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# A New Method of Measuring the Impedance of the Human Respiratory System at Moderate Frequencies

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# A NEW METHOD OF MEASURING THE IMPEDANCE OF THE HUMAN RESPIRATORY SYSTEM AT MODERATE FREQUENCIES

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SUMMARY

The stability of aircrew breathing equipment depends not only on the stability of the oxygen regulator, but also on the impedance of the system it feeds, which includes the user's respiratory system. A method of measuring the human respiratory impedance, in the range 5-90 c/s, by comparing oscillating pressures at two points in an external reference system, is described; and results are given for nose and mouth, heavy and light, breathing. The effect of altitude, and of increased external resistance to breathing are briefly discussed. The importance of correct representation of man's impedance in dynamic testing of oxygen equipment is illustrated by reference to the characteristics of a present-day simulator and system, and an improved type of simulator is suggested.

*Replaces RAE Technical Report 66296 - A.B.C. 30004

## CONTENTS

## 1 INTRODUCTION

Oxygen equipment in military aircraft consists essentially of a high-pressure supply; a regulator delivering oxygen, or an air and oxygen mixture, on demand at pressures appropriate to ambient conditions; and a system of piping, valves and mask connecting the regulator to the man. The regulator is a complicated servo-mechanism feeding the system and the man's respiratory system, and endeavouring to maintain a constant datum pressure at its outlet, while man is also a servo-mechanism, trying to extract a desired flow of gas from the system. The arrangement is shown schematically in Fig.1. Having two sensitive control mechanisms, and piping and valves with their own pneumatic and mechanical characteristics, such an arrangement is likely to give rise to undesirable pressure oscillations, and, in fact, practically all aircrew demand oxygen equipments exhibit some form of damped or sustained pressure oscillation with frequencies from 5 c/s upwards. The word 'instability' has been used in this connection to describe any oscillation of pressure which is of such frequency, amplitude and duration as to cause physiological, psychological or mechanical trouble. This somewhat loose use of the word will be continued in this Report.

As far as the mechanical effects of instability are concerned, permissible limits may be established by endurance testing; but physiological and psychological limits are less easily determined. Samuel[1] could detect no decrement in performance of aircrew in a flight task simulator when subjected to a continuous inspiratory pressure oscillation at about 5c/s of undefined amplitude; but the pilots gave the impression that this would worry them in the air, and one stated that he would declare an aircraft unserviceable if he perceived such an instability. Thus it seems that the problem is mainly psychological; and therefore that no instability perceptible by the user should be allowed. The results obtained from some brief experiments to establish the level of perception of subjects wearing a standard aircrew mask are shown in Fig.2. Although the scatter between subjects was large, all were particularly sensitive to pressure oscillations in the region 20-50 c/s (where some could apparently perceive amplitudes as low as ±0.05 am w.5) and least sensitive to low frequency oscillations, where they subconsciously adapted their breathing to the pressure oscillation.

It is thought that no breathing equipment would be acceptable in aircraft if instability above the lower level of perception were sustained for a large part of the breathing cycle, and it is therefore necessary to establish tests which will show up any such effects.

Instability may sometimes be traced to one component such as a valve, and cured by small modifications. Occasionally the regulator may be unstable in itself, so that it acts as a pressure oscillator, independent of the system. This is basically a fault in design or construction, and its cure must lie in the hands of the makers. Both these types of instability will be detected in steady flow testing, or dynamic testing with almost any type of simulator, or any user.

But sometimes a less obvious type of instability occurs, which is a function of the whole system. A slight change in the piping between the regulator and the mask may produce instability in a previously stable system: or a system which is stable when the user breathes through his mouth may be unstable when he breathes through his nose: or it may be unstable for a small proportion of users only. This is the case when the stability depends on the impedance fed by the regulator, that is, the combined impedance of the system and the user. This effect will be discussed in more detail in Section 2.2.

Hence, in order to ensure complete stability, it is necessary to use a dynamic method of test including an impedance representative of man's respiratory system. But in order to construct the necessary analogue, either mechanical or pneumatic, we need to know the human impedance presented to aircrew oxygen equipment, that is the impedance of the respiratory system and face of healthy young men, seated, for various types of breathing.

A study of the literature showed that several methods have been developed in recent years for measuring the impedance of parts or all of the human respiratory system. Unfortunately, from our point of view, a large proportion of the literature is devoted to finding variations due to pathological conditions, often with the subjects lying down, so that data on healthy seated subjects is limited. Also, since aircrew equipment sees the complete human system, (including part of the face, the mouth or nose airways, the airways and lungs, ad the mechanical effects of the diaphragm and chest wall,) none of the numerous papers in which the flow through the mouth is compared with alveolar pressure (measured either by an interrupter technique or by an oesophagal balloon) is strictly relevant to our problem, since the chest wall is here deliberately excluded. The nearest approach to our conditions is described in papers by Dubois et al[2] and McIlroy et al[3], where the impedance of the whole respiratory system is measured, though excluding the face. Dubois applied a sinusoidal pressure oscillation at the mouth, measuring the flow and pressure there. Though the results given extend to about 15 c/s, it is stated that abwe about 8 c/s there were difficulties in measuring the flow, due, apparently, to phase shift in the flowmeter used. McIlroy used a method for obtaining the resistance and compliance of the respiratory system, by comparing the expired volume with the flow rate in relaxed expiration at constant flow rates. His method obviously gives characteristics at very low frequency, and any inductive component of impedance is deliberately excluded.

An alternative approach to the problem is to construct an analogue, usually electrical, based on the extensive anatomical data available. Again, most such analogues described in the literature refer to parts of the respiratory system only, and, even so, are extremely complicated. (See, e.g., Ref.4.) Shephard, however[5] gives a comparatively simple circuit, including all the system we need except the face. A copy of this analogue was constructed at R.A.E., and the impedance results obtained from it are shown in Fig.3. The author however states that this analogue did not agree exactly with his experimental results using the interrupter technique.

Thus it was thought that the published data available was insufficient for our purposes, since except for Shephard's somewhat inconclusive work none referred to frequencies above about 15 c/s, and also none included the face. The published data will be discussed. further in Section 3.2.2.

A programme of work was therefore undertaken to determine the impedance of the human respiratory system as it affects aircrew oxygen equipment, that is, the face within the mask cavity and the overall effect of the nostrils and/or mouth, the airways and lungs, for the various types of breathing likely to occur in aircraft. The frequency range was determined by the equipment in use at 5-90 c/s, covering the range of at least many of the instabilities noticed previously. The investigation was confired to tie sort of subjects likely to use aircraft equipment, that is, healthy young men, and in the posture usually taken by aircrew, that is, seated.

There is no reason why the methods used should not be applied to other groups, or inert systems, and in fact some tests on breathing simulators and aircrew systems were also carried out.

## 2 GENERAL THEORY AND EXPERIMENTAL "METHOD

### 2.1 Definition of impedance

The impedance of a pneumatic or pneumo-mechanical assembly may be defined as the ratio of the pressure drop across the assembly to the volumetric flow of gas into it. This is equivalent to the definition used in acoustics, although the acoustic wording - the ratio of sound pressure at a source of sound to the strength of the sound at the source - appears rather different.

If the pressure is measured in dynes/$cm^2$, and the flow in $cm^3/s$, the unit of impedance which might be called a pneumatic ohm, is the $(gm/cm\ s^2)/(cm^3/s)$, or gm $cm^{-4}$ $s^{-1}$. In medical respiratory work, the pressure is usually measured in cm water gauge (cm w.g.) and the flow in litres per second (l/s), so that the medical unit of impedance is (cm w.g.)/(l/s). The numerical value obtained by the first definition must be multiplied by 1000/g = 1.02 to give the values under the second definition, so that, for most practical purposes the two units are numerically identical. The first unit will be used in this Report, and called an ohm unless confusion with eleotrioal units is possible.

The impedance can be shown to consist of resistanoe, inductance and capacity (or compliance) analogous to the eleotrioal quantities, which gives rise to an overall impedance comprising 8 resistive and a reactive term. It is possible, if certain limits on dimensions are observed, to lump the impedance of some pneumatic and pneumo-mechanical items as simple resistances, inductances, or capacities. The calculation of the values for different pneumatic and mechanical oomponents, derived partly from Ref.6, is given in Appendix A, but we shall here state the forms for some of' the oomponents used in the experimental work to be described later in this Section and in Section 3.

A smooth-bore pipe of length $\ell$ om, radius r cm, where $\ell$ is short compared with the wave-length of sound at the relevant frequencies, may be treated as an inductance and resistance in series. The inductance is given by

$$L = (\ell\rho/\pi r^2) \text{ gm cm}^{-4} \text{ or Henry} \qquad (1)$$

where the gas density is $\rho$ gm $cm^{-3}$.

For air at normal temperature and pressure, the inductance of a pipe 1 metre long, radius 1 cm is 0.038 Henries.

The resistance is given by

$$R = (8\ \mu\ell/\pi r^4) \text{ ohm} \qquad (2)$$

where the dynamic viscosity of the ges is $\mu$ gm $cm^{-1}$ $s^{-1}$. For air at normal temperature and pressure in viscous flow the resistance of a pipe 1 metre long, radiua 1 cm is 0.047 ohm. Equation 2 holds so long as the flow is viscous, but end-effects in pipes, equivalent to the inclusion of orifices in the line, and irregularities in the bore, may make the flow turbulent, so that a more reliable definition of the resistanoe is

$$R = \text{pressure drop } (P)/\text{flow } (F) \tag{3}$$

measured for steady flow. The value of R will generally depend on the magnitude of the steady flow, but for small changes relative to the steady flow it may usually be treated as a constant.

A rigid container of simple form and of small dimensions compared with the wave-length, may be treated as a pure capacity. For a volume of' V $cm^3$, the capacitance is given by

$$C = (V/c^2\rho)\ gm^{-1}\ cm^4\ s^2 \text{ or Farad} \tag{4}$$

where the velocity of sound in the gas is c cm $s^{-1}$.

For air at normal temperature and pressure, the capacitance of a 1 litre container is 0.00072 Farad.

The above definitions can only be treated as constants if the pressure changes involved are small compared with the mean pressure, that is, if the changes in gas density are small, and it is only permissible to use lumped components in this say if the dimensions are small compared with the wavelength of sound in the gas at the relevant frequency. It is usually held that the dimensions should not exceed $\frac{1}{8}$ of the wavelength.

An assembly consisting of a pipe leading into a closed volume will obey the equation

$$L(dF/dt) + RF + \int (F/C)\ dt = P \tag{5}$$

If the pressure is oscillating sinusoidally at a frequency $f = \omega/2\pi$, $P = p_i e^{j\omega t}$,

$$(j\omega L + R + 1/j\omega C)\ F = p_i \tag{6}$$

so that the impedance of the assembly is given by

$$\begin{aligned} Z = X + jY &= j\omega L + R + 1/j\omega C \\ &= R + j(\omega L - 1/\omega C) \end{aligned} \tag{7}$$

That is,

$$\left.\begin{array}{l}\text{the resistive term } X = R \\ \text{and} \\ \text{the reactive term } Y = \omega L - 1/\omega C\end{array}\right\} \quad (8)$$

This circuit with the equivalent electrical circuit is shown in Fig.4(a).

## 2.2 Oxygen equipment and man as a control system

In order to understand how a man using breathing equipment can affect the stability of the system, we shall look briefly at the schematic control circuit shown in Fig.1(b). In the diagram, $Z_r$ is the impedance of the regulator, $Z_{s1}$ and $Z_{s2}$ form the impedance of the piping and mask, and $Z_m$ is the impedance the man presents to the equipment.

The man requires a volumetric flow $F_i$. He accordingly sucks on the system, which reduces the pressure at the regulator outlet to pr. The regulator responds by emitting a gas flow, $F_r$, obtained from the high pressure oxygen store at Q, if it is set for pure oxygen, or from a mixture of this store and ambient air if it is set for air-mix. $F_r$ is linked to the pressure at the outlet by the equation

$$F_r = N(p_d - p_r) \quad (9)$$

where $p_d$ is the datum pressure, and N is the transfer function of the regulator.

The flow is transmitted through the system to the man, and modified by it, so that he receives a flow $F_o$. His adaptive mechanism represented by G in the diagram, compares the flow he is receiving with what he requires, and he either sucks more or less hard, or modifies his flow pattern. At the same time, the flow of gas $F_r$ into the system will tend to raise the pressure at the regulator outlet, so that, in the absence of further excitation from the man, the demand will be reduced, and $F_r$ will decrease.

Thus, there are in effect two servo-mechanisms, the regulator controlling the pressure at its outlet, and the man controlling the flow he receives, the two linked by the system.

All the functions and impedances in the diagram arc *complex* and complicated. The function G is not *fully* understood, but is undoubtedly very involved[7]. The function N, for a very simple regulator, is derived from two second order, non-linear differential equations for continuity and mass balance in the demand chamber of the regulator[8].

However, to see how the impedance of system and man can affect stability, consider a pneumatic circuit with the regulator replaced by a simple reducing valve, having the transfer function

$$N = K\omega_o^2/(D^2 + 2h\omega_o D + \omega_o^2) \qquad (10)$$

where D is the operator (d/dt) which can be replaced by jw for a sinusoidal input of frequency $f = \omega/2\pi$, $\omega_o$ *is* the *circular* natural *frequency* of the valve, h is the damping coefficient, and K is a gain factor.

Let the valve feed a passive system of impedance

$$Z = X/(1 + DT) \qquad (11)$$

where T is the time constant of the system fed by the valve, and X tho resistance to steady flow.

Then, following Sivyer's work[9], the open loop equation is

$$p_r = NZ(p_d - p_r) \qquad (12)$$

end applying the Nyquist criterion, we can show that the mechanism will be stable if

$$(T\omega_o)^2 + 2h(T\omega_o) + 1 > KX(T\omega_o)/2h \qquad (13)$$

From this inequality we can draw two general conclusions.

(I) Large values of $\omega_o$, $T$ and h tend *to* promote stability,

(2) Large values of K and X tend to decrease stability.

But for rapid response we need h to be fairly small, and for good sensitivity we need K to be large, so that, even in this very simplified case, we need very large values of T, and very small values of X to be sure of stability.

Actual breathing equipment is too complex to be dealt with by this simple type of analysis, but since, even with this simplified form, the stability can be shown to depend on the pneumatic loading, it is plain that the more complex equipment will do so too. Dynamic testing with breathing simulators therefore will not reproduce instabilities unless the breathing simulator has the correct impedance for a man.

## 2.3 Principles of impedance measurement

Any method of measuring the dynamic characteristics of a pneumatic or pneumo-mechanical impedance requires a source of dynamic pressure or flow injecting either a sinusoidal wave or a step of pressure into the unknown impedance either in isolation or in combination with known impedances. It is necessary then either to measure the pressure drop across the unknown impedance and the flow into it, or else to compare the pressure drop across two parts of the circuit, one or both of which include the unknown impedance in combination with known impedances.

A sinusoidal input is generally preferred, compared with a step input, as in electrical measurements, because of the relative simplicity with which results may be analysed. There is also practical difficulty in producing a true step function of pressure in a pneumatic system. It was therefore decided to use a sinusoidal pressure generator as the power source.

Since the primary intention was to measure the impedance man presents to oxygen equipment, including the part of the face enclosed by the mask, it was necessary to use some system external to the man, terminating in a mask of the correct shape which could be sealed onto the face. It was thought at first that the best solution would be to use a typical aircraft system, and some attempts were made with an actual system of flexible piping leading to an aircrew-type mask complete with inspiratory and expiratory valves. It was found impossible, however, to obtain any consistency from such a system, owing to the flexibility of piping and mask which varied widely according to their mechanical support and the effect of the valves in varying the impedance between inspiration and expiration. It was therefore decided that the external system should terminate in a rigid valveless mask, and be connected by a moderately stiff pipe to the pressure source. It was found convenient to use a pipe of moderate length, since this helped in adapting the system to different subjects and also protected them to a certain extent from vibration and noise.

The alternatives then available for measuring the impcdanoe were:

(1) To measure the flow into the mask, and the pressure at the inlet or in the mask.

(2) To measure the pressure at two points in the external system.

It has already been explained in the introduction, that other experimenters have found difficulty in dynamic flow measurements at frequencies as low as 8 c/s, and it was intended to work up to 90 c/s. In any case, comparison of pressure and flow involves measurement of two factors with instruments which are unlikely to have equal responses over the full range of frequencies. Moreover, almost all methods of measuring flow involve the inclusion of additional pneumatic impedance in the circuit, which was thought to be undesirable as it would increase the work required of the subject in breathing.

It was therefore decided to use the second alternative, and compare the pressure within the mask with the pressure at the source as shown in Fig.4(b). Similar transducers, similarly connected to the system could then be used, so that their response was also similar. This method greatly simplifies calibration and analysis of results.

