berw[illegible] T.W. and Liu, F.F., [illegible]

were completed some five years later. At that time, a flowmeter frequency output of 1 KHz was considered unusually high, therefore these computers were designed to accommodate input frequency up to 1000 Hz. For the present MIRACLE System, we stretch their bandwidth to 1500 Hz; or, in the case of the FPAC, to 2000 Hz at slightly decreased computing accuracy. If we put forth some additional effort, we can easily improve these computers to accommodate input frequencies of up to 6 Khz. Because of the limitation of the present design, we can only make use of the frequency output from one of the three cryotronic pickups, thus sacrificing some of the achievable high resolution advantage of metabolic measurement.

a) Improvement of FPAC Computer

The FPAC Computer is essentially a non-linear K/t reciprocal computer, achieving linear input-output relationship by means of non-linear impedance network using diodes and resistors - somewhat analogous to "function generator". There are at present fifteen (15) section-by-section linearization processes. If the input frequency is below 1000 Hz, then the equalization would be adequate; the input-output relationship is linear; and a $\pm$ 0.25% accuracy is achievable. This high performance is evidenced by the linear transfer function and high accuracy shown in the calibration charts of the MIRACLE II System's Instruction Manual, when the one-inch Lower Resolution Flowmeter is used. However, as reported in the "Instructon Manual for High-Resolution Bi-Directional 3/4" ID Respiratory Flowmeter", if we force the present FPAC to accept 2500 Hz, which the 3/4" flowmeter is capable of producing then we suffer some non-linearity over the low frequency region. The improvement is largely aimed at the computer's non-linear impedance network. By increasing the number of straight line approximating sections, this non

frequency non-linearity problem of wide-band operation can be solved, using today's hybrid circuit construction techniques.

b) Improved PT/T Computers

For PT/T Computers, it is desirable to use higher frequency input; particularly in order to minimize the dynamic "counting" errors during the ascending and descending slopes of the Mass Spectrometer output. The computer is based on the so-called "skip pulse count" concept. If the count rate is high, then the dynamic error can be made very small. With the present MIRACLE II System, a discrepancy of $\pm$ 6 counts non-repeatability can occur, although this is only about 1 to 2 percent of the total print-out values.

c) Summary

The improvement of these two computers is merely a matter of extending their frequency range, and circuit revision using higher quality circuit components. No major technical difficulty is anticipated, and the cost involved is relatively small. The requirement of high computing density, as justified by studying the results of recent preliminary tests at NASA Environmental Physiology Laboratory, particularly when the extreme cases of low flowrate/high RR are concerned. Although good computing accuracy has been demonstrated, when operating within its designed realm, the proposed improvement of these two computers should nevertheless be considered as necessary tasks to be undertaken as soon as possible.