

Airflow Perturbation Device for Respirator Research and Medical Screening*

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Abstract

The Airflow Perturbation Device (APD) is presented as a means of routinely and noninvasively measure airways resistance of humans for the purpose of respirator research and medical surveillance. The theoretical basis of the APD is presented, APD construction is outlined, and some preliminary experimental results are described. Results thus far have been consistent with expectations and can be used as the basis for further development.

Introduction

Medical examinations will continue to be important wherever workers are required to wear respiratory protective devices. When research has clearly demonstrated the utility of their results, and measurement interpretation is much less questionable, pulmonary function tests will probably be given during pre-employment physical examinations, annual medical examinations, and at post-exposure medical examinations (perhaps including bronchial challenge). Questions which are sure to be asked include: Is the worker a health risk? Can the worker wear a respirator? Which respirator(s) can the worker wear? Will the worker be impaired while wearing a respirator? How can the worker be matched to the job and to the respirator? These are fairly sophisticated questions, and the answers require sophisticated techniques, equipment, respirators, and knowledge. However, a prescription of respiratory protection

will eventually require development of this degree of sophistication.

As it appears now, require pulmonary function tests should include measurements of respiratory resistance, compliance, dead volume, carbon dioxide sensitivity, and maximum developed intrapleural pressures. Each of these has some predictive ability concerning the ability of workers to comfortably wear respiratory masks.

Widespread measurement of respiratory resistance will require a simple, light weight, easily used device which will require no special breathing maneuvers. Automatic data analysis is advantageous because operator variability will be eliminated.

In respirator research, there is need for an instrument which can be used to measure resistance during exercise, as well as at rest, and while wearing a mask, as well as while unmasked. Such a device should be portable if possible, and be able to allow the introduction of contaminants or bronchial-active agents while the measurement of resistance is being made. A strong candidate for such an instrument is the Airflow Perturbation Device (APD).

*Scientific Article Number A3724 Contribution Number 6700 of the Maryland Agricultural Experiment Station (Department of Agricultural Engineering).

Device Theory

The method by which the APD works is based upon the simple respiratory system model of Fig. 1, wherein the respiratory system is considered to be composed of a resistance and compliance. An external resistance is periodically switched into the circuit. Otherwise, the mouth delivers the air through the pneumotach to the atmosphere. A pressure balance on the above model, with external resistor switched in, gives:

$$\dot{V}R_s + \dot{V}R_{pt} + \dot{V}R_R + \frac{1}{C_R} \int \dot{V}dt = p_{mus} \tag{1}$$

where

- $\dot{V}$ = flow rate,
- R_s = external resistance,
- R_{pt} = pneumotach resistance,
- R_R = respiratory resistance,
- C_R = respiratory compliance,
- t = time, and
- p_{mus} = muscle pressure.

A small flow perturbation, $\Delta\dot{V}$, is introduced by means of the switched resistor (R_s becomes zero):

$$(\dot{V} + \Delta\dot{V})R_{pt} + (\dot{V} + \Delta\dot{V})R_R + \frac{1}{C_R} \int (\dot{V} + \Delta\dot{V})dt = p_{mus} \tag{2}$$

Next, consider mouth pressure, p_m :

$$p_m = p_{mus} - \left(\dot{V}R_R + \frac{1}{C_R} \int \dot{V}dt \right) \tag{3}$$

when the flow perturbation begins,

$$p_m + \Delta p_m = p_{mus} - R_R(\dot{V} + \Delta\dot{V}) - \frac{1}{C_R} \int (\dot{V} + \Delta\dot{V}) dt \tag{4}$$

Mechanical properties of the respiratory system and muscle pressure are assumed to be unaffected by the small perturbation.

$$\Delta p_m = p_m + \Delta p_m - p_m = -R_R\Delta\dot{V} - \frac{1}{C_R} \int \Delta\dot{V}dt \tag{5}$$

and

$$\frac{\Delta p_m}{\Delta\dot{V}} = - \left(R_R + \frac{1}{\Delta V C_R} \int \Delta\dot{V}dt \right) \tag{6}$$