At an early stage of the investigation, it was thought that sufficient was known about the structure of the human respiratory system, to determine the impedance by amplitude measurements only, as recorded on a galvanometer recorder. That is, we should assume a circuit similar to that proposed by Shephard (shown in Fig.3) with *the* addition of a circuit for the face, and assess the values of the different components solely from the ratio of amplitudes of pressure at two points in the external circuit. This supposition, which has been described in detail in Ref.10, broke down for two main reasons, apart from the difficulty of solving the equations involved.

(I) Though the generator used produced a pure sine wave when fed into a transducer alone, the addition of non-linear elements distorted the wave form, especially at lower frequencies, so that it was extremely difficult to estimate amplitude accurately. Examples of some records are shown in Fig.6.

(2) It became obvious that our information was inadequate, that is, the model was too simple.

It was therefore decided to measure both amplitude ratio and phase shift between the two transducers using a Resolved Components Indicator, which compares only the fur&mental frequency, ignoring harmonics; and to *consider* the result as a single impedance, without reference to the sub-elements of which it must be composed.

2.4 Experimental equipment

A diagram of the experimental set-up is shown in Fig.4(b), (c) and (d), together with an electrical analogue, and a photograph of an experiment in progress is shown in Fig.5. The output of the pressure generator is fed through an adjustable restriction into the end of a reference system comprising a stiff tube leading to a rigid mask sealed onto the subject's face. A short length of pipe is connected to the other side of the source, to prevent complete loss of signal through the effective open circuit an open tube would present. Pressure transducers are connected near the source and into the mask, and their amplified outputs are fed in turn to the Resolved Components Indicator.

(1) Pressure generator

This apparatus, made by the Hymatic Engineering Co. Ltd., produces a sinusoidal pressure wave by means of the oscillation of a piston in a closed cylinder. The piston is driven through a scotch yoke and cam mounted on a shaft driven by a constant speed motor through an eleotronioally controlled magnetic clutch. The frequency range is 5-90 c/s, and the maximum pressure range is ±6 in w.g.

(2) External reference system

The external system consisted of a length of thick-walled rubber tubing, 54 cm long, of nominal inside diameter $\frac{1}{2}$ in, leading to a rigid mask of internal volume, when sealed to a dummy head, of 290 $cm^3$. In order to ensure that the same area of face was enclosed by the rigid mask as by a standard mask, the contour of contact was drawn on the dummy head, and the mask built up in fibre-glass from this line. A thin layer of rubber was then glued to the edge to form a seal. In some later tests a closed volume was used instead of the rigid mask, consisting of a cylinder of the same internal volume.

(3) Pressure transducers

The oscillating pressure was measured by means of variable inductance, differential transducers, type SE 70, made by SE Labs. Ltd., range ±5 psi. The oscillating pressure to be measured was only of the order of 1.5 cm w.g., but the quasi-static pressure was sometimes of the order of 5 cm w.g. It would have been preferable to use transducers of a range about ±25 cm w.g. (about 1/3 psi), but none were readily available which possessed good frequency response over the required range, and were of equally small size and weight - an important consideration, since one transducer was mounted on the mask. The output was fed into Type IT 1-6-51 amplifiers.

(4) Resolved components indicator

This equipment, made by Solartron Ltd., was used to compare the output of eaoh transducer in amplitude and phase, with the output of a synchro-resolver mounted on the shaft of the pressure generator. The instrument compares the pure sine-wave output of the synchro with the fundamental of the transducer output, so that any distortion is disregarded, The results obtained are two complex numbers $p_i$ and $p_o$, $p_i$ being the ratio of the output of the transducer at tie source to the synchro output, and $p_o$ the corresponding ratio for the mask-mounted transducer. The output of the R.C.I. was fed to a digital voltmeter, so that readings could be taken to 3 figures.

(5) Bicycle ergometer

This equipment, made by Zentralwerkstatt Gottingen GmbH, was used in some tests to enable the subjects to work up a heavy breathing rate. The work load for a subject pedalling on it can be adjusted by varying the position of a permanent magnet relative to a copper disc on the rear wheel, so that the subject has to work against the magnetic forces.

## 2.5 Preliminary tests

(I) Transducer calibration

The two transducers, connected to amplifiers which gave approximately the same sensitivity in static calibration, were joined to a T-piece the third arm of which was connected to the pressure generator, The full range of frequency (5-90 c/s) was then run through at several different amplitudes, and the outputs compared using the Resolved Components Indicator, which had been carefully adjusted according to the maker's instructions. It was found that the amplitude ratio was constant throughout the range, and the phase difference never exceeded $1^\circ$. This test was repeated at intervals throughout subsequent testing.

(2) Measurement of characteristics of reference system

The rigid mask was sealed onto a dummy head, the input transducer connected near the source and the output transducer to the mask cavity. The amplitude ratio and phase shift between the two transducers was determined for the range 5-90 c/s, with several different flow rates through the system. The amplitude ratio gave the familiar second order response curve, with a resonance at 45 c/s, agreeing very well with the figure calculated from the dimensions, and the damping factor varied from 0.03 to 0.05 for different flow rates.

The external system could thus be regarded as

$$\left.\begin{array}{ll}\text{an inductance} & L = 0.062 \text{ Henry} \\ \text{and capacitance} & C = 0.0002 \text{ Farad}\end{array}\right\} \qquad (14)$$

associated with negligible resistance.

A similar test was carried out with the same pipe and the closed volume, the latter being adjusted to give the same characteristics. The figures quoted above were used in all subsequent work, except in some tests at altitude, where the values of L and C are modified by the changed air density.

## 2.6 Derivation of impedance from experimental results

The impedance required was in every case paralleled with the capacitance of the mask or closed volume, as shown in Fig.4(b). The value of' the impedance at a frequency $f = \omega/2\pi$ is given by

$$Z = X + jY = (R + j\omega L)/[(p_i/p_o) - (1 - LC\omega^2 + jRC\omega)] \qquad (15)$$

where $(p_i/p_o)$ is the complex ratio of the outputs of the source and mask transducers. The detailed derivation of this equation and of the values for X and Y are given in Appendix B. Equation (15) is the basic equation used for all the experimental work.

## 2.7 Accuracy of impedance measurements

The basic accuracy of the ratio of the outputs of the pressure transducers when coupled to the Resolved Components Indicator is reckoned to be nearly as good as that of the R.C.I. itself, which is better than 2%. Some analysis of the probable errors in impedance which this might give rise to is given in Appendix C, where it is shown that the probable error depends not only on the error in the original readings, but also on the relative values of the resistive and reactive components of the impedance and. on the frequency. Based on the figure of 2%, calculations show that for most of the tests the probable error throughout the range of frequencies is of the order of ±0.5 ohm. Another possible source of error is a variation of the capacity in the mask due to different facial contours. In Appendix C it is shown that a possible error of 10% in the volume would produce a maximum error of about 0.5 ohm at the maximum frequency, with correspondingly less at lower frequencies, and this error would, of course, remain sensibly constant for any one subject.

Exceptions must be made for the measurements of the human impedance excluding the face with nose-breathing (see 3.2.2), where difficulties in sealing onto the nose caused wide scatter; and for the measurement of the face impedance (see 3.2.3), where the relative values of the resistance and reactance could cause quite large errors, especially at low frequencies. Fortunately, both these tests are of somewhat academic interest only.

Variations in ambient pressure, temperature and humidity lead to changes in air density and consequently to changes in the values of the reference impedance. The variation in gas density over the range of atmospheric pressure 740-780 mm Hg, temperature $14^{o}-26^{o}C$, humidity 0-100% is from 1.135 to 1.262 gm/litre. Hence, over this rather extreme range the density is 1.2 gm/litre ±0.064, and an error of 5% is possible. Thus, for the maximum values obtained for the human impedance is equivalent to about 0.3 ohm, so that errors from this source are not likely to be important.

Another apparent source of error was the variation between values obtained in inspiration and expiration with human subjects. While at frequencies above about 20 c/s, the readings on the R.C.I. were almost independent of whether the subject was breathing in or out, at lower frequencies there were quite large swings between the two conditions, that is, all four components fluctuated over quite a wide range. Attempts were made to measure either during inspiration or expiration, by making the subject inhale or exhale slowly over a long period, but the time available was still insufficient to obtain steady readings on the R.C.I., partly, it is believed, because the subject subconsciously varied his geometry during the period. In any case, such breathing is artificial, and may introduce effects which would not be present in normal breathing. Some records were therefore taken of the output of both transducers during complete braatning oyoles, using a galvanometer recorder with high paper speeds, so that records with about 1 in/cycle of the oscillating pressure were obtained. Part of a record is shown in Fig.6. It will be seen, that, though the traces are too complex to determine the actual amplitude ratio and phase shift without using Fourier analysis methods, the relations between the input and output remain unchanged, though the amplitude *is less in expiration* than in inspiration. It was therefore concluded that it was legitimate to use the mean values observed on the R.C.I. at these frequencies, without loss of accuracy, though there is obviously error in judging the mean values.

Another source of variation in measurements of human impedance is variation in the individual himself, That is, the subject *may* tighten or relax his facial muscles, or his stomach muscles, or shift the mask on his face, or vary the depth of his breathing, or the width of his mouth opening. All these effects, though they will tend to increase the scatter between results, cannot be classified as errors, *since* they represent possible conditions.

We may therefore conclude that for most of the tests, the error was unlikely to exceed 1 ohm, with a probable mean error not greatly in excess of

0.5 ohm. Errors at the lower frequencies were somewhat greater, owing to the difficulty of judging mean values.

## 3 TESTS WITH HUMAN SUBJECTS

### 3.1 Range of tests and experimental procedure

The tests conducted can be divided into three groups. The first Croup, oomprising tests A, B and C in Table 1 (below), were direoted to determining the impedance of the respiratory system and face, for light and heavy, nose and mouth breathing, and the effect on it of reduced gas density, and increased resistance to breathing.

In the second group, Tests D and E, the face was by-passed. These results might be compared directly wich published work.

The third group, Tests F and G were of academic interest only.

The tests are summarized in Table 1.

Table 1

Range of tests with human subjects

| Test No. | External system | Type of breathing | No. of subjeots | Impedance determined |
|---|---|---|---|---|
| A1 | 1 | Mouth/light | 6 | Respiratory system with face, at ground level |
| 2 | 1 | Mouth/heavy | 3 | |
| 3 | 1 | Nose/light | 6 | |
| B1 | 1 | Mouth/light | 1 | Respiratory system with face, at 8000 ft |
| 2 | 1 | Nose/light | 1 | |
| C | 1 | Nose/light | 3 | Respiratory system with face, and high external resistance |
| D 1 | 2 | Mouth/light | 9 | Respiratory system exoluding face, at ground level |
| 2 | 2 | Nose/light | 7 | |
| E | 2 | Mouth/light | 1 | Respiratory system excluding face, at 8000 ft |
| F | 1 | - | 3 | Face only |
| G | 2 | Nose/light | 5 | Mouth cavity |

Notes to Table 1: For tests A. B, C, F see Fig.4(b)
For tests D 1, E sea Fig.4(c)
For tests D 2, see Fig.4(d)

External system 1 means the rigid mask and pipe.

External system 2 means the closed volume and pipe.

For light breathing the subject was seated comfortably and breathed at an easy rate. The peak flow was about 15 l/min.

For heavy breathing the subject peddled on a bicycle ergometer until his breathing rate reached approximately 35 l/min as indicated by a flowmeter downstream of the pressure source.

In tests A3, B2, C, D2, nose breathing, the subject kept his mouth dosed.

In tests Al, A2, B1, D1 and E, mouth breathing, his nostrils were plugged with cotton wool.

In tests D and E, mouth breathing, the subjeot closed his lips round a 1 in outside diameter short pipe leading to the closed volume.

In test D2 he fitted his nose into a special end-plate to the closed volume keeping his mouth closed.

Tests B1, B2 and E were conducted in a decompression chamber.

In all the tests except C the static resistance to breathing was about 1 in w.g. at a flow rate of 30 l/min (about 5 ohm). In test C this was increased to about 10 ohm.

In test F, the subject held his breath for the period necessary to take the readings - about 30 seconds.

In test G, the subject placed his mouth over the outlet pipe of the closed volume, breathing through his nose and sealing off the mouth cavity with his tongue.

The subjects used throughout the tests were healthy men in the age range 18-45 years, all technicians or B.A.F. Officers working in the Human Engineering Division of Mechanical Engineering Dept.

The tests A and D were carried out with as many subjeots as could conveniently be fitted in. The other tests were carried out on fewer numbers. Only one subject (G F R) was used in every type of test. No-one with a cold was tested since aircrew are usually grounded when they have colds, but otherwise no enquiry was made into the subjects' fitness at the time they were tested.

In every test, the subjeot was first familiarized with the apparatus and the type of experiment to be undertaken, and, in tests using the rigid mask, shown how to oheok that the mask was leak-tight. This was done by blocking the open end of the system, and allowing the subject to exhale into it. Any leaks caused a flow of air past the mask seal over the face, which was easily detected by the subject. The system was passed when no leak occurred for a pressure of 5 in water.

Readings of the inphase and out-of-phase components of input and output pressures relative to the synchro output ware taken at 5 c/s intervals in the range 5-90 c/s, with an input amplitude of approximately ±2 cm w.g., maintained by adjusting the restriction between *the pressure generator* and the system. After each set of readings at one frequency, the subject **removed** the mask and breathed ambient air for a few minutes so that re-breathing the $CO_2$-rich sir of his exhalate should not over-stimulate his breathing. Each test was repeated three times with each subject. The derivation of impedance was made on a digital computer along the lines described in Appendix B. The average impedance for the three runs for each subject and condition was determined, rejecting only a few gross inconsistencies, and plotted, and from these curves the results and scatter-bands shown in Figs.7-10 were plotted.

The range of tests does not include all the oases possible in aircraft, in particular heavier breathing rates than those considered are possible, and only very brief tests at altitude and with increased external resistance to breathing werecarried out. It is hoped to extend the tests to cover these cases more thoroughly at a later date, but it is felt that within the time available the present tests cover a significant range of conditions.

### 3.2 Results and discussion

It may be as well, before embarking on detailed analysis of the results obtained, to re-state the primary object of the investigation, and the limitations of the experimental method.

The object of the most important tests - those using the rigid mask - was to determine the range of impedance whioh might be present to aircraft oxygen equipment in ordinary use. Therefore, it was important not to impose artificial restriotions on the subjects, and no attempt was made to standardize some of the variables which could lead to different impedanoes. For example, in the mouth-breathing tests with the mask, the subject was allowed to keep his mouth as wide open as he preferred. The light breathing rate was not carefully controlled, and neither was the subject's posture which might affect impedance by modifying the characteristics of chest walls and diaphragm. Also, since the subjeots were not medically examined prior to test, some may, at one time or another have been suffering *from* slight catarrh or incipient colds. Hence we should expeot fairly wide divergence between subjects, and between repeated tests on the same subject. One variable was eliminated from the mouth-breathing tests with the volume, in that the subject had to fit his lips round a standard cutlet pipe. We should therefore expect greater consistency from these tests, and this proved to be the case, as is seen by comparison of' the results shown in Fig.7(a) and 8(a).

The scatter-bands shown coincide very nearly with the curves for ±2 standard deviations from the mean, and might therefore be expected to cover 95% of a population of healthy young men not suffering from colds.

However, correlation could not be expected between certain tests which might appear to yield comparative results. It might be supposed that the face impedance (Test F) could be combined witn the results for mouth breathing cn the closed volume (Test D1) to give the same values as the tests for mouth-breathing with the mask (Test Al). But since the mouth opening was not controlled in Test Al, and was almost certainly less than in Test D1, no such correlation is possible.

3.2.1 Impedance of human respiratory system including the face

Some general conolusicns can be drawn at a glance from the scatter bands of Fig.7.

(I) Resistive component at ground level, for low resistance to breathing

In all three cases the resistance level remains fairly constant throughout the frequency range, though with a dip in the neighbourhood of 20 c/s in light breathing. The mean value is greatest for nose-breathing, and least for heavy mouth-breathing, a difference which may reasonably be ascribed to the difference in area between the nasal passages, the mouth slightly open for light mouth-breathing and more widely open for heavy mouth-breathing. The full range of resistance is between 1.7 ohm and 7.0 ohm.

(2) Reactive component at ground level for low resistance to breathing

The reactive component is mainly inductive in all three cases. This is a little surprising at first glance, since it would appear likely on the face of it that the volume of the lung, giving a capacative effect, would be the predominating factor. It seems from the results that the dominant effect (in the range of frequencies considered) is due to the inertia of the gas in the airways and mechanical inertia of the tissues. Table 2 shows the approximate values of simple inductance and capacitance in series which would cover the limits of the scatter in the three cases, for frequencies greater than 10 c/s.

The results at 5 c/s in light breathing lie outside this band, and would be regarded as doubtful if they were not so consistent with each other.

Table 2

Reactive components of respiratory system including face

| Test | Lower limit | | Upper limit | |
|---|---|---|---|---|
| | Inductance (Henries) | Capacitance (Farads) | Inductance (Henries) | Capacitance (Farads) |
| Al Mouth/light | 0.0076 | very large | 0.013 | very large |
| A2 Mouth/heavy | 0.0084 | 0.048 | 0.035 | very large |
| A3 Nose/light | 0.0098 | 0.024 | 0.013 | 0.081 |

The reactive component of impedance Y, is here given by $Y = (2\pi f L - 1/2\pi f C)$, so that the larger the value of C the less importance it has in the combined impedance.

In all case* except the upper limit of A2, the figures quoted in Table 2 would give curves close to the limits, but in this case the large value of inductance is required to fit the values of impedance in the neighbourhood of 5-20 c/s, and much lower values would fit the curve at higher frequencies,

It is not suggested that the impedance actually consists of a pips and volume having these values, indeed, suoh a thing is physically impossible, since the capacities given are equivalent to volumes up to more than 100 litres. These are simply figures whioh might be used in constructing analogues.

Moreover, looking at the results for an individual, denoted by (x) in Fig.7, we see that though the points all lie within the aoattrr band, the actual shapes differ widely from the shape of the limits, and are obviously of a more oomplicated form. Thus, while an analogue to represent an individual throughout the full frequency range would be very complicated, a range of comparatively simple adjustable analogues might be constructed to cover the scatter bands, the individual being represented at different frequencies by different settings.

(3) Effect of increased resistance to nose-breathing

The results for subjects breathing against a higher resistance show a wider range of resistance and reactance. The results plotted for one individual, denoted by o in Fig.-/(c), show an increase of resistance, but for other subjects a decrease was observed. Similarly, the capacitive effect was increased with this subject but not for all subjects. The chief effect then seems to be to produce a larger scatter between individuals, possibly owing to their different methods of achieving the increased effort required. The upper limit of resistance is raised a little, in this case, to 7.5 ohm. The reactive upper limit is hardly altered, but the lower limit can be represented by an inductance of 0.0091 Henry and a capacitance o? 0.C11 Farad.

(4) Effect of decreased air density

The density of sir at 8000 ft is approximately 0.8 times the density at ground level. We should therefore expect, if the reactance were due to pneumatic effects, since both L and 1/C are proportional to the density, that the reactance would decrease by 20% at 8000 ft compared with the values at ground level. The results, denoted by . in Fig.7(a) and 7(c), show that this is not so, in fact, on the whole, where it is possible to observe a significant difference, the values at altitude are somewhat higher than those at ground level, except possibly in the neighbourhood of 50 c/s in nose-breathing, and at low frequencies in mouth-breathing. This suggests that the reactance is mainly due to pneumo-mechanical rather than pure pneumatic effects.

The resistance when mouth-breathing shows very little difference from the results at ground level, but in nose-breathing there was an average decrease, of about 20% over the whole frequency range. If the resistance were due to mechanical or viscous pneumatic effects, we should expect it to be unaltered by gas density; if it were due to orifice effects, we should expect it to be proportionalto the gas density. This latter case is approximately true for nose-breathing, suggesting that the nose acts as an orifice, or series of orifices, while flow through the mouth is more nearly viscous - not an unreasonable conclusion, considering the relative size of the openings.

It is recognized, however, that the differences to be expected are too small for firm conclusions to be drawn from the limited data, since only one subjeat was tested at altitude, and it is possible that his physical condition might have varied between the tests.

3.2.2 Impedance of human respiratory system excluding the face

Comparison of Figs.7(a) and 8(a) shows tiiat the resistance is somewhat lower when breathing through the pipe than when wearing the mask, but the reactive oomponent is very little affected. This suggests that the face has little effect on the impedance, the difference in resistance probably being due to the rather wider mouth opening enforced by the pipe diameter.

The resistance is in fact very similar to that measured in heavy breathing when wearing the mask.

Comparing 7(c) and 8(b), the differences in nose-breathing appear more marked. But great difficulty was experienced in sealing onto the nose, and if a good seal was achieved, it was generally at the expense of pinching the nostrils. These difficulties probably account for the very wide scatter, and the higher mean values of resistance and reactance, since any decrease in the cross-sectional area of the nasal airways would increase both proportionately.

The resistance range *in* mouth-breathing was from 1-4 ohm. The lower and upper limits of reactance can be matched to an inductance of 0.0098 Henry ad a capacitance of 0.063 Farad for the lower limit; and an inductance of 0.014 Henry for the upper limit.

Comparison with published data

In comparing the results obtained in these tests with results obtained by other experimenters, it must be borne in mind that all figures for resistance, inductance and capacitance are attempts tc find simple forms to fit the much more complicated respiratory system, so that the figures will refer in general only to specific frequency bands. Hence, unless tests covering the same frequency bands are compared, we should not expect close agreement for inductance and capacitance. Since the resistance is likely to be more independent of frequency, we should expect better agreement here.

From the figures given by Dubois et al[2], we can extract the mean values quoted, R = 4.6 ohm, L = 0.042 Henry, C = 0.018 Farad. These averages appear to be based on the lower end of the results shown, at frequencies up to about 6 c/s, and agreement with the experimental results is less good above about 10 c/s. The resistance is very close to the mean of our results, the inductive term is higher than any of ours, the capacitive term is much lower. But these figures probably apply to a frequency range lower than we investigated.

McIlroy et al[3] give values *obtained* from 12 healthy subjects, with a range from 2.7 ohm to 9.0 ohm in resistance, and a mean value of 5.26 ohm; and a range from 0.035 Farad to 0.13 Farad in capacitance, with a mean value of 0.087 Farad. The method used, relaxed expiration with steady flow rates, is obviously applicable for very low frequencies, and excludes inductance. The method is perhaps open to objection in that the breathing is in expiration only, and somewhat artificial; and that the values are obtained by comparing time constants with two different external resistances, and we have shown that increased external resistance may alter the impedance.

However, once again, the resistive measurement is near our mean. Both Dubois and McIlroy refer to mouth-breathing.

The results obtained from Shephard's analogue[5] and shown in Fig.3, again give very reasonable agreement on the resistive side for both nose and mouth-breathing, and comparison with the individual results in Fig.7 for the reactive component suggest that the type of circuit given for nose-breathing is correct, though the turning points do not match very well with those observed here.

On the whole then, the published results and ours agree fairly closely as to the resistive component of impedance, but less well as to reactance. Since, however, the frequency ranges covered are different in each case, we should expect considerable differences, and the results are probably not Incompatible.

3.2.3 Impedance of face

As has been shown in 3.2.2, the inclusion of the face makes little difference to the human respiratory impedance, and this is confirmed, to a oertain extent, by the results for the face impedance shown in I'ig.9. We cannot expect perfect correlation between the figures for mouth-breathing with the mask, and the paralleled figures for the face, and for mouth-breathing through a pipe, as explained in 3.2.2, however, paralleling the mean values from Fig.9 and Fig.8(a), we find that the effect is to reduce the values in 8(a) by less than 0.5 ohm in resistance, throughout the range, and the reactance by about 2 ohm at 90 c/s, diminishing to 0.2 ohm at 5 c/s, which is not very far from the actual effect, allowing for errors in the measurement of face impedance. As explained in detail in Appendix C, the probable errors in the face impedance at low frequency were very large, least at about 15 c/s, but increasing again to a probable error of about ±1.5 ohm at 90 o/s.

The mean value for the face reactance is equivalent to 0.009 Henry and 0.004 Farad, as shown in the dotted line in Fig.9; but a variable resistance would be needed for the resistive component.

3.2.4 Impedance of mouth cavity

The results, shown in Fig.10, give a resistance that varies widely with frequency, but the reactance is effectively a simple oombination of inductance and capacitance. The dotted line is the calculated value for L = 0.0223 Henry and C = 0.004 Farad.

## 4 DYNAMIC TESTING OF OXYGEN EQUIPMENT

It is not intended here to discuss in detail the construction of suitable simulators, or the effects of such simulators in testing. This subject is covered in detail in Ref.10. We shall only illustrate the importance of correct simulation by describing the behaviour of one present-day simulator in conjunction with one current aircrew oxygen system, and suggest methods of improved simulation.

The simulator is the British Oxygen Co's "Beaver" Mk.II respirator, used *at present* for routine testing of regulators. The system is the Normalair seated-mounted system, consisting of about 1 ft of $\frac{1}{2}$" diameter pipe, 2 ft of $\frac{3}{4}$" diameter pipe, with a quick-release connector, together with the flexible mask-tube and mask. This system is in the middle range of *size* of systems in current use.

In all tine tests to be described, the equipment used was the same as in the tests using humans, the impedance to be measured being connected to the outlet of the closed volume.

### 4.1 'Beaver" impedance

In normal use, the "Beaver" has two *different* settings, for nominal 'light' breathing with a peak flow of 30 l/min, and nominal 'heavy' breathing, with a peak flow of 110 l/min, and it has not hitherto been considered necessary to standardize the pipe-work connecting the "Beaver" to the system under test. Tests were therefore carried out at both settings, both with the shortest possible pipe connector and also with a 4 ft length of flexible piping. The results obtained are shown in Fig.11. The results show that the impedance of the "Beaver" varies widely according to the connector used, as might be expected; and to a less extent according to the setting when a long connector is used. But even with the minimum connector, the impedance of the "Beaver" is, for most frequeroies several times greater than the human impedance.

### 4.2 Comparison of impedance of' man and a typical system, with that of' the "Beaver" and the same system

In order that the impedance the system presents in normal inspiration should appear over the whole breathing cycle, the inspiratory valve of the mask was removed and the expiratory valve blocked. Measurements were made with twc subjects both for mouth and nose light breathing. The scatter bands obtained are shown in Fig.12.

The mask was then mounted on a standard mask clamp, attached to the "Beaver" by a short connector, and the impedance measured for both 'light' and 'heavy' breathing. These results are also shown in Fig.12.

It will be observed that there are very few frequency ranges in which the "Beaver" results lie within the scatter bard for the men for both resistance and reactance. With the light breathing setting, the match is fairly good in the neighbourhood of 50 c/s, and again above 80 c/s; for the 'heavy' breathing, there is only a possible match at about 65 c/s.

It is rather surprising, considering the difference shown to exist between man and the "Beaver", that there should be even so much agreement; but it has to be remembered that the system is itself modified in attaching it to a simulator, since the mask has to be fitted onto a clamp, thus probably varying its characteristics.

Thus, with this particular type of system, tested in this manner, any instability discovered except in a few restricted ranges must be viewed with suspicion. And the fact that no instability occurs in those ranges, does *not* mean that the system is stable even there, since the "Beaver" here represents only one possible human impedance.

It is probable that the effect of the man, or simulator, on the larger systems in current use would be less than on the seat-mounted system, and that the effect would be greater on miniature man-mounted systems. This subject is explored further in Ref.10.

However, it has been demonstrated that there can be significant differences due to simulator connectors and mask clamps, and that the "Beaver" is inadequate to represent man when a question of stability arises, although its usefulness in more routine dynamic investigations has been fully proved (see Ref.11). There is an obvious need of a simulator with more representative mask-fixing, and whose impedance can be varied over the range likely to occur in a user.

## 4.3 Simulation of human impedance for dynamic testing

All the components of the human impedance may be simulated either pneumatically or mechanically, or by a combination of both types of element. The choice is largely a matter of convenience, and depends on the ability to combine the configuration with a simple method of producing the necessary flows. One possible solution will be briefly described. Others may be found in Ref.10.

The resistive oomponent may be produced as a pneumatic resistance, by means of an arrangement of gauzes such as that shown in Fig.13. In this scheme, a variable resistance is conveniently combined with a fixed resistance by means of which the flow may be measured. The graph shown is for 400 mesh phosphor-bronze gauzes, diameter 2 in.

The inductive component may be obtained pneumatically by pipes of convenient dimensions, as can be seen by comparing equation (1) with the values required in Table 2.

The capacitive component is more difficult to simulate with a pure element, since it demands very large volumes, whose dimensions would exceed the allowable limits to which simple theory could be applied. It is therefore likely that the capacitive effect can best be achieved mechanically.

The arrangement shown in Fig.14 offers a possible solution. The flow is generated by the piston moving in the cylinder, the resistance and inductance are incorporated in the outlet pipe. The capacitance is a combination of the pneumatic effect of the volume of the cylinder, and the spring-rate of the light bellows closing it. If for example, the cylinder has a volume of 4 litre, then by equation (4) its capacitance is approximately 0.003 Farad. Then, to produce the capacitance necessary for, say, light nose-breathing (0.081 Farad from Table 2), the capacity of the bellows must be 0.078 Farad. Using equation (19) of Appendix A, for a circular bellows of diameter 20 om, we need a spring-rate given by $k = A^2/0.078 = 1.28 \times 10^6$ dynes/cm. This is equivalent to a pressure of $3.2 \times 10^3$ dynes/cm$^2$ (or 3.2 cm w.g.) within the cylinder for 1 am displacement, which appears to be an attainable value.

## 5 CONCLUSION3

A method of measuring the pneumatic impedance of the human respiratory system over a wide range of conditions (nose, mouth, heavy and light breathing) and fairly large frequency range has been demonstrated. Normal breathing is used, and no artificial breathing techniques are required.

The impedance man presents to oxygen equipment has been measured over the frequency range 5-90 c/s.

Tne resistive component remains fairly constant over the frequency range, varying from about 1 ohm (gm om$^{-4}$ s$^{-1}$) to 7 ohm according to the conditions, being greatest for nose-breathing, and least for breathing with the mouth widely opened. Hence, the resistance tends to be lowest in heavy breathing. The density of the gas breathed appears to have little effect on the resistance in mouth-breathing, but reduces the resistance in nose-breathing proportionately to the density: but this requires further confirmation.

The reactive component is chiefly inductive over the frequency range, lying for the most part between 0.007 and 0.043 Henry ($gm\ cm^{-4}$), though for heavy breathing in the lower part of the frequency range the value may be as high as 0.035 Henry. The capacitance has little effect, except at the lowest frequencies, and may be represented by values greater than 0.024 Farad ($gm^{-1}\ cm^{4}\ s^{2}$). There is little difference between the values for nose and mouth breathing, and the effect of variation of gas density is small. This, and the large volumes which would be required to represent the capacitance with rigid pneumatic components, suggest that the reactance is mainly due to pneumo-mechanical effects.

The importance of standardized simulators in investigating the stability of oxygen equipment, and the inadequacy of one present-day simulator to represent man for this purpose, have been demonstrated. Suggestions for an improved simulator have been made.

There is an urgent need for a simulator which will truly represent man's impedance, as well as producing the required flow patterns, and it is thought that the data given here will be useful in assisting the design and development of such a simulator,

The data given apply only to healthy young men, and the range of breathing conditions is not fully comprehensive. It in hoped at a later date to fill in some of the gaps, and also to extend the measurements to higher frequency ranges, and to give more detailed cover to the lower frequencies.

The application of the method to clinical work would obviously need great caution in view of the scatter observed between individuals, and the variation in impedance of the same individual in different conditions. It is thought that unless detailed observation of an individual subject could be carried out, only gross pathological effects are likely to be detected by the method.

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# Appendix A

## IMPEDANCE OF SOME BASIC ELEMENTS OF PNEUMATIC AND PNEUMO-MECHANICAL SYSTEMS

In this Appendix the derivation of the dynamic constants will be given in simple terms which can be applied to the dimensions and frequencies used in the experimental work. For more detailed analysis applicable to wider ranges of dimensions and frequencies, the reader is referred to works on acoustics, or to detailed studies of particular elements, such as Ref.6.

Throughout the analysis it is assumed that the pressure changes are small compared with the mean pressure, and that all the dimensions are small compared with the wavelength of sound in the gas for the appropriate frequencies.

We shall, in each case, produce an equation connecting the Plow, F $cm^3/s$, with the pressure P $dynes/cm^2$, of the form

$$L\, dF/dt + RF + (1/C) \int F\, dt = P \qquad (A1)$$

where L, the coefficient of $dF/dt$, represents inductance ($gm/cm^4$)

R, the coefficient of F, represents resistance ($gm/s\ cm^4$)

C, the reciprocal of the coefficient of $\int F\, dt$, represents capacitance ($cm^4\ s^2/gm$).