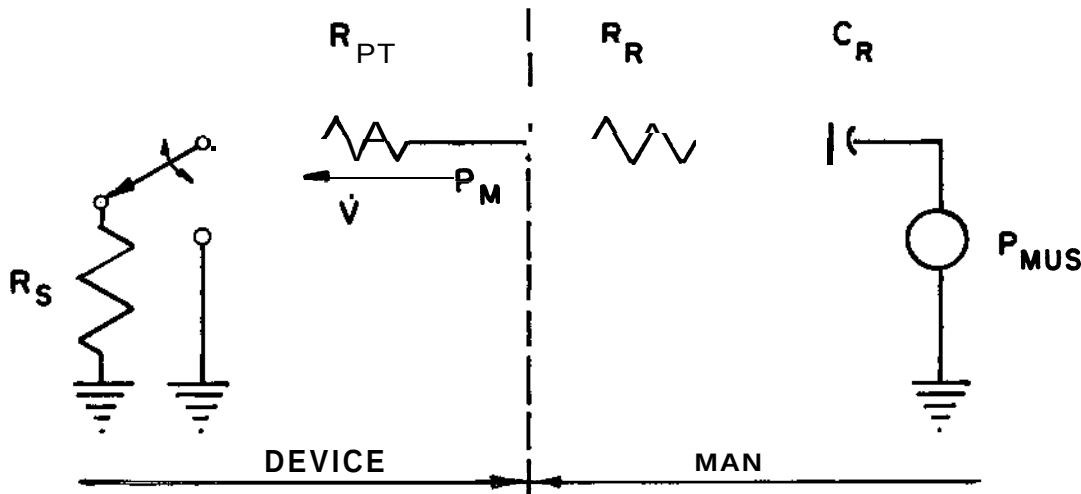


Figure 1. Simplified model of the respiratory system and APD including respiratory compliance (C_R), respiratory resistance (R_R), pneumotach resistance (R_{PT}), and screen resistance (R_s).

Instead of switching the external resistor, it is caused to undergo a smooth transition from minimum to maximum value. The resistance is chosen such that

$$\dot{A}\dot{V} = \Delta\dot{V}_o \sin \omega_o t \quad , \quad (7)$$

where

$\Delta\dot{V}_o$ = magnitude of flow perturbation sine wave,
 ω_o = angular frequency of sine wave;

then

$$\Delta p_m \sim \Delta p_o \sin (\omega_o t + \theta) \quad ,$$

where

Δp_o = magnitude of pressure perturbation sine wave and
 θ = phase angle between pressure and flow perturbations.

Pressure and flow perturbations form a set of rotating vectors with magnitudes Δp_o and $\Delta\dot{V}_o$ and angles $(\omega_o t + \theta)$ and $\omega_o t$, respectively. The phase angle between the two is a constant, θ . The pressure perturbation vector, Δp_m , forms components parallel to (in phase with) $\dot{A}\dot{V}$ and perpendicular to (out of phase with) $\Delta\dot{V}$. These two components are $\Delta p_o \cos \theta$ and $\Delta p_o \sin \theta$. Thus,

$$\begin{aligned} \Delta p_o \sin (\omega_o t + \theta) &= (\Delta p_o \cos \theta) \sin \omega_o t \\ &+ (\Delta p_o \sin \theta) \cos \omega_o t \quad . \end{aligned} \quad (9)$$

Inserting these relationships into Eq. (6),

$$\begin{aligned} \frac{\Delta p_o \cos \theta \sin \omega_o t}{\dot{A}\dot{V} \sin \theta} + \frac{\Delta p_o \sin \theta \cos \omega_o t}{\Delta\dot{V}_o \sin \omega_o t} &= \\ - \left(R_R + \frac{1}{C_R \omega_o \sin \theta} \right) \quad . \end{aligned} \quad (10)$$

Equating in-phase and out-of-phase components:

$$R_R = \frac{-\Delta p_o}{\Delta\dot{V}_o} \cos \theta \quad (11)$$

and

$$C_R = \frac{-\Delta\dot{V}_o}{\Delta p_o \omega_o \sin \theta} \quad . \quad (12)$$

Therefore, respiratory resistance can be obtained by computing the ratio of pressure perturbation magnitude to flow perturbation magnitude of the in-phase component. The exact magnitude of the external resistance need not be known.

Device Description

The APD is formed from a pneumotach with flow straightening vanes (Fleisch #2 or #3; Gould, Biltoven, Netherlands) to measure airflow rate, pressure transducer (Validyne DP15; Northridge, CA) to measure mouth pressure, and a rotating wheel containing openings alternately open and covered with 400 mesh stainless steel wire cloth (Figs. 2 and 3). A Heath (Benton Harbor, MI) H-11A microcomputer is presently used to collect and analyze the data (Hochheimer, 1982).