### Pneumatic elements

### Impedance of a pipe

Consider an element of length $\ell$ cm of a pipe, radius r cm, with a pressure difference P $dynes/cm^2$ across it. The gas is flowing at a velocity v cm/s, so that

$$F = v\pi r^2$$

### Inductance

The mass of gas in the element is $\rho\ell\pi r^2$, so that the acceleration resulting from the pressure difference is given by

$$\rho\ell\pi r^2\ (dv/dt) = P\pi r^2 \qquad (A2)$$

that is,

$$(\rho\ell/\pi r^2)\ (dF/dt) = P \qquad (A3)$$

The reactive component is chiefly inductive over the frequency range, lying for the most part between 0.007 and 0.013 Henry ($gm\ cm^{-4}$), though for heavy breathing in the lower part of the frequency range the value may be as high as 0.035 Henry. The capacitance has little effect, except at the lowest frequencies, and may be represented by values greater than 0.024 Farad ($gm^{-1}\ cm^{4}\ s^{2}$). There is little difference between the values for nose and mouth breathing, and the effect of variation of gas density is small. This, and the large volumes which would be required to represent the capacitance with rigid pneumatic components, suggest that the reactance is mainly due to pneumo-mechanical effects.

The importance of standardized simulators in investigating the stability of oxygen equipment, and the inadequacy of one present-day simulator to represent man for this purpose, have been demonstrated. Suggestions for an improved simulator have been made.

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# Appendix A

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Throughout the analysis it is assumed that the pressure changes are small compared with the mean pressure, and that all the dimensions are small compared with the wavelength of sound in the gas for the appropriate frequencies.

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The mass of gas in the element is $\rho\ell\pi r^2$, so that the acceleration resulting from the pressure difference is given by

$$\rho\ell\pi r^2\,(dv/dt) = P\pi r^2 \qquad (A2)$$

that is,

$$(\rho\ell/\pi r^2)\,(dF/dt) = P \qquad (A3)$$

Hence, the inductance, L, is given by

$$L = \rho\ell/\pi r^2 \tag{A4}$$

This formula only applies strictly for parts of the pipe distant from the ends. To allow for end effects it has been suggested[12] that an effective pipe-length $\ell' = 4\ell/3$ should be used.

Resistance

If it is assumed that the flow of gas in the pipe is viscous, the relation between a constant flow and constant pressure drop, as given in works on aerodynamics is

$$(8\mu\ell/\pi r^4)\ F = P \tag{A5}$$

Hence, the resistance, R, is given by

$$R = 8\mu\ell/\pi r^4 \tag{A6}$$

But this law only holds for parts of long straight smooth bore pipes distant from the ends. Roughness of the pipe-bore, bends in the pipe, end-effects, or high gas velocity may all tend to produce turbulent flow, and consequent increased resistance. The inclusion of an orifice, for example, where the flow is governed by a formula of the form

$$F^2 = P \times \text{constant} \tag{A7}$$

gives a resistance proportional to the gas flow.

However, the pipe used for the reference impedance in the experimental work was found to have resistance almost independent of flow. The numerical value was however, approximately 10 times the value calculated from equation (A6), though still so low as to be negligible in subsequent calculation.

Impedance of simple volume

Consider a simple container of volume V $cm^3$, with internal pressure P dynes/$cm^2$, and gas flowing into it at a rate F $cm^3/s$. Then the pressure and density of the gas are connected by the formula

$$P/\rho^n = \text{constant}$$

where $n = 1$ for isothermal. changes; and

$n = y$, the ratio of specific heats, for adiabatic changes;

$\gamma = 1.4$, for air.

Differentiating,

$$d\rho/dt = (\rho/nP)\ dP/dt \qquad (A8)$$

The mass of gas in the container is $\rho V$, so that

$$d(\rho V)/dt = F \qquad (A9)$$

that is,

$$(V/nP)\ (dP/dt) = F \qquad (A10)$$

or,

$$(nP/V) \int F\ dt = P \qquad (A11)$$

For isothermal changes, therefore,

$$C = V/P \qquad (A12)$$

For adiabatic changes,

$$C = V/\gamma P \qquad (A13)$$

or, since the speed of sound is

$$c = (\gamma P/\rho)^{\frac{1}{2}}\ ,$$

$$C = V/\rho c^2 \qquad (A14)$$

Extension of this formula to a pipe of moderate bore, where the capacity term cannot be neglected, shows that for a length $\ell$, radius $r$,

$$C = \pi \ell r^2/\rho c^2 \qquad (A15)$$

Mechanical elements

Impedance of bellows system

Consider a bellows of stiffness $k$ dynes/cm, damping force $q$ dynes s/cm, carrying a rigid diaphragm of area A $cm^2$ and mass W gm, with a pressure difference P dynes/$cm^2$ between the inside and outside.

Then the diaphragm displacement is given by

$$W\, d^2x/dt^2 + q\ ax/at + kx = PA \qquad (A16)$$

The flow of gas involved in the diaphragm displacement is

$$F = A\,dx/dt \tag{A17}$$

so that

$$(W/A^2)\,dF/dt + (q/A^2)\,F + (k/A^2)\int F\,dt = P \tag{A18}$$

Hence, for this mechanical system

$$\left.\begin{aligned} L &= W/A^2 \\ R &= q/A^2 \\ C &= A^2/k \end{aligned}\right\} \tag{A19}$$

Thus for a weight of 1 kg on a spring of stiffness $10^6$ dynes/cm,

$$\left.\begin{aligned} L &= 0.01 \text{ Henry} \\ C &= 0.1 \text{ Farad} \end{aligned}\right\}$$

Similar forms apply to simple flexible diaphragms, except that the flow and the effective mass W', depend on the shape of deformation of the diaphragm, so that

$$F = A'\,dx/dt \quad , \quad \text{say} \tag{A20}$$

Therefore,

$$\left.\begin{aligned} L &= W'/AA' \\ R &= q/AA' \\ C &= AA'/k \end{aligned}\right\} \tag{A21}$$

Two particular cases will be considered. The formulae for diaphragm deflection and natural frequency are taken from Ref.12.

Stretched circular membrane

Consider a stretched membrane of radius b cm, thickness t cm, of material of density d gm/cm$^3$, with peripheral stress s dynes/cm. The membrane deflection under pressure is spherical, hence, if the centre deflects a short distance x, the volume of gas displaced is $\pi x b^2/2$.

Therefore

$$A' = A/2 \tag{A22}$$

The displacement

$$x = b^2 P/4S \tag{A23}$$

so that

$$k = 4\pi S \tag{A24}$$

The lowest natural frequency is given by

$$\omega_{o_1}^2 = (2.45/b)^2 S/td \tag{A25}$$

and since

$$\omega_{o_1}^2 = k/V' \quad , \quad V' = 4\pi\, b^2\, td/2.45^2 = 0.67\ Atd \tag{A26}$$

Hence

$$\left.\begin{aligned} L &= V'/AA' = 1.33\ td/A \\ C &= AA'/k = A^2/25.1\ S \end{aligned}\right\} \tag{A27}$$

Built-in diaphragm

For a built-in circular diaphragm of the same dimensions, made of material with Young's modulus $E_o$ dynes/cm$^2$, and Poisson's ratio $\nu$, the deflection at a distance $r_1$ from the centre of the diaphragm when the centre deflects a distance x, is $x(b^2 - r_1^2)^2/b^4$, so that the volume of gas displaced is $\pi b^2 x/3$.

Therefore

$$A' = A/3 \tag{A28}$$

The displacement

$$x = 3(1 - \nu^2)\, b^4\, P/16\, E_o\, t^3 \tag{A29}$$

so that

$$k = 16\pi\, E_o\, t^3/3(1 - \nu^2)\, b^2 \tag{A30}$$

The lowest natural frequency is given by

$$\omega_{o_2}^2 = (5.12t/b^2)^2 E_o/3d\ (1 - \nu^2) \tag{A31}$$

The flow of gas involved in the diaphragm displacement is

$$F = A\,dx/dt \tag{A17}$$

so that

$$(W/A^2)\,dF/dt + (q/A^2)\,F + (k/A^2)\int F\,dt = P \tag{A18}$$

Hence, For this mechanical system

$$\left.\begin{aligned} L &= W/A^2 \\ R &= q/A^2 \\ C &= A^2/k \end{aligned}\right\} \tag{A19}$$

Thus for a weight of 1 kg on a spring of stiffness $10^6$ dynes/cm,

$$\left.\begin{aligned} L &= 0.01 \text{ Henry} \\ C &= 0.1 \text{ Farad} \end{aligned}\right\}$$

Similar forms apply to simple flexible diaphragms, except that the flow and the effective mass W', depend on the shape of deformation of the diaphragm, so that

$$F = A'\ dx/dt\ ,\ \text{say} \tag{A20}$$

Therefore,

$$\left.\begin{aligned} L &= W'/AA' \\ R &= q/AA' \\ C &= AA'/k \end{aligned}\right\} \tag{A21}$$

Two particular cases will be considered. The formulae for diaphragm deflection and natural frequency are taken from Ref.12.

<u>Stretched circular membrane</u>

Consider a stretched membrane of radius b cm, thickness t cm, of material of density d gm/cm$^3$, with peripheral stress s dynes/cm. The membrane deflection under pressure is spherical, hence, if the centre deflects a short distance x, the volume of gas displaced is $\pi x b^2/2$.

Therefore

$$A' = A/2 \tag{A22}$$

The displacement

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so that

$$k = 4\pi S \tag{A24}$$

The lowest natural frequency is given by

$$\omega_{o_1}^2 = (2.45/b)^2 \; S/td \tag{A25}$$

and. since

$$\omega_{o_1}^2 = k/W' , \quad W' = 4\pi \, b^2 \; td/2.45^2 = 0.67 \; Atd \tag{A26}$$

Hence

$$\left.\begin{aligned} L &= W'/AA' = 1.33 \; td/A \\ C &= AA'/k = A^2/25.1 \; S \end{aligned}\right\} \tag{A27}$$

Built-in diaphragm

For a built-in circular diaphragm of the same dimensions, made of material with Young's modulus $E_o$ dynes/cm$^2$, and Poisson's ratio $\nu$, the deflection at a distance $r_1$ from the centre of the diaphragm when the centre deflects a distance $x$, is $x(b^2 - r_1^2)^2/b^4$, so that the volume of gas displaced is $\pi b^2 x/3$.

Therefore

$$A' = A/3 \tag{A28}$$

The displacement

$$x = 3(1 - \nu^2) \, b^4 \, P/16 \, E_o \, t^3 \tag{A29}$$

so that

$$k = 16\pi \, E_o \; t^3/3(1 - \nu^2) \; b^2 \tag{A30}$$

The lowest natural frequency is given by

$$\omega_{o_2}^2 = (5.12t/b^2)^2 \, E_o/3d \; (1 - \nu^2) \tag{A31}$$

and since

$$\omega_{o_2}^2 = k/W' , \quad W' = k/\omega_{o_2}^2 = 0.61 \text{ Atd} \tag{A32}$$

Hence,

$$\left.\begin{aligned} L &= W'/AA' = 1.83 \text{ td}/A \\ C &= AA'/k = A^3/176 \, E_o \, t^3 \end{aligned}\right\} \tag{A33}$$

---

## Appendix B

## DERIVATION OF IMPEDANCE FROM EXPERIMENTAL RESULTS

### Constants of reference system

The reference system, when closed by a dummy head, was found to give rise to the normal results for a second-order linear system with light viscous damping. With a sinusoidal pressure input $p_i e^{j\omega t}$, and output $p_o e^{j(\omega t - \phi)}$,

$$(p_i/p_o)\, e^{j\Phi} = (1 - LC\omega^2) + jRC\omega \qquad \text{(B1)}$$

When $\Phi = \pi/2$, at $\omega = \omega_o$, say, we have

$$j(p_i/p_o)_{\omega_o} = 1 - LC\omega_o^2 + jRC\omega_o \ ,$$

so that

$$\left.\begin{aligned} LC\omega_0^2 &= 1 \\ RC\omega_o &= (p_i/p_o)_{\omega_o} \end{aligned}\right\} \qquad \text{(B2)}$$

and

For convenience in later work, we write

$$f_o = \omega_o/2\pi \qquad \text{(B3)}$$

and

$$h = RC\omega_o/2$$

Thus, the values of $\omega_o$, $f_o$, and h may be determined experimentally, and the absolute value of L or C found from the dimensions of the system.

### Measurement of human impedance

Referring to Fig.4, where Z represents the impedance to be measured, we find that

$$(p_i/p_o)\, e^{j\Phi} = (1 - LC\omega^2 + jRC\omega) + (R + jL\omega)/Z \qquad \text{(B4)}$$

whence

$$Z = X + jY = (R + jL\omega)/[(p_i/p_o)\, e^{j\Phi} - (1 - LC\omega^2 + j\omega RC)] \qquad \text{(B5)}$$

Using equation (B3), and putting $f/f_o = \omega/\omega_o = a$, and $p_i/p_o = \theta$, we find

$$\left.\begin{aligned} X &= (a\theta \sin \Phi + 2h\theta \cos \Phi - 2h)/\lambda \\ Y &= (a^3 + a\theta \cos \Phi - a + 4h^2 a - 2h\theta \sin \Phi)/\lambda \\ \lambda &= [(\theta \cos \Phi - 1 + a^2)^2 + (\theta \sin \Phi - 2ha)^2]\, 2\pi C f_o \end{aligned}\right\} \qquad (B6)$$

where

In the experimental system, h was found to be so small that the error in omitting it was negligible. The forms actually used were, therefore,

$$\left.\begin{aligned} X &= a\theta \sin \Phi/2\pi C f_o [(\theta \cos \Phi - 1 + a^2)^2 + (\theta \sin \Phi)^2] \\ Y &= a(\theta \cos \Phi - 1 + a^2)/2\pi C f_o [(\theta \cos \Phi - 1 + a^2)^2 + (\theta \sin \Phi)^2] \end{aligned}\right\} \qquad (B7)$$

Using the experimental values of $f_o$ and C,

$$\left.\begin{aligned} a &= f/45 \\ 2\pi C f_o &= 1/17 \end{aligned}\right\} \qquad (B8)$$

## Appendix C

## PROBABLE ERRORS IN MEASUREMENT OF IMPEDANCE

The readings on which all. the results are based are, for each condition and frequency, the two complex numbers $(a_i + jb_i)$ and $(a_o + jb_o)$, representing the ratio of the transducer outputs to the reference voltage. From these numbers, the ratio $\theta$, and the phase angle $\Phi$ between the output and input pressures are calculated, using the forms

$$\left.\begin{aligned} \theta^2 &= (a_i^2 + b_i^2)/(a_o^2 + b_o^2) \\ \text{and} \qquad \Phi &= \tan^{-1}(b_i/a_i) - \tan^{-1}(b_o/a_o) \end{aligned}\right\} \qquad \text{(C1)}$$

If the probable fractional errors in each of the four figures is the same, say E, the probable fractional error in $\theta$ can be shown to be 2E, and the probable error in $\Phi$ is 2E radians.

Now differentiating equation (B7) of Appendix B with respect to $\theta$ and $\Phi$, we have

$$\delta X = -(\delta\theta/\theta)\,A + (\delta\Phi)\,B\ , \quad \delta Y = -(\delta\theta/\theta)\,A - (\delta\Phi)\,B \qquad \text{(C2)}$$

where

$$\left.\begin{aligned} A &= X + 4\pi C f_o\, XY\,(1 - a^2)/a \\ \text{and} \qquad B &= Y - 2\pi C f_o\,(X^2 - Y^2)\,(1 - a^2)/a \end{aligned}\right\} \qquad \text{(C3)}$$

Then, since

$$(he/e) = (\delta\Phi) = 2E\ ,$$

the probable errors $E_x$ and Ey in X and Y, are given by

$$\begin{aligned} E_x^2 = E_y^2 &= 4E^2(A^2 + B^2) \\ &= 4E^2\,(X^2 + Y^2)\,[\{1 + 2\pi C f_o\,Y(1 - a^2)/a\}^2 + \{2\pi C f_o\,X(1 - a^2)/a\}^2] \end{aligned} \qquad \text{(C4)}$$

The probable errors in the electronic equipment were of the order of 1-2%. Calculations based on this figure, show that for most of the experiments, the probable error was less than 0.5 unit. But in the experiments to determine the impedance of the face, the probable error at low frequencies was so large a.3 to make the results almost meaningless. The minimum error of about ±0.5 unit occurred at 15 c/s, increasing again with increasing frequency to about 1.0 units at 90 c/s.

Error due to variation in capacity

Another possible source of error in the experiments with the rigid mask was a variation in the volume within the mask due to differences of facial contour between the subjects. Such variations would obviously alter the value of C in equation (B7) of Appendix B, and by shifting the natural frequency of the reference system would also vary the value of the frequency ratio a.

Since

$$a = f/f_0 \quad \text{and} \quad f_0 = 1/2\pi\sqrt{LC}\,, \quad a^2 = 4\pi^2 f^2 LC\,,$$

so that

$$a\,(da/dC) = 2\pi^2 f^2 L \tag{C5}$$

Differentiating X and Y with respect to C, we find

$$\left.\begin{aligned} (\delta X) &= -(\delta C)\,4\pi\, f X Y \\ \text{and} \quad (\delta Y) &= (SC)\,2\pi\, f\,(X^2 - Y^2) \end{aligned}\right\} \tag{C6}$$

The nominal value of C used in all the calculations was $2.0 \times 10^{-4}$, and it is thought that the variation in volume, to which C is directly proportional, was not likely to exceed 10%. Calculations based on this figure, show that the errors in most of the experiments were unlikely to exceed ±0.5 units, at the maximum frequency, with proportionately less at lower frequencies. Onoe again, the error in the measurements of the face impedance was larger, giving an error of ±1 ohm at 90 c/s, but less at lower frequencies.

## SYMBOLS

| Symbol | Description | Units |
|---|---|---|
| Z | complex impedance, (X ± jY) | gm $s^{-1}$ $cm^{-4}$ |
| X | resistive component of impedance | gm $s^{-1}$ $cm^{-4}$ |
| Y | reactive component of impedance | gm $s^{-1}$ $cm^{-4}$ |
| $Z_r$ | regulator impedance | gm $s^{-1}$ $cm^{-4}$ |
| $Z_{s_1}$ | piping impedance | gm $s^{-1}$ $cm^{-4}$ |
| $Z_{s_2}$ | mask impedance | gm $s^{-1}$ $cm^{-4}$ |
| $Z_m$ | human respiratory impedance | gm $s^{-1}$ $cm^{-4}$ |
| L | inductance | gm $cm^{-4}$ |
| C | capacitance | $s^2$ $cm^4$ $gm^{-1}$ |
| R | resistance | gm $s^{-1}$ $cm^{-4}$ |
| F | flow | $cm^3$ $s^{-1}$ |
| $F_i$ | demand flow of human | $cm^3$ $s^{-1}$ |
| $F_r$ | regulator exit flow | $cm^3$ $s^{-1}$ |
| $F_o$ | flow received by human | $cm^3$ $s^{-1}$ |
| P | pressure difference | dyne $cm^{-2}$ |
| $p_i$ | pressure amplitude of sinusoidal input | dyne $cm^{-2}$ |
| $p_o$ | pressure amplitude of sinusoidal output | dyne $cm^{-2}$ |
| $p_d$ | regulator datum pressure | dyne $cm^{-2}$ |
| $p_r$ | regulator outlet pressure | dyne $cm^{-2}$ |
| c | velocity of sound in gas | cm $s^{-1}$ |
| v | velocity | cm $s^{-1}$ |
| μ | dynamic viscosity of gas | gm $cm^{-1}$ $s^{-1}$ |
| ρ | gas density | gm $cm^{-3}$ |
| g | acceleration due to gravity | cm $s^{-2}$ |
| V | volume | $cm^3$ |
| θ | ratio of moduli of input and output pressures, $p_i$ and $p_o$ | - |
| a | ratio of angular frequency to natural angular frequency of reference system | - |
| Φ | phase angle between input and output pressure signals $p_i$ and $p_o$ | degrees |
| ω | angular frequency | radians $s^{-1}$ |

probable error was less than 0.5 unit. But in the experiments to determine the impedance of the face, the probable error at low frequencies was so large as to make the results almost meaningless. The minimum error of about ±0.5 unit occurred at 15 c/s, increasing again with increasing frequency to about 1.0 units at 90 c/s.

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Since

$$a = f/f_0 \quad \text{and} \quad f_0 = 1/2\pi\sqrt{LC}\,, \quad a^2 = 4\pi^2 f^2 LC\,,$$

so that

$$a\,(da/dC) = 2\pi^2 f^2 L \tag{C5}$$

Differentiating X and Y with respect to C, we find

$$\left.\begin{aligned} (\delta X) &= -(\delta C)\;4\pi\; f X Y \\ \text{and}\qquad (\delta Y) &= (\delta C)\;2\pi\; f\;(X^2 - Y^2) \end{aligned}\right\} \tag{C6}$$

The nominal value of C used in all the calculations was $2.0 \times 10^{-4}$, and it is thought that the variation in volume, to which C is directly proportional, was not likely to exceed 10%. Calculations based on this figure, show that the errors in most of the experiments were unlikely to exceed ±0.5 units, at the maximum frequency, with proportionately less at lower frequencies. Once again, the error in the measurements of the face impedance was larger, giving an error of ±1 ohm at 90 c/s, but less at lower frequencies.

## SYMBOLS

| Symbol | Desoripticn | units |
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| $Z_m$ | human respiratory impedance | gm $s^{-1}$ $cm^{-4}$ |
| L | inductance | gm $cm^{-4}$ |
| C | capacitance | $s^2$ $cm^4$ $gm^{-1}$ |
| R | resistance | gm $s^{-1}$ $cm^{-4}$ |
| F | flow | $cm^3$ $s^{-1}$ |
| $F_i$ | demand flow of human | $cm^3$ $s^{-1}$ |
| $F_r$ | regulator exit flow | $cm^3$ $s^{-1}$ |
| $F_o$ | flow received by human | $cm^3$ $s^{-1}$ |
| P | pressure difference | dyne $cm^{-2}$ |
| $p_i$ | pressure amplitude of sinusoidal input | dyne $cm^{-2}$ |
| $p_o$ | pressure amplitude of sinusoidal output | dyne $cm^{-2}$ |
| $p_d$ | regulator datum pressure | dyne $cm^{-2}$ |
| $p_r$ | regulator outlet pressure | dyne $cm^{-2}$ |
| c | velocity of sound in gas | cm $s^{-1}$ |
| v | velocity | cm $s^{-1}$ |
| μ | dynamic viscosity of gas | gm $cm^{-1}$ $s^{-1}$ |
| ρ | gas density | gm $cm^{-3}$ |
| g | acceleration due to gravity | cm $s^{-2}$ |
| V | volume | $cm^3$ |
| θ | ratio of moduli of input and. output pressures, $p_i$ and $p_o$ | - |
| a | ratio of angular frequency to natural angular frequency of reference system | - |
| Φ | phase angle between input and output pressure signal3 $p_i$ and $p_o$ | degrees |
| ω | angular frequency | radians $s^{-1}$ |

## SYMBOLS (Contd)

| Symbol | Description | units |
|---|---|---|
| $\omega_o$ | natural angular frequency of reference system | radians $s^{-1}$ |
| $f$ | periodic frequency | (cycles/sec) |
| $f_o$ | natural periodic frequency of reference system | (cycles/sec) |
| $n$ | polytropic index | |
| $\gamma$ | ratio of specific heats of gas | |
| $N$ | regulator transfer function | s $cm^4$ $gm^{-1}$ |
| $G$ | summing unit of human | |
| $h$ | damping factor | |
| $K$ | gain factor | |
| $T$ | time constant | s |
| $\ell$ | pipe length | cm |
| $\ell'$ | effective pipe length | cm |
| $r$ | pipe radius | cm |
| $b$ | radius of stretched membrane | cm |
| $t$ | thickness of stretched membrane | cm |
| $x$ | displacement of membrane centre | cm |
| $r_1$ | distance from centre of diaphragm | cm |
| $A$ | cross-sectional area of rigid diaphragm | $cm^2$ |
| $A'$ | effective surface area of flexible diaphragm | $cm^2$ |
| $d$ | density of stretched membrane | gm $cm^{-3}$ |
| $\omega_{o_1}$ | lowest natural angular frequency for stretched circular membrane | radians $s^{-1}$ |
| $\omega_{o_2}$ | lowest natural angular frequency for built-in circular diaphragm | radians $s^{-1}$ |
| $k$ | spring rate | dynes $cm^{-1}$ |
| $q$ | damping force per unit velocity | dynes s $cm^{-1}$ |
| $s$ | peripheral stress of membrane | dynes $cm^{-1}$ |
| $W$ | diaphragm mass | gm |
| $W'$ | effective mass for simple flexible diaphragm | gm |
| $E_o$ | Young's modulus for diaphragm | dynes $cm^{-2}$ |
| $\nu$ | Poisson's ratio for diaphragm | |
| $E$ | probable fractional errors in $\theta$ and $\Phi$ | |

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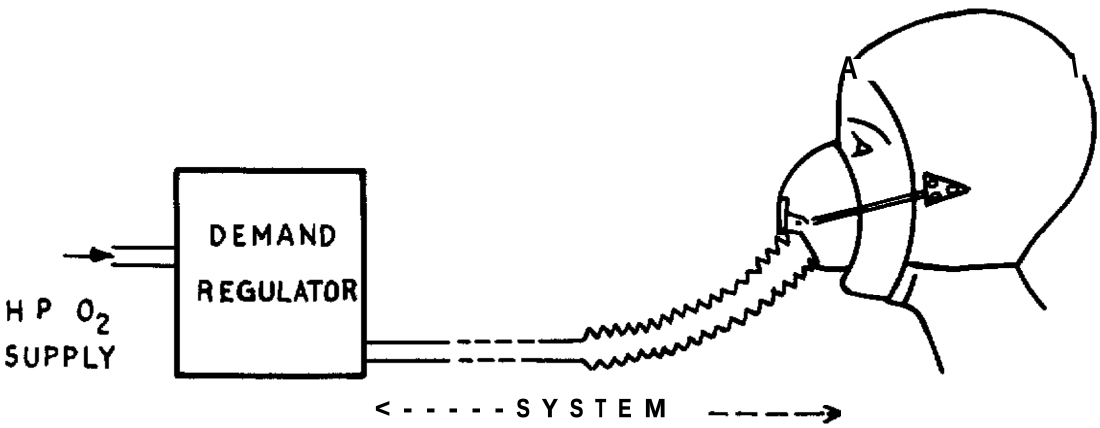