A sample annotated strip chart recording of pressure and flow waveforms while breathing through the APD is shown in Fig. 4. Perturbations at 6 cps are superimposed upon exhalations and inhalations. Pressure perturbations are typically a larger proportion of the breathing wave than flow perturbations (Johnson and Lin, 1983A), and thus are easier to estimate.

Preliminary Experiments

Experiments were performed to investigate APD characteristics and to determine the experimental realizability of the device theory. At this point, measurements which were at least consistent with previously established physiological ob-

servations would demonstrate the merit of the APD and justify further development.

Of central importance is the depth of penetration of perturbations within the respiratory system, because the APD cannot measure resistance

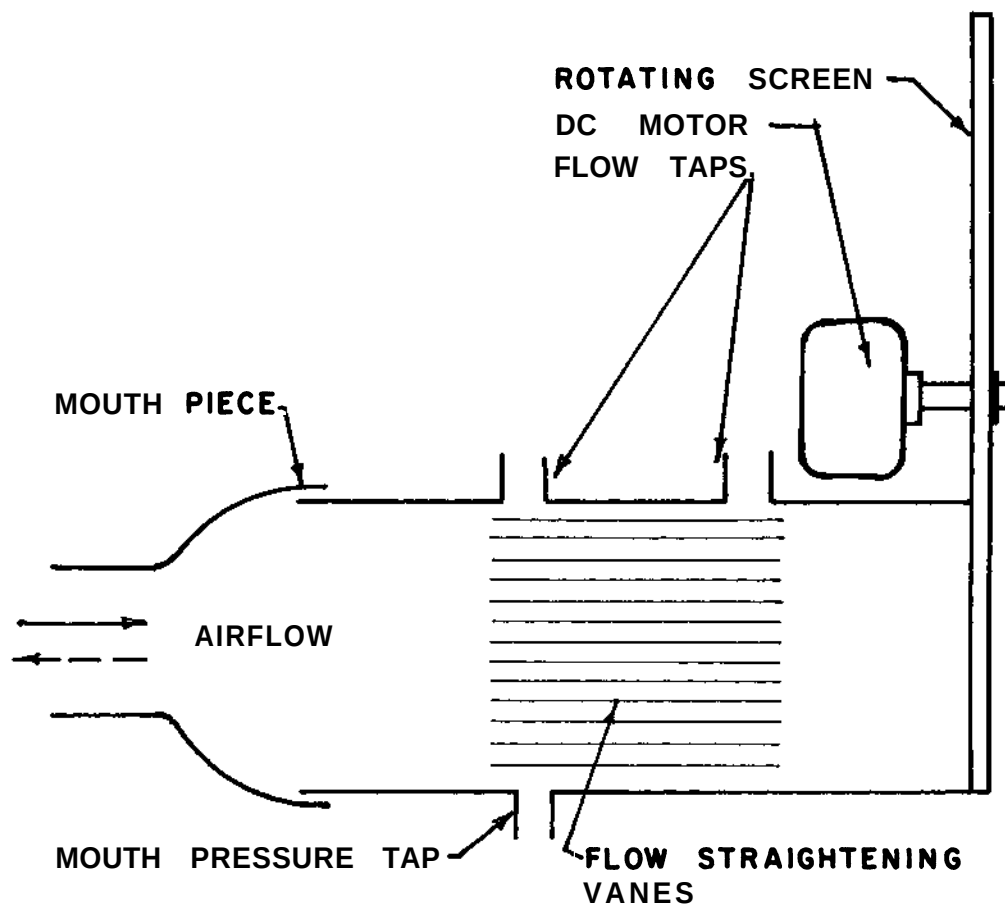


Figure 2. Schematic diagram of the APD.

beyond the point where the perturbations disappear. For this purpose, seven swine (females and castrated males 70-115 kg) were lightly anesthetized with intravenous infusion of Nembutol (Sodium pentobarbitol) at a rate of 25 mg/kg body mass. Pigs were laid on their sides on the operating table and their facial hair removed with a depilatory. The APD was fixed to a large semirigid cone which was placed over the face and sealed to the pig (with peanut butter!). While the APD was operating, a 1 mm diameter cannula which had been passed with the aid of a bronchoscope from the mouth into the respiratory airways was slowly and carefully pushed in further. Between periods of movement, the cannula was flushed with air from a hypodermic syringe to assure that the end was open. Pressure measurements were made while the external end of the cannula was connected to a Validyne DP15 pressure transducer with 0.1 psi (6.9 N/m^2) diaphragm and displayed on a Validyne CD12 transducer indicator. Maximum resolution of the pressure-measuring system

was less than $\pm 0.07 \text{ N/m}^2$. Within that resolution, noticeable perturbations were obtained until the region of the twelfth bifurcation, as estimated from the length of the inserted cannula and its diameter.