**FIG. 1(a) SCHEMATIC LAYOUT**

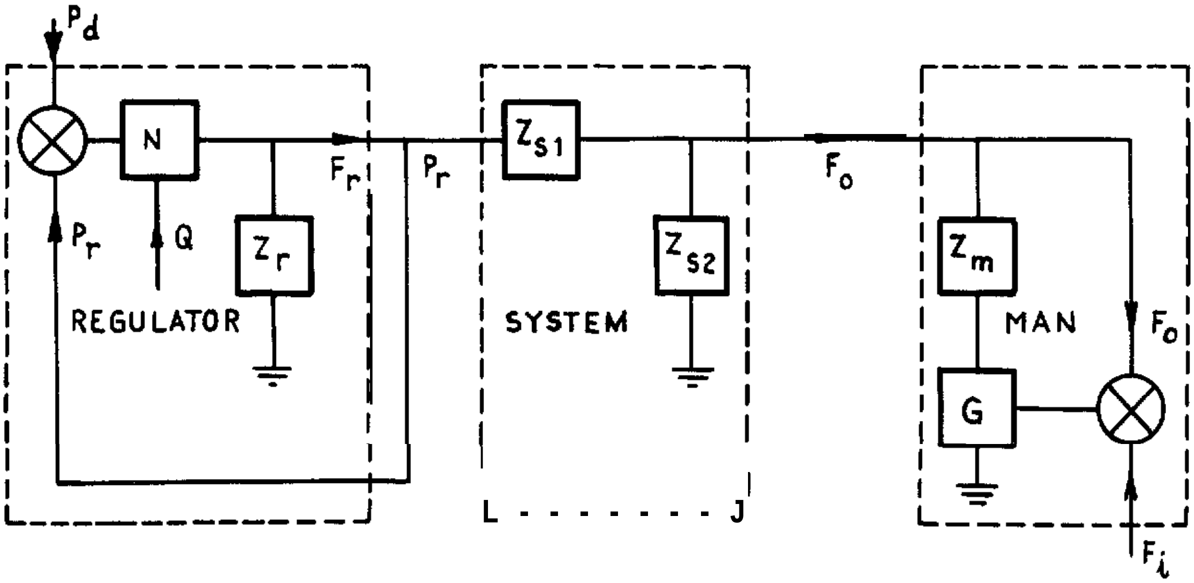



**FIG. 1 (b) CONTROL SYSTEM**

FIG.1 AIRCRAFT BREATHING EQUIPMENT AS A CONTROL SYSTEM

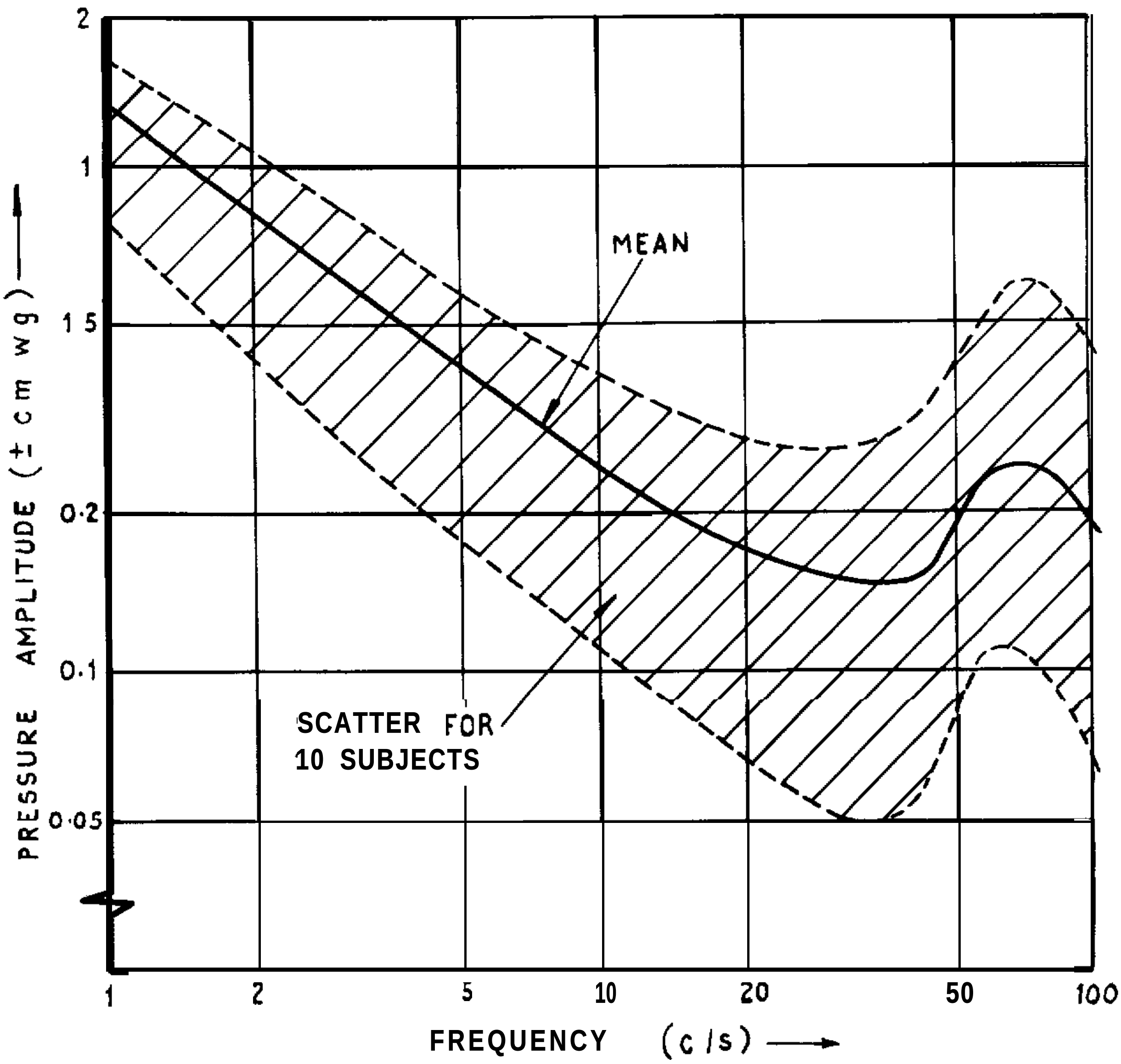



FIG. 2 PERCEPTION OF OSCILLATING PRESSURE

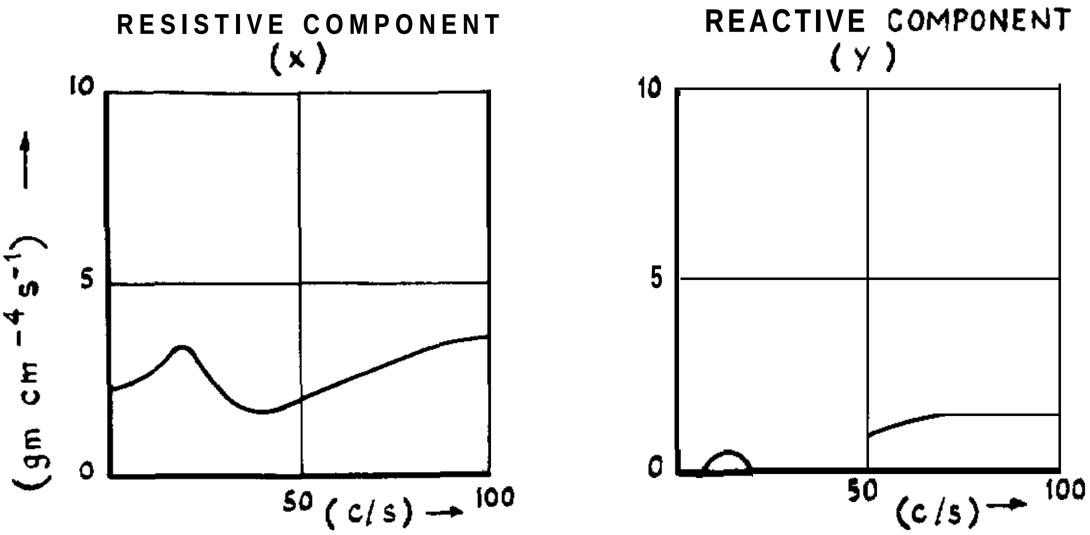



FIG.3 (a) MOUTH BREATHING

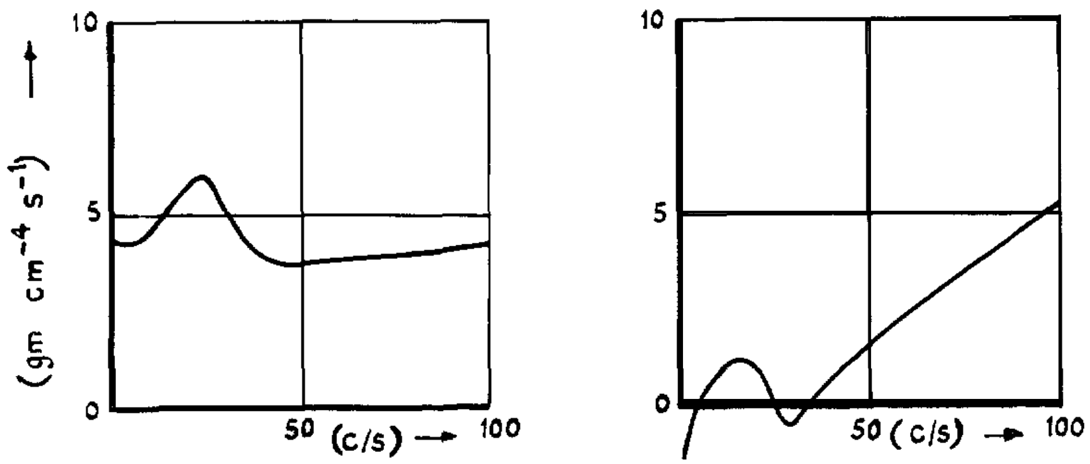



FIG. 3(b) NOSE BREATHING

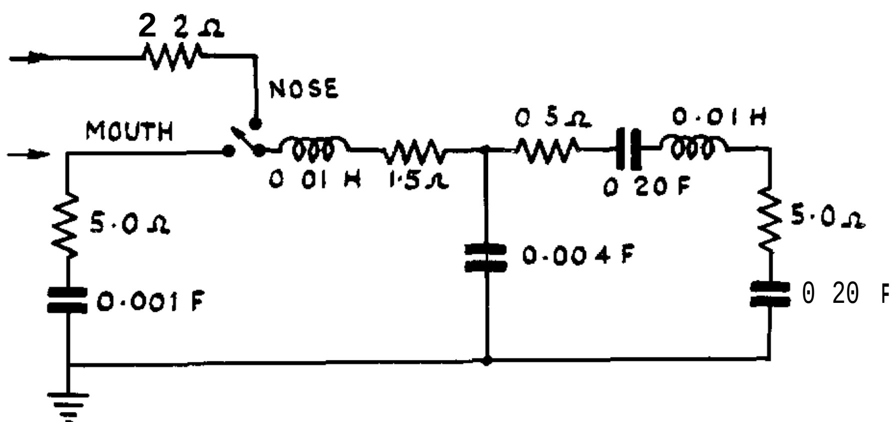
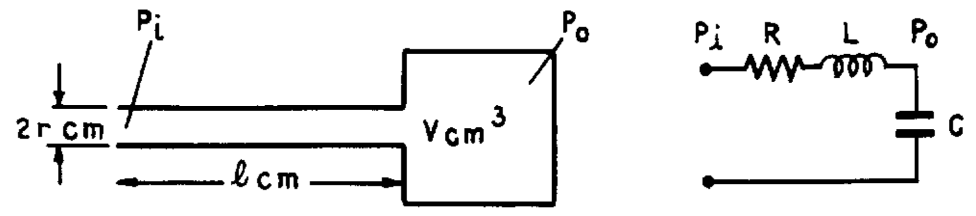



FIG.3 ELECTRICAL ANALOGUE OF HUMAN RESPIRATORY SYSTEM
(AFTER SHEPHARD ref 5 )




$$L = \ell\rho / \pi r^2$$

$$C = V / \rho c^2$$

$$P_i / P_o = 1 - LC\omega^2 + jRC\omega$$

FIG. 4(a) REFERENCE SYSTEM

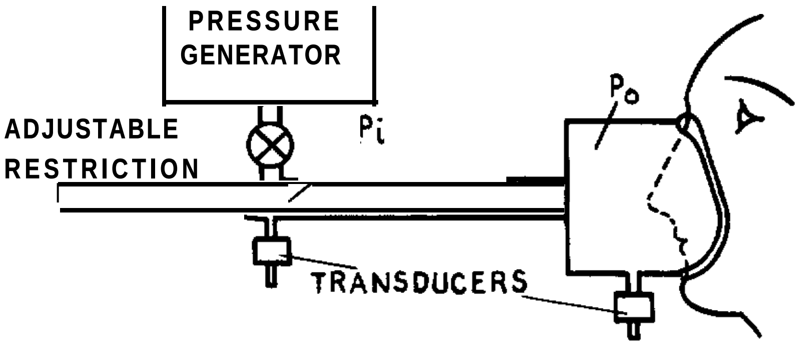



FIG. 4 (b) TESTS A, B, C & F

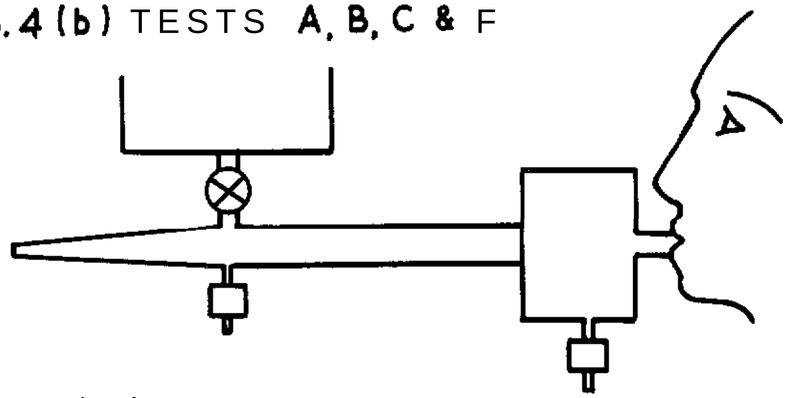
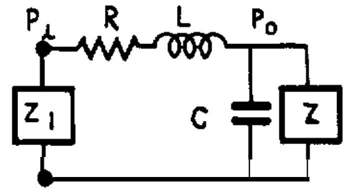


FIG 4 (c ) TESTS DI, E & G




R, L, C, CONSTANTS OF REFERENCE SYSTEM

$Z_1$, IMPEDANCE OF DOWNSTREAM SYSTEM

Z, HUMAN IMPEDANCE

$$P_i / P_o = (R + j\omega L)/Z + (1 - LC\omega^2 + jRC\omega)$$

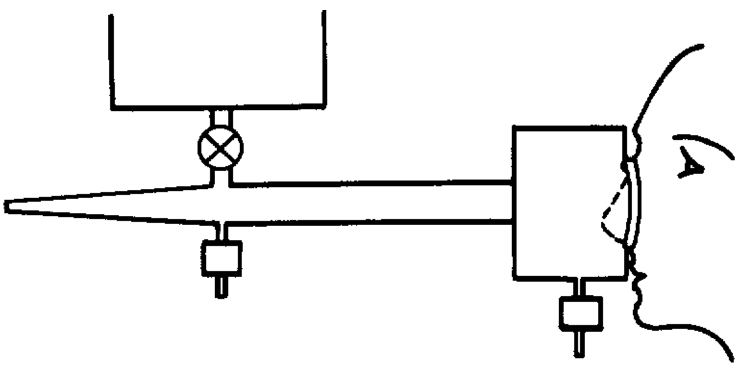
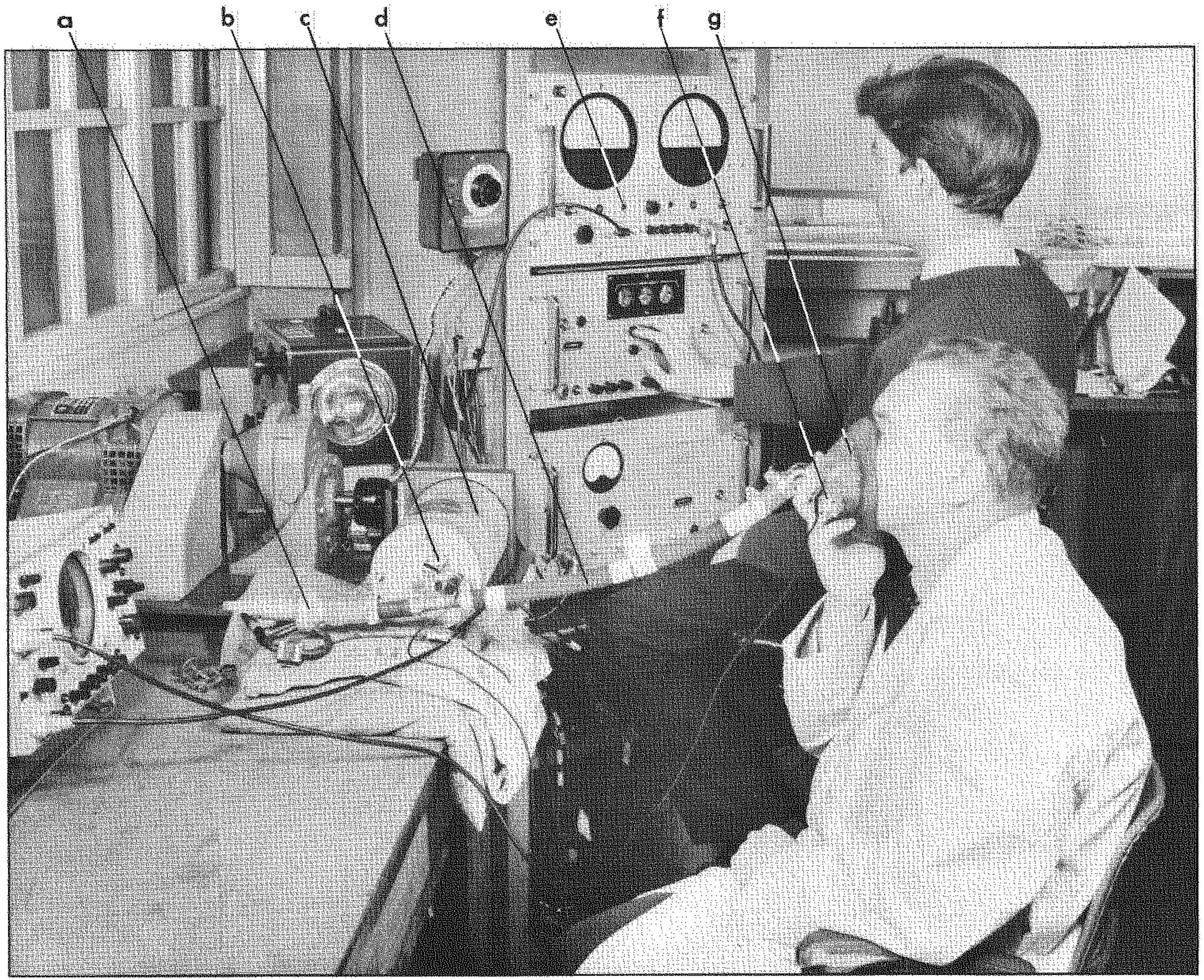


FIG 4(d) TEST D2

FIG. 4 EXPERIMENTAL SET- UP WITH ELECTRICAL ANALOGUES




a. FLOW TRANSDUCER
b. INPUT PRESSURE TRANSDUCER
c. PRESSURE GENERATOR
d. REFERENCE SYSTEM PIPE
e. RESOLVED COMPONENTS INDICATOR
f. OUPUT PRESSURE TRANSDUCER
g. RIGID MASK

Fig 5 Test in progress

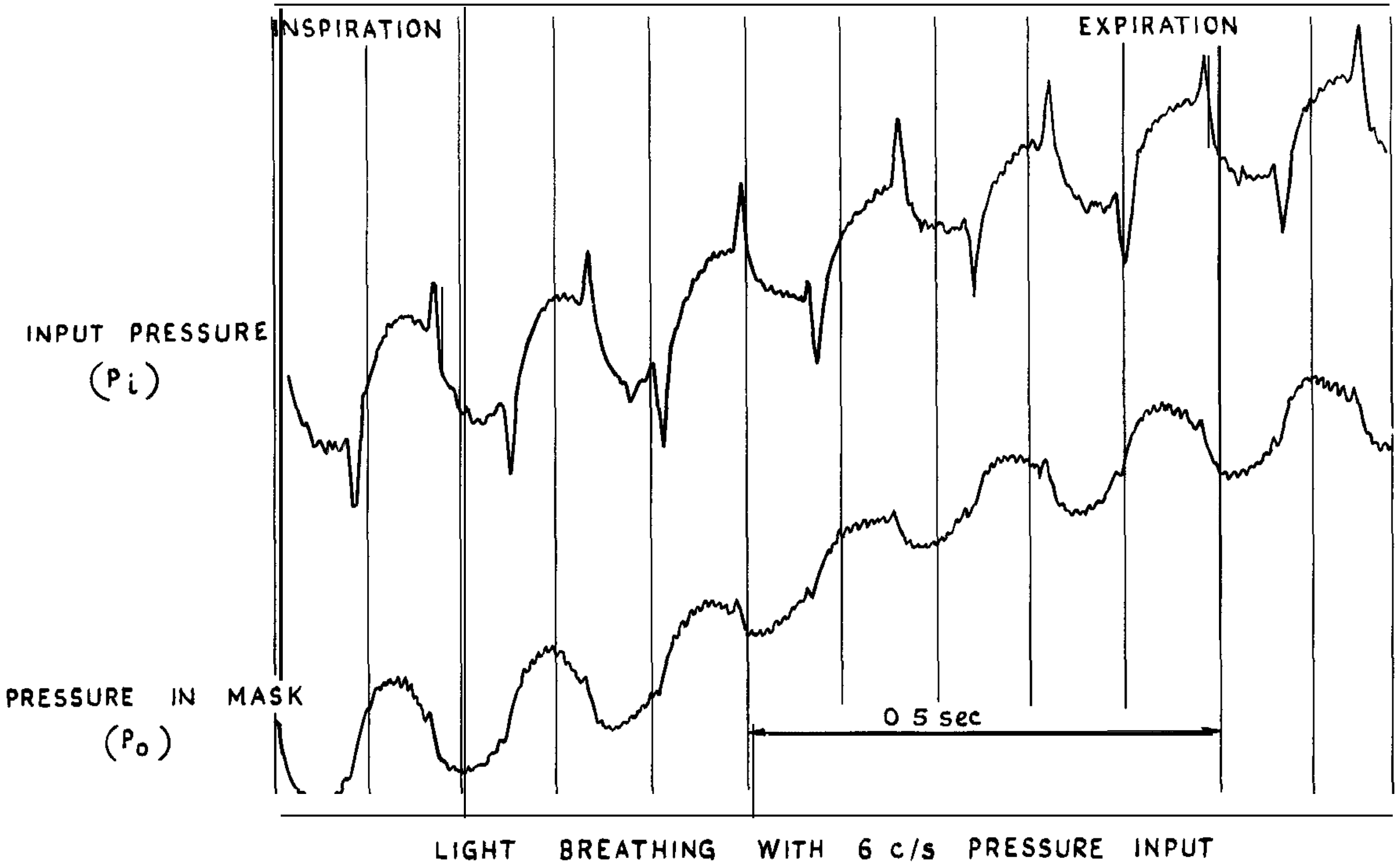



FIG. 6 TYPICAL PRESSURE RECORD

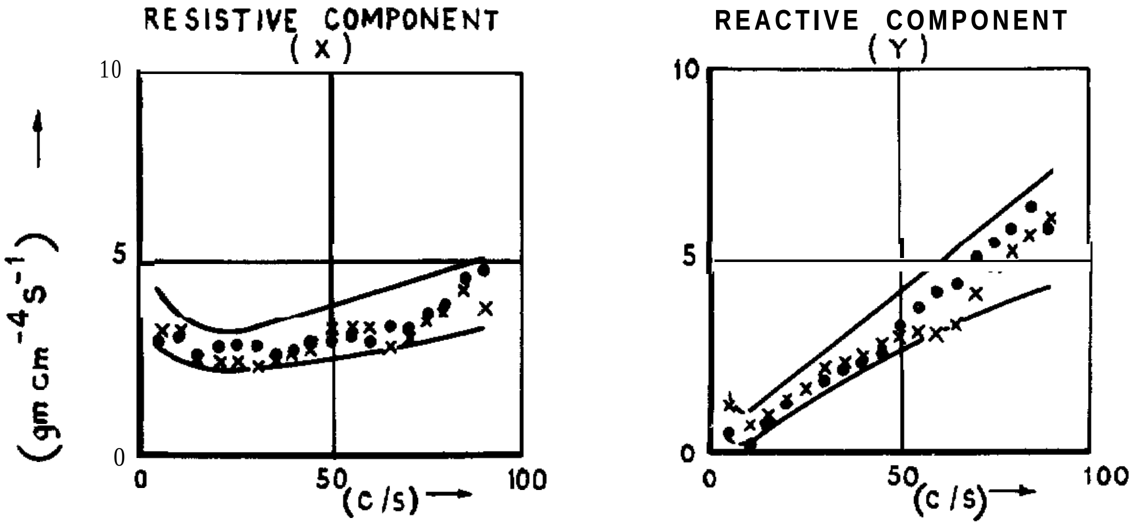



FIG. 7(a) MOUTH BREATHING (LIGHT)
TESTS A1, B1

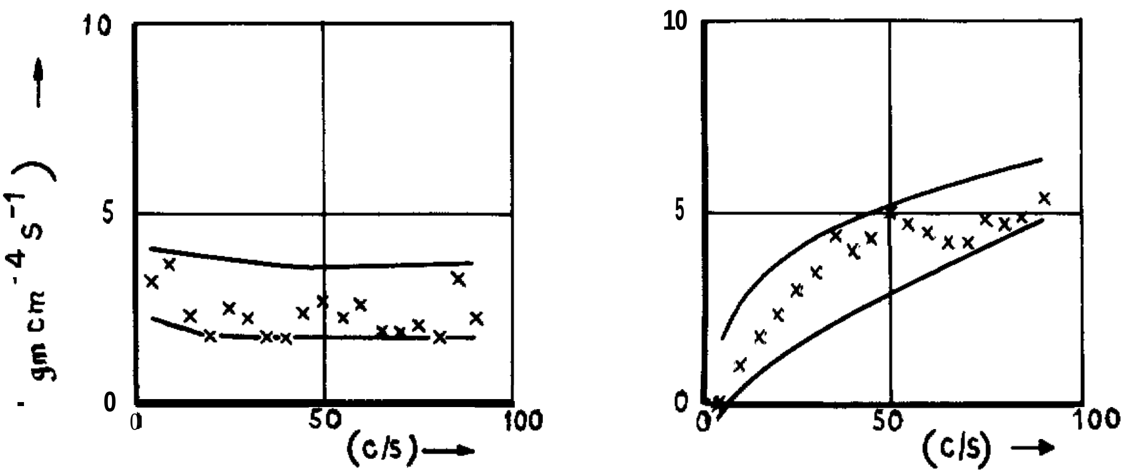



FIG. 7 (b) MOUTH BREATHING (HEAVY)
TESTS A 2

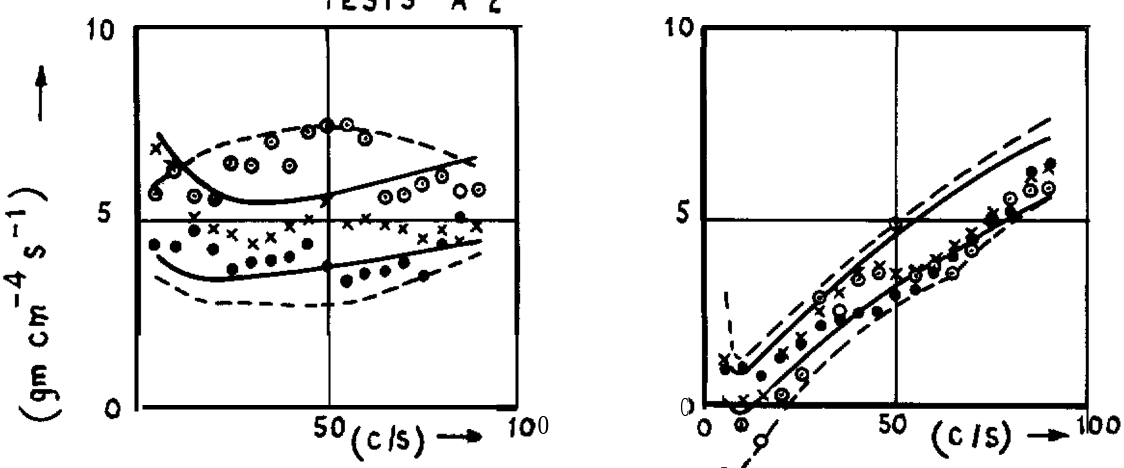



FIG. 7(c) NOSE BREATHING (LIGHT)
TESTS A3, B2, C

/ LIMITS OF SCATTER BAND AT GROUND LEVEL
-- LIMITS OF SCATTER BAND AT GROUND LEVEL WITH INCREASED RESISTANCE TO BREATHING
X G F R AT GROUND LEVEL
⊙ G F R AT GROUND LEVEL WITH INCREASED RESISTANCE
· G F R AT 8000 ft

FIG.7 IMPEDANCE OF HUMAN RESPIRATORY SYSTEM INCLUDING FACE

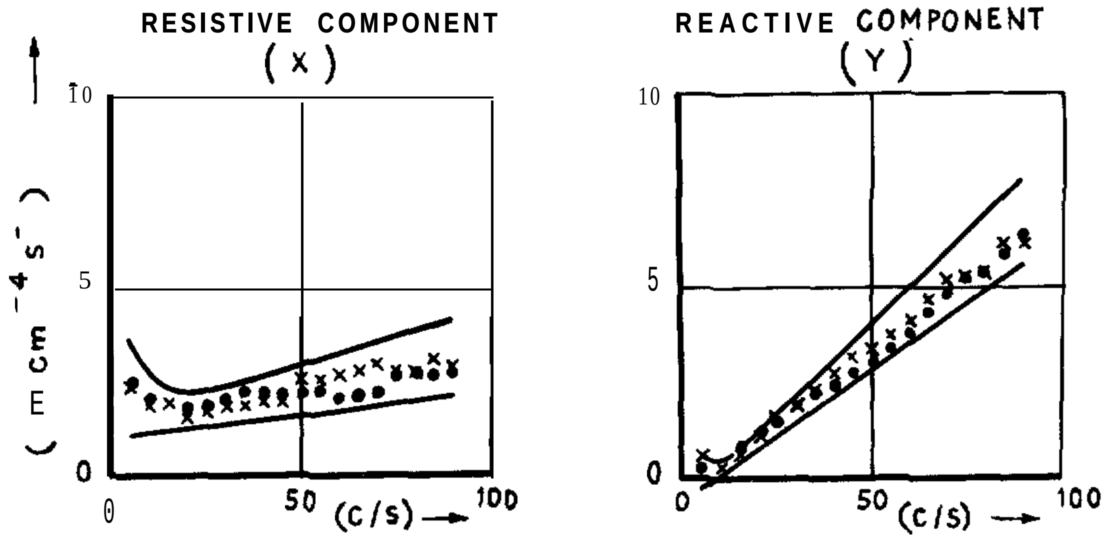



FIG. 8(a) MOUTH BREATHING ( LIGHT )

TEST D1, E2

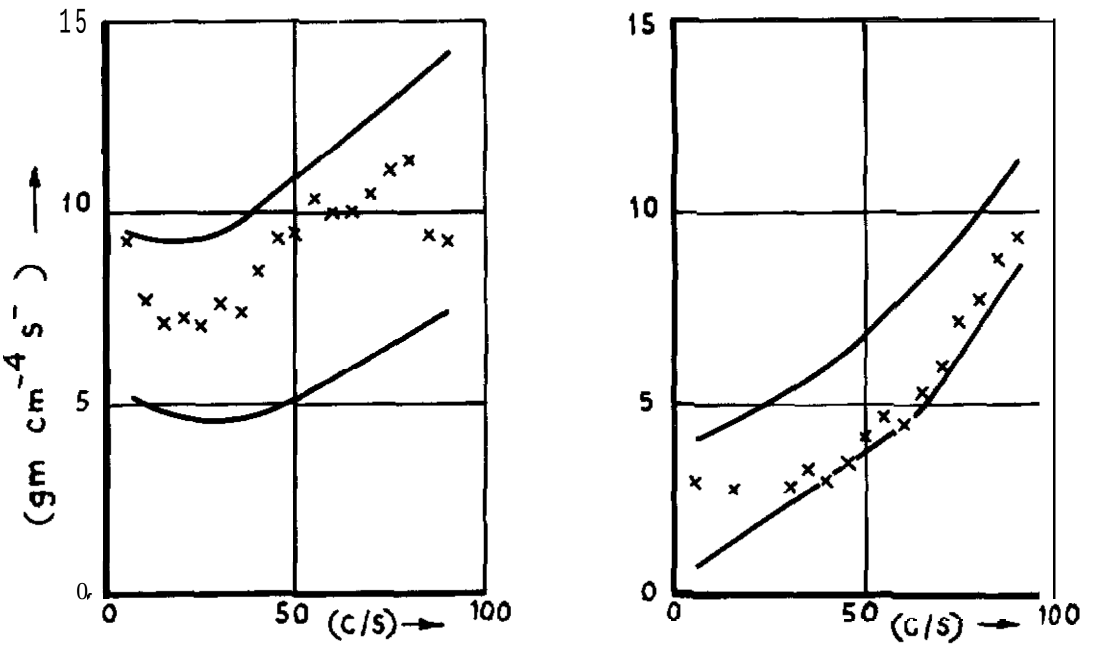



FIG. 8(b) NOSE BREATHING ( LIGHT )

TEST D2

/ LIMITS OF SCATTER BAND AT GROUND LEVEL

x G F R AT GROUND LEVEL

. G F R AT 8000 ft

FIG.8 IMPEDANCE OF HUMAN RESPIRATORY SYSTEM EXCLUDING FACE

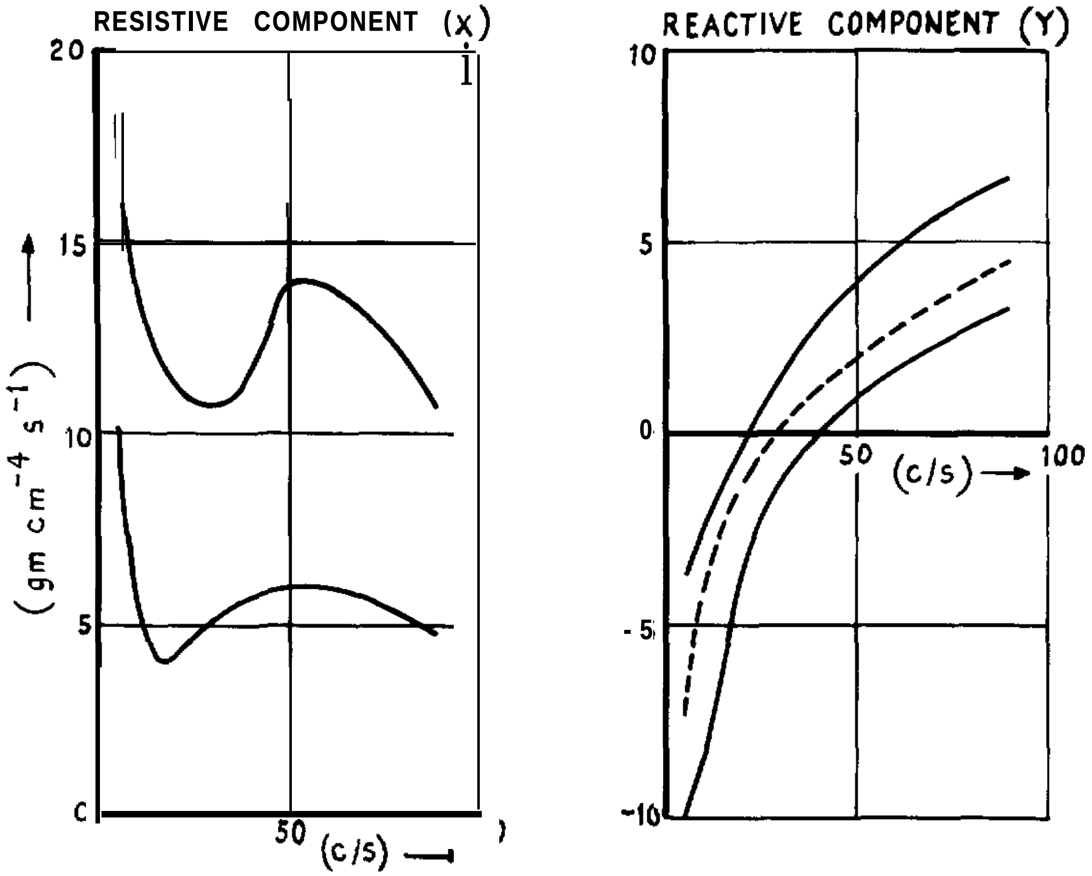



FIG.9 IMPEDANCE OF FACE (TEST E)