In a second experiment, five human subjects (males, aged 24-31, reasonably fit) were asked to breathe through the APD while sitting and watching an indicator breath flow rate. Each subject was asked to maintain his flow rate nearly constant during each breath and to breathe at three different flow rates. No attempt was made to hold the cheeks with the hands or to correct for lung volume changes. Peak-to-peak data was removed from the charts by hand and at least ten samples were taken during periods of constant flow rate. Results appear in Table 1.

In a third experiment, five human subjects (3 males, 2 females, aged 20-23, fit) were asked to pedal on a bicycle ergometer at constant rates of work which would cause them to perform for 15-20 minutes before they reached their voluntary

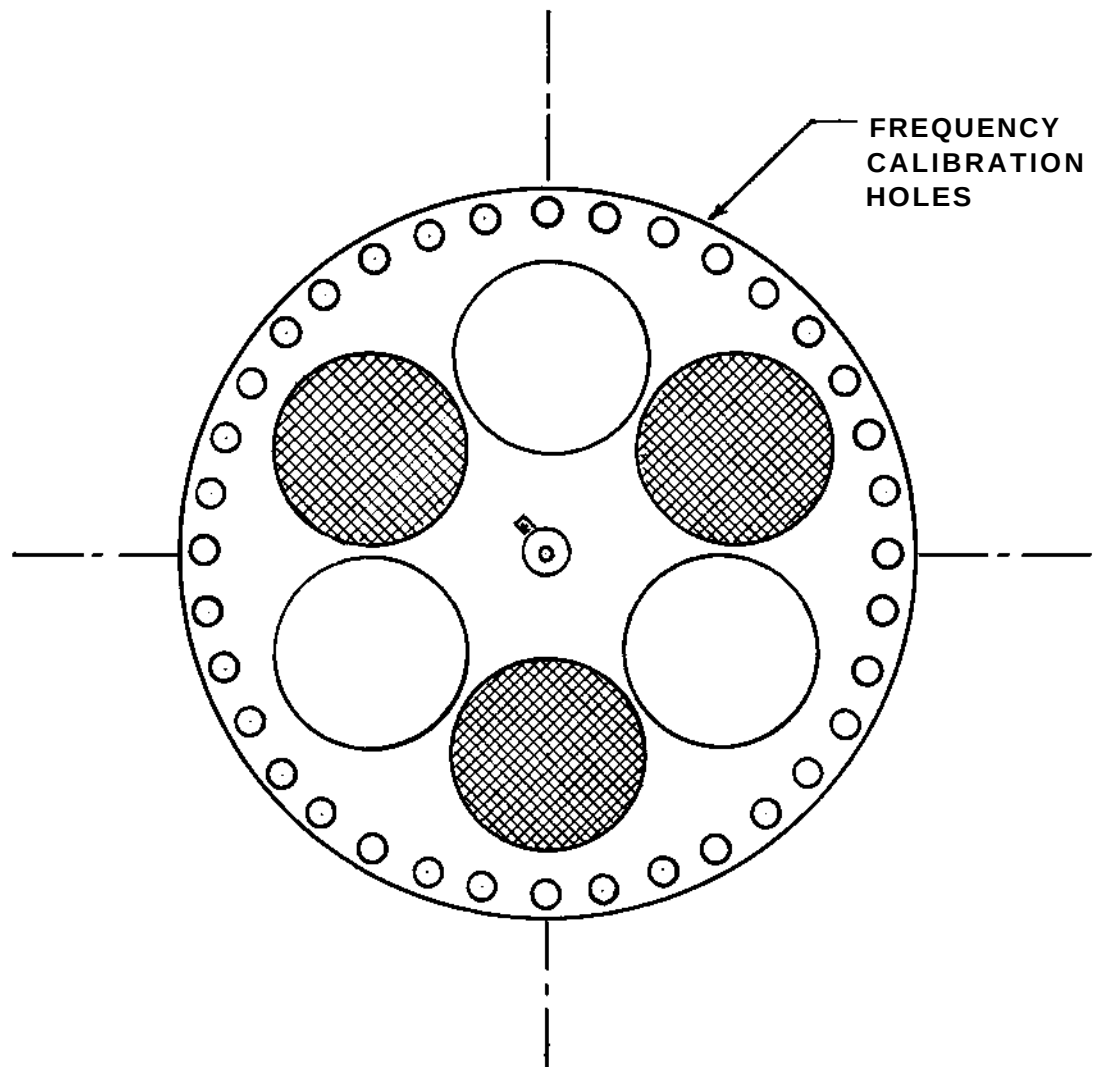


Figure 3. Schematic diagram of the perturbation generator wheel.

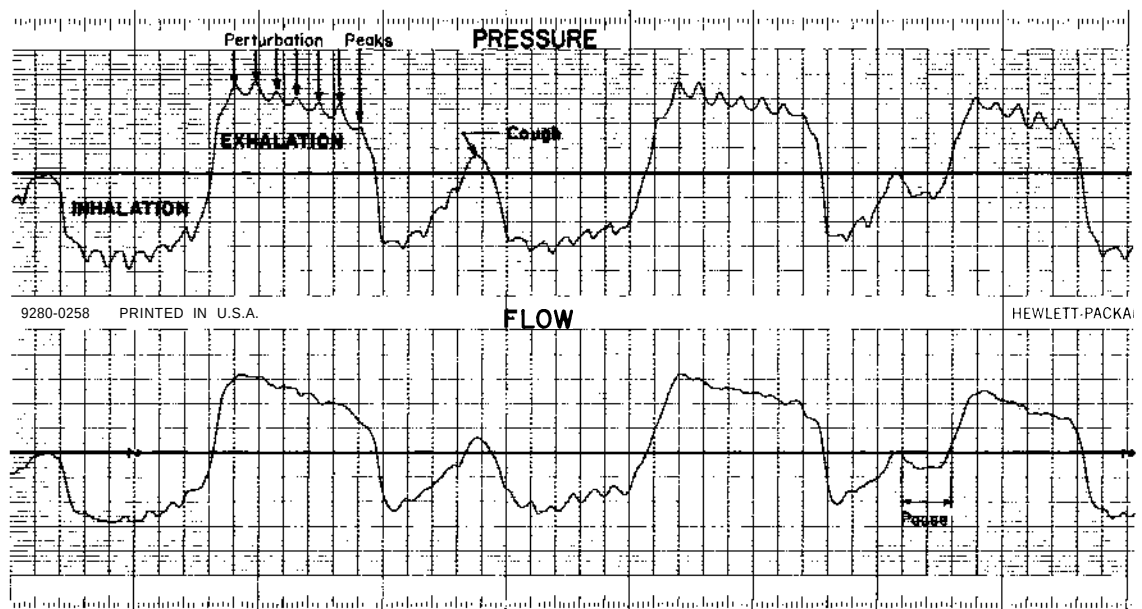


Figure 4. Strip-chart recording of APD waveforms.

Table 1. Inhalation and exhalation resistance of the airways of human subjects.

Subject		Airway resistance (cmH ₂ -sec/L)		Flow rate (L/sec)
		Mean	Standard deviation	
H.B.*	(EX)	1.03	0.24	0.92
	(IN)	0.85	0.23	
	(EX)	1.39	0.30	1.67
	(IN)	1.10	0.21	
	(EX)	1.86	0.24	3.33
	(IN)	1.53	0.15	
O.B.	(EX)	1.00	0.27	0.92
	(IN)	0.74	0.15	
	(EX)	1.23	0.12	1.67
	(IN)	0.95	0.15	
	(EX)	1.55	0.27	3.33
	(IN)	1.05	0.12	
A.J.	(EX)	0.92	0.17	0.92
	(IN)	0.62	0.05	
	(EX)	1.22	0.20	1.67
	(IN)	1.03	0.14	
	(EX)	1.47	0.22	3.33
	(IN)	1.23	0.24	
M.L.*	(EX)	1.40	0.24	0.92
	(IN)	1.22	0.12	
	(EX)	1.67	0.36	1.67
	(IN)	1.36	0.21	
	(EX)	2.12	0.32	3.33
	(IN)	1.71	0.23	
E.M.	(EX)	1.24	0.19	0.92
	(IN)	0.76	0.16	
	(EX)	1.44	0.29	1.67
	(IN)	1.20	0.26	
	(EX)	2.16	0.43	3.33
	(IN)	1.72	0.35	

* Designates smoker.