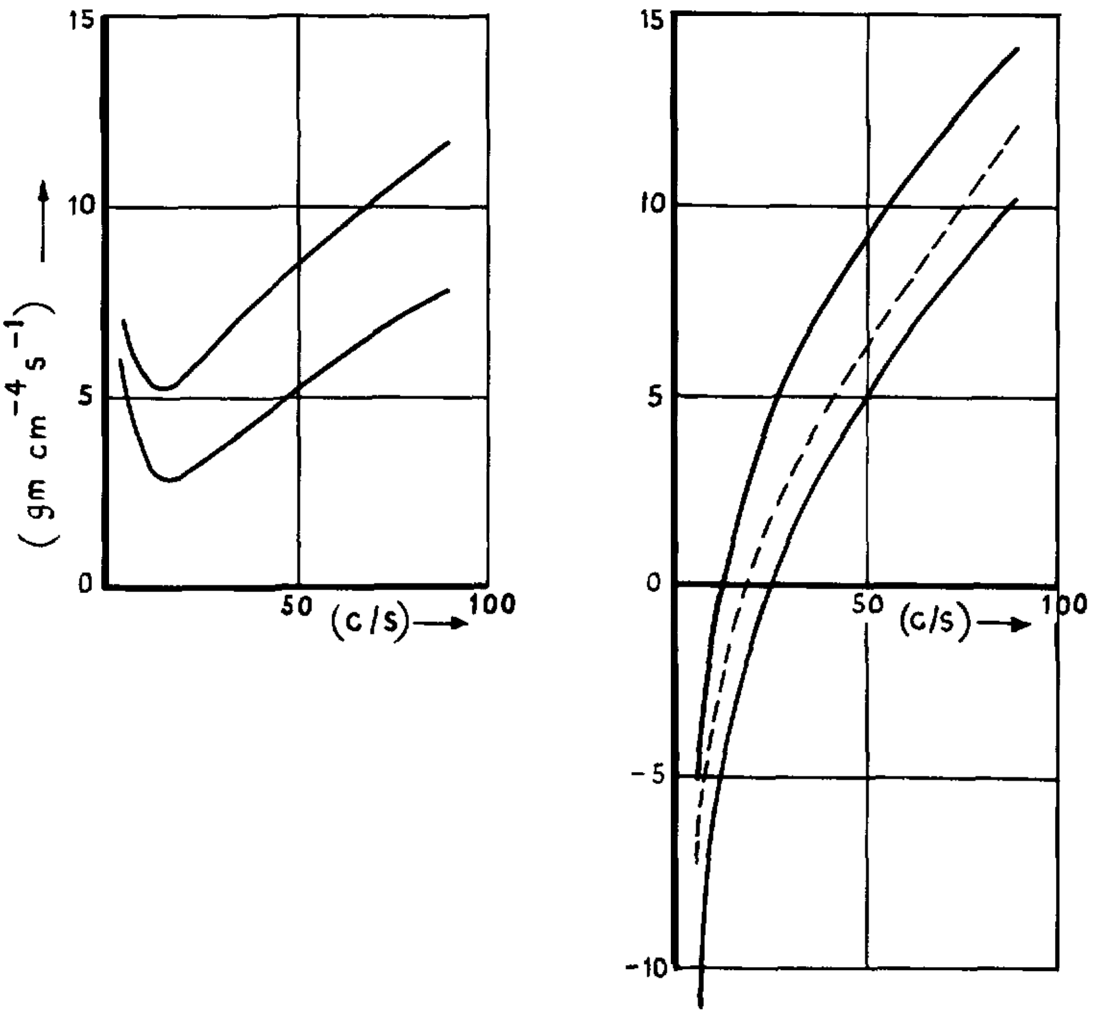



FIG.10 IMPEDANCE OF MOUTH INTERIOR

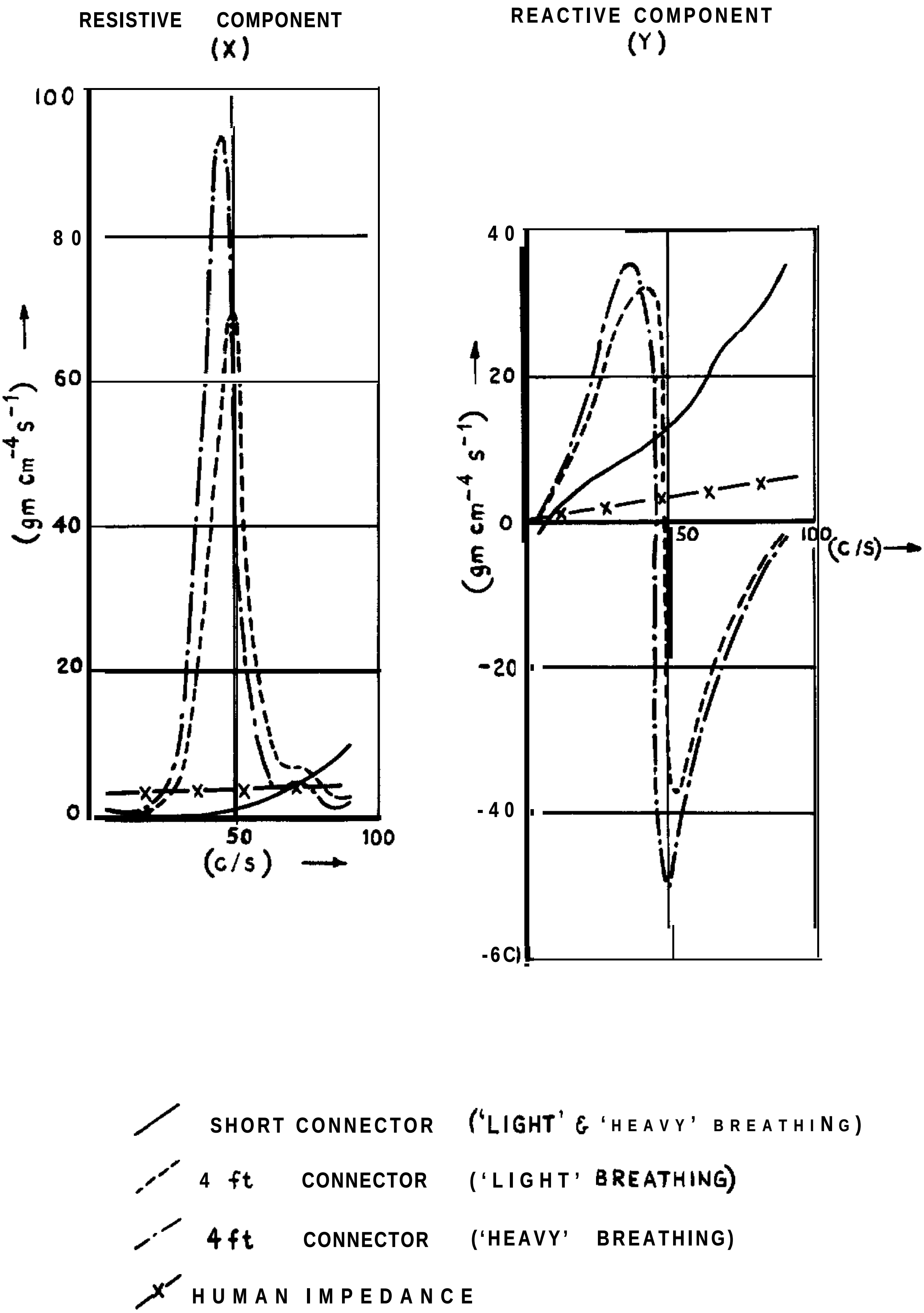



FIG.11 IMPEDANCE OF Mk II 'BEAVER'RESPIRATOR

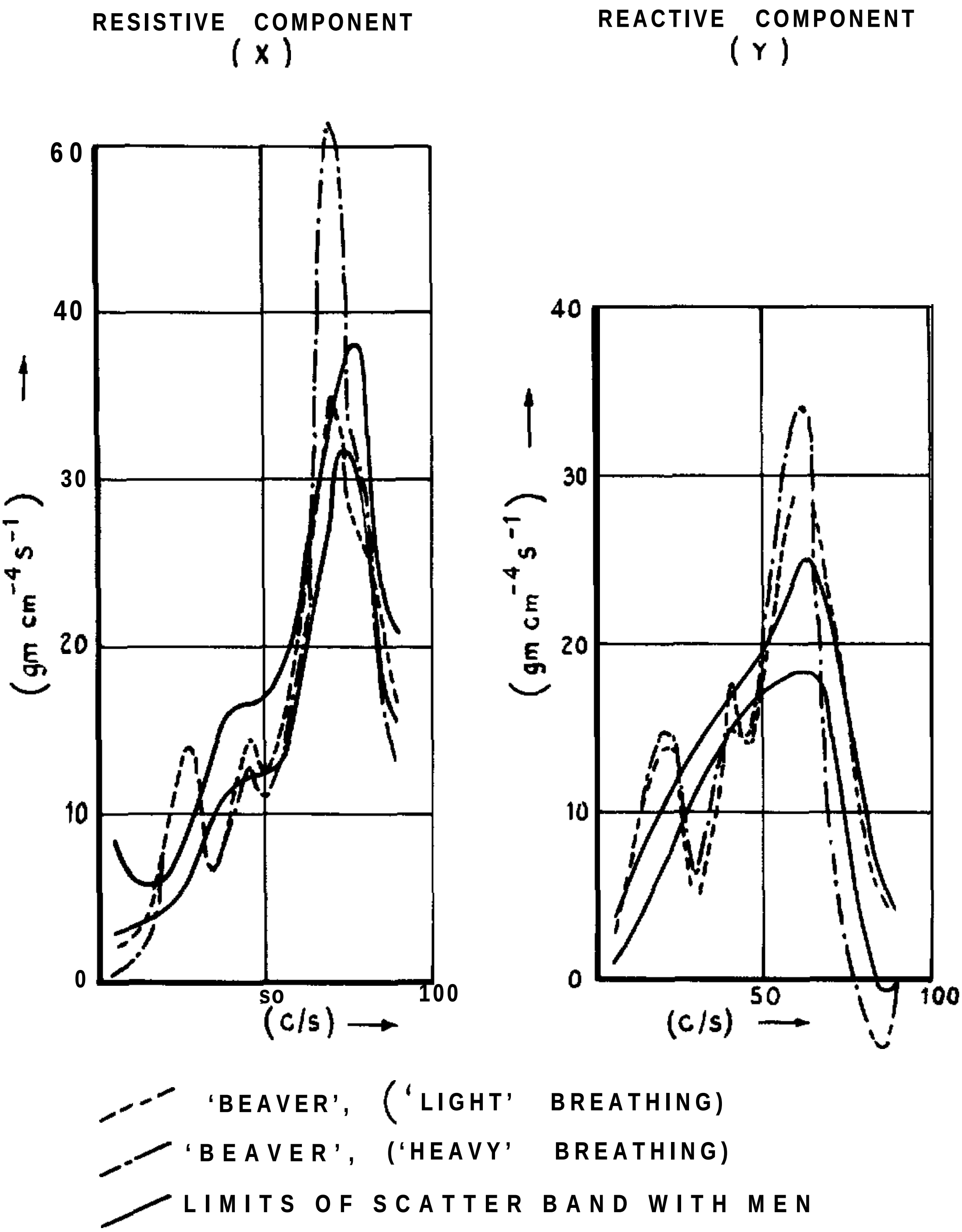
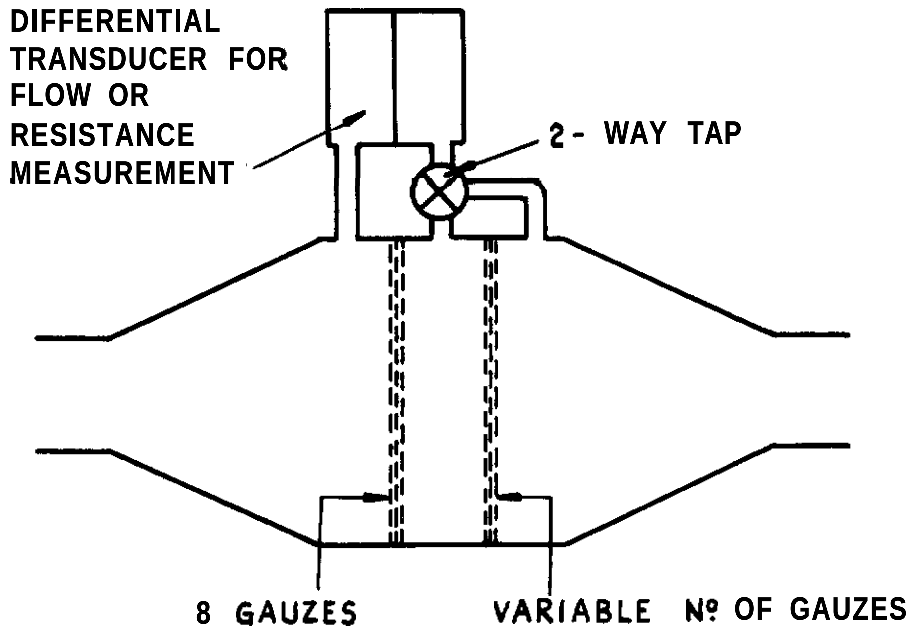



FIG.12 IMPEDANCE OF SEAT-MOUNTED SYSTEM WITH 'BEAVER' Mk II AND WITH MAN



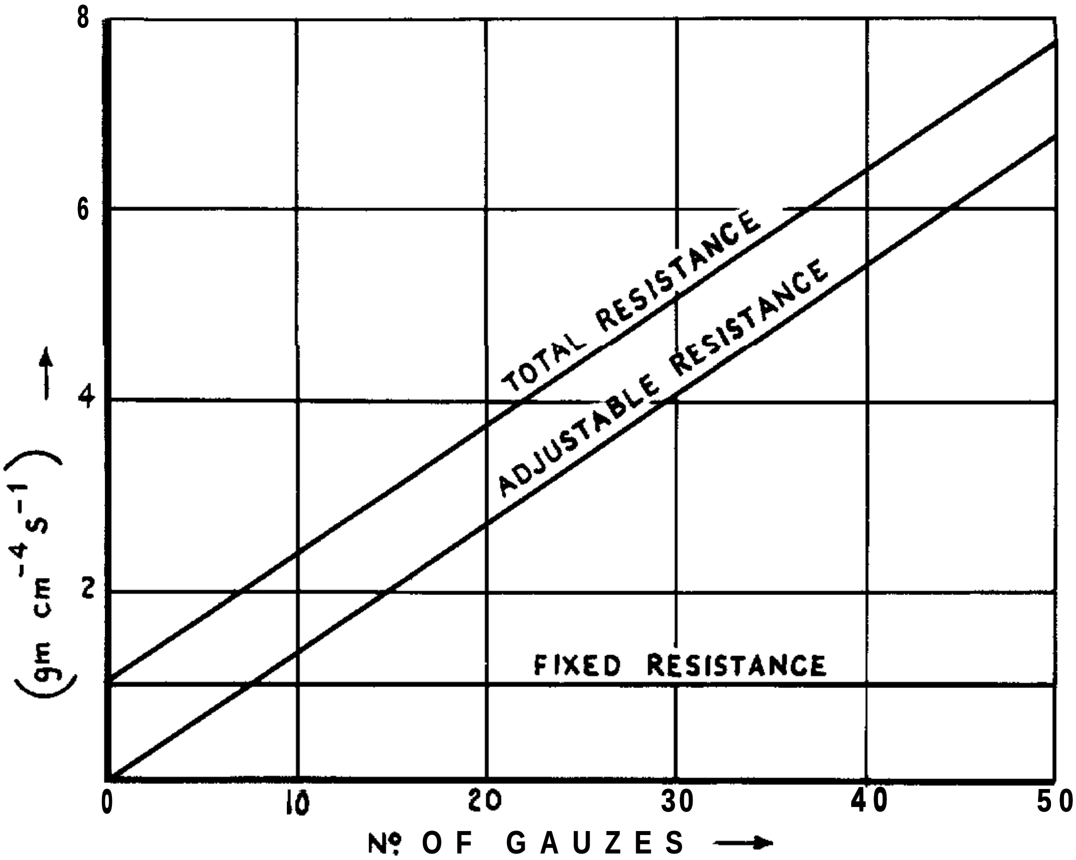




FIG.13 RESISTANCE SIMULATION AND FLOW MEASUREMENT

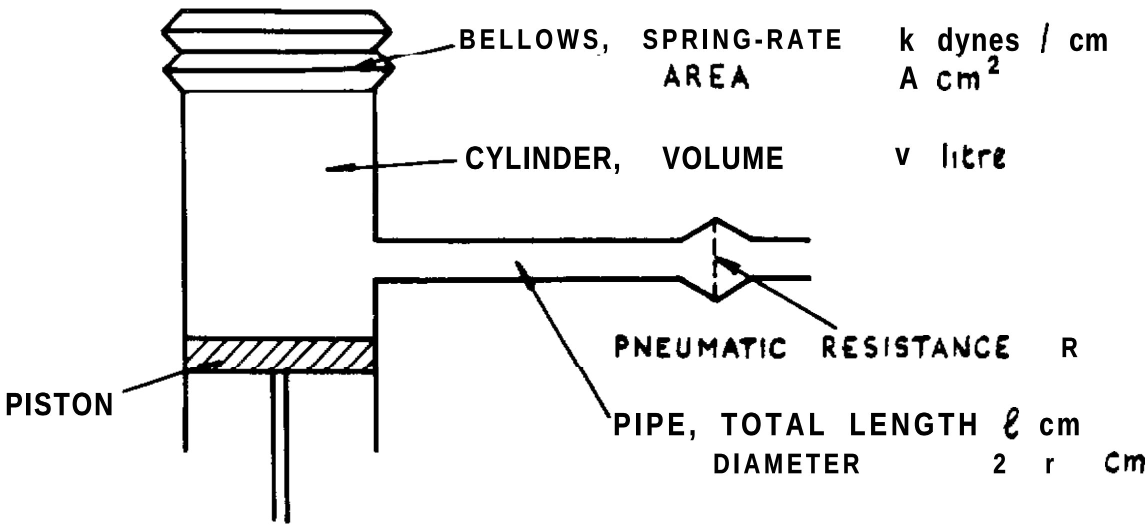



FIG. 14 (a) SCHEMATIC LAYOUT

$$L = \rho \ell / \pi r^2$$

$$C_1 = V / \rho c^2$$

$$C_2 = A^2 / k$$

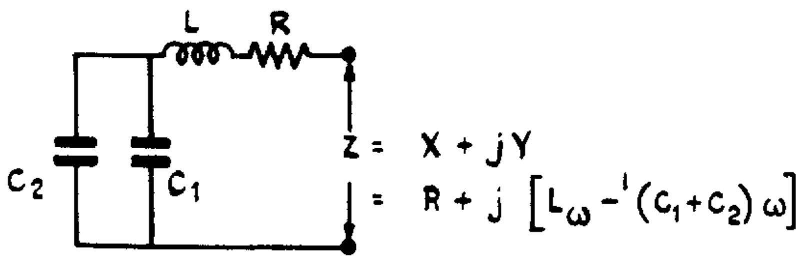



FIG. 14 (b) ELECTRICAL ANALOGUE

FIG. 14 FEATURES OF BREATHING SIMULATOR

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Maslen, K.R.
Rowlands, G.F.

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