(EX) refers to exhalation and (IN) to inhalation.

end-points (Johnson, 1976). After a familiarization session, subjects underwent three measured test periods: one while breathing through the APD, another while breathing through the APD and external resistance (a screen which fit inside the APD mouthpiece and which had a pressure drop of 5 cm H₂O at 85 L/min), and a third while wearing an air purifying mask [MSA (Pittsburgh, Penn.) full facepiece with Ultra-Filter Cartridge, #464807] with APD attached to the exhalation valve by means of a length of tubing. Subjects

performed for as long as they were able, for each condition, and conditions were not replicated. Recordings of mouth pressure and flow were made while the subjects exercised. As with the previous experiment, all data was calculated by hand from these charts. Typical results for the three conditions appear in Figs. 5, 6, and 7. Because data was obtained manually from the charts, and because of the preliminary nature of these tests, resistance is presented in relative units only.

Discussion

The swine results indicate that the respiratory model of Fig. 1 is of limited use for APD measure-

ments. Compliance is not localized, but is distributed throughout lung tissue, and acts to dampen

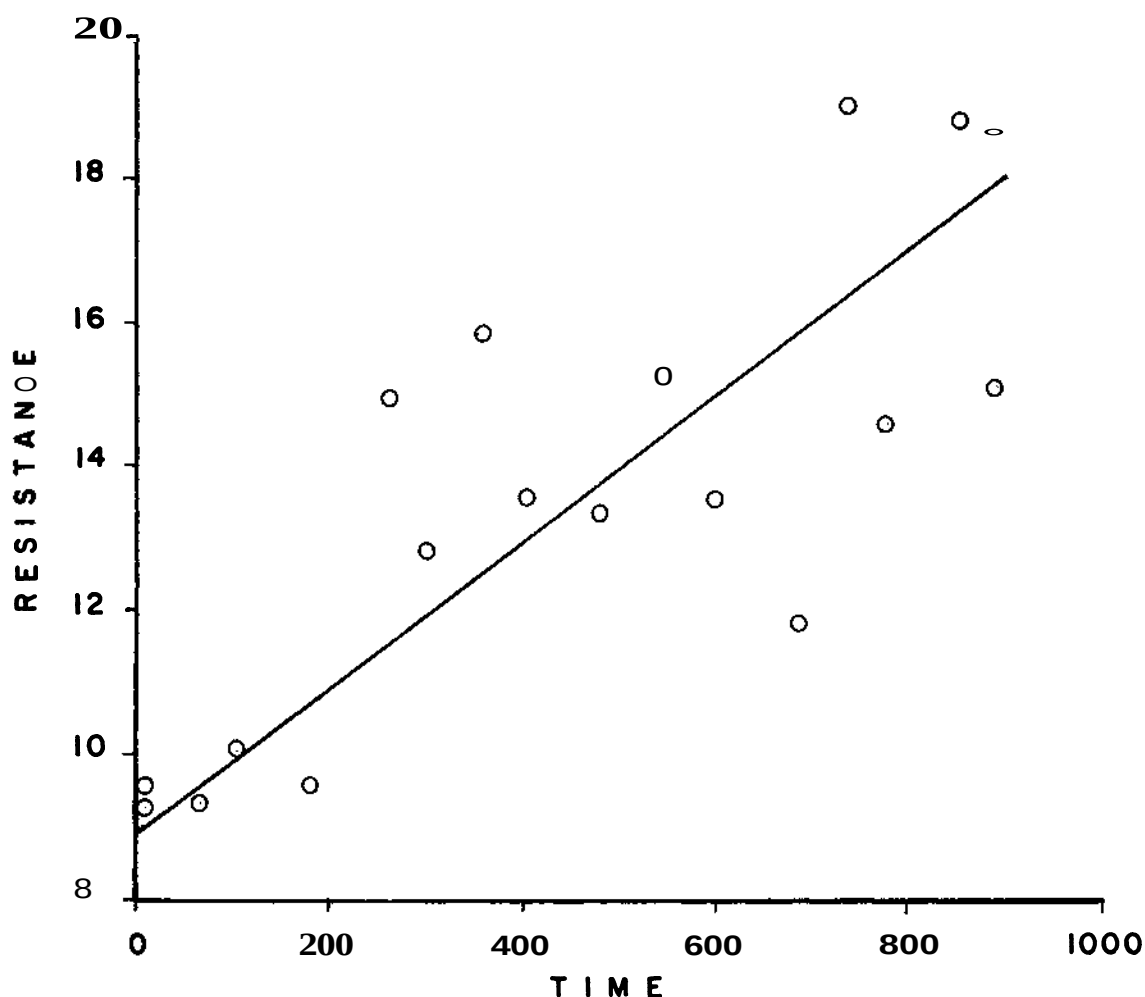


Figure 5. Resistance measurements during bicycle ergometer exercise of a subject (RB) breathing through the APD. Time is in seconds elapsed since the start of testing. Resistance units are relative only. The line represents an estimate of the overall trend.

induced perturbations much above the alveolar level. Wherever the perturbations disappear, pressure remains constant as far as the APD is concerned. Thus the muscle pressure appearing in Eqs. (3) and (4), and which disappears in Eq. (5), should be changed to be pressure closest to the mouth without perturbations. The APD measures resistance from the mouth to the point of disappearance of perturbations. Therefore, the swine results indicate that the APD measures at least upper airway resistance. Because the diameter of the cannula prevented insertion much deeper, it is difficult to say whether lower airway resistance is also measured by the APD.

Lower airway resistance is greater than upper airway resistance for very quiet breathing and for certain pathological states. For normal individuals, and especially exercising normals, the nonlinearity of the pressure-flow characteristic of

the upper airway (Ferris et al., 1964) translates into upper airway resistance much higher than lower airway resistance. Therefore, even if the APD is limited to only upper airway resistance measurement, we do not interpret the limitation to be severe.

A perturbation frequency of 6 cps was chosen partly because it is in the range of the natural frequency of the respiratory system (DuBois et al., 1956). At the natural frequency, compliance effects cancel with inertance effects, and impedance equals resistance. Typical APD tracings for healthy animals and humans show very little, if any, phase angle between pressure and flow perturbations. This would indicate that compliance is not a major contributor to the APD measurements.

Results from the sitting subjects are consistent with physiological expectations. First, values are in the range of 1.0 cm H₂O-sec/L, as expected

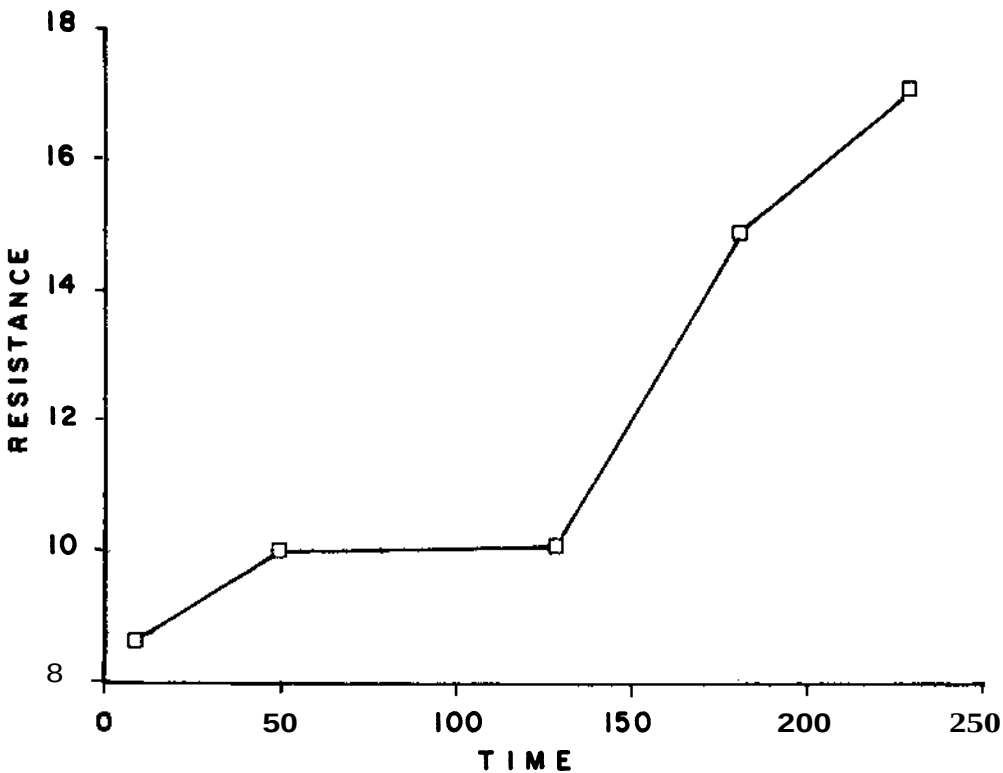


Figure 6. Resistance measurements during bicycle **ergometer** exercise of subject EK breathing through the APD and added resistance. Time and resistance scales are relative.

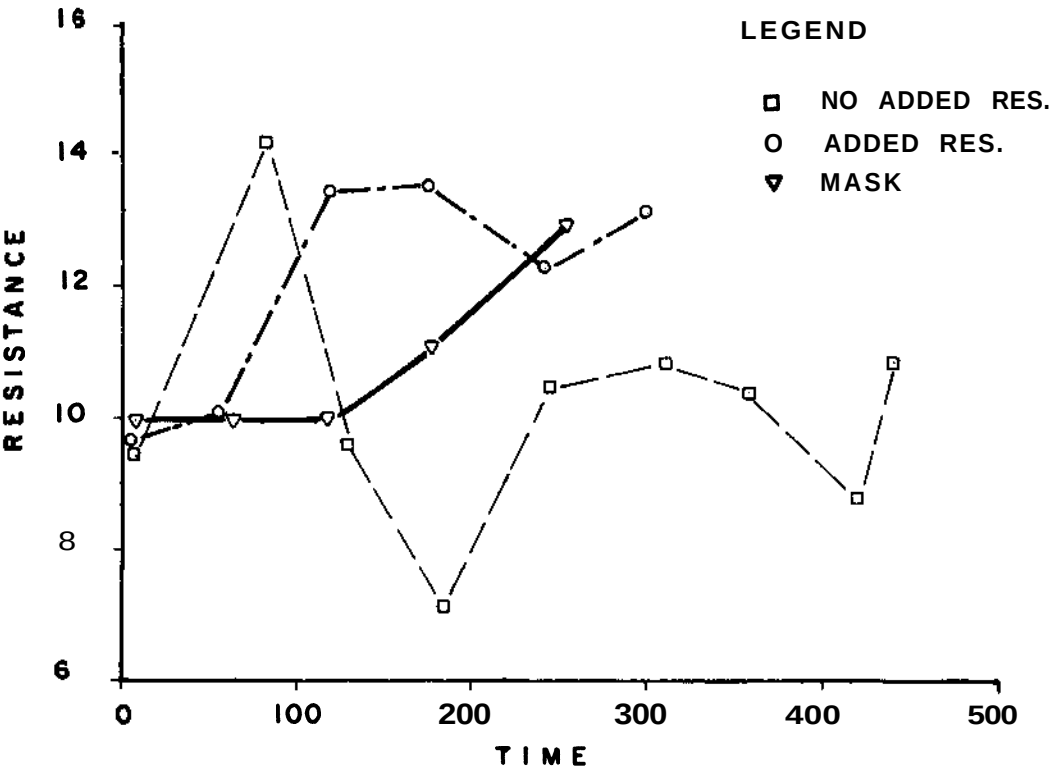


Figure 7. Resistance measurements during bicycle ergometer exercise of subject JSC for all three breathing conditions. Time and resistance scales are relative.

(Ferris et al., 1964). Second, exhalation resistance measurements are higher than inhalation resistance measurements. This trend has been reported many times (Ferris et al., 1964; Bouhuys and Jonson, 1967), and is explained by the direction of the transmural pressure acting on the airways. Third, the sitting subjects results show an increase in resistance measurements as flow rate increases. This trend was recognized long ago by Rohrer (1915) and Ainsworth and Eveleigh (1952). Both of these observations, while they don't by themselves validate APD measurements, increase our confidence in the APD and its measurements.

Results from the exercising subjects show the increase in resistance that would be expected from exercising subjects. As exercise progresses, flow rate increases, and, as flow rate increases, resistances of both human (Mead, 1961) and mask (Johnson, 1976) increase.

It is indeed unfortunate that the variation in data presented in Figs. 5, 6, and 7 is so great, and

also that resistance values for the same figures are not presented in absolute terms. Both difficulties are the result of the hand calculation procedure, and cannot be corrected without an unduly large effort. We roughly estimated that actual resistance of the airways (which should begin at about 100 kN-sec/m², or 1 cm H₂O-sec/L) increased between 5- and 10-fold during the test. Because of the difficulties with manual data analysis, it is doubtful that the APD would ever find wide use without computer data analysis. We are presently developing the procedure to accomplish this, and results thus far are very encouraging.

With further development and more experimental data, the APD can become an important instrument for measuring respiratory stress in the workplace. Its noninvasive features and computer operation will enable a wide range of personnel to take respiratory resistance measurements confidently.

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