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THE ARMY EMERGENCY RESPIRATOR

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U.S. ARMY MATERIEL COMMAND

HARRY DIAMOND LABORATORIES
WASHINGTON, D.C. 20438

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by

J. W. Joyce, Jr.

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ABSTRACT

This report describes the Army Emergency Respirator, a small pressure-cycled respirator that uses a fluid amplifier as its basic component. The calibration technique applied to achieve the desired functional operation is presented. To determine performance capability of the respirator, engineering tests were performed using tanks to simulate lung compliances and a perforated disk to simulate airway resistance. From these tests, the switching pressures, cycling rates, tidal and minute volumes, and gas consumption of the respirator are defined quantitatively for given load conditions. The engineering tests, along with some earlier medical tests, suggest that the respirator can provide adequate ventilation for patients.

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ILLUSTRATIONS

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1. INTRODUCTION

The Army Emergency Respirator (fig. 1) is a joint development of
the Harry Diamond Laboratories (HDL) and the Walter Reed Army Institute
of Research (WRAIR).[1]   This device is pressure cycled and can either
assist or control the ventilation of a patient. This small
(5 by 10 by 15 cm), lightweight (250 gm) respirator consists of a
bistable fluid amplifier, a flow control valve, and a breathing valve
(fig. 2). The fluid amplifier provides the pressure-cycling function,
the flow control valve regulates the minute volume delivered to the
patient, and the breathing valve allows exhalation to take place easily
and rapidly. The entire unit contains only three moving parts (all in
the breathing valve).

The Army Emergency Respirator described in this report is con-
sidered to be an advanced developmental design. Twenty-five of these
units have been produced for experimental evaluation (medical and
engineering) purposes. Because this respirator has been designed for
emergency use and has only one control to regulate ventilatory function,
it has been calibrated to satisfy the needs of the average patient.
Therefore, it will not necessarily find application in those instances
where patient needs vary widely from the average or where functions
other than those designed into the respirator are required.

2. OPERATIONAL DESCRIPTION

The basic part of the Army Emergency Respirator consists of two
molded plastic plates bonded together. One plate contains the fluid
amplifier channels; the other plate serves to seal these channels. A
schematic diagram of the respirator is shown in figure 3. A description
of one complete respiratory cycle follows.

Breathing gases are supplied to the fluid amplifier through the
flow control valve. As these gases issue from the power nozzle of the
fluid amplifier, the resulting jet attaches to the left wall and flows
out of the left receiver (fig. 3a). Jet attachment to the left wall is
due to limited entrainment from the left control nozzle. Gases flow-
ing from the left receiver pass through the breathing valve and into
the patient's lungs, increasing the pressure in the lungs. As the
pressure increases in the lungs (and hence, in the left receiver of
the amplifier), flow through the left feedback line also increases.
Eventually the pressure in the left receiver is great enough to cause
the feedback flow in the left control nozzle to be sufficient to switch
the power jet from the left wall to the right wall. This terminates the

1. Straub, H. H., "Design Requirements and Proposal for Army
Respirators," TR-1249, Harry Diamond Laboratories, June 1964.

inspiratory phase of the respiratory cycle. The pressure at which switching occurs depends on the setting of the calibration setscrew in the left control feedback line.

When the jet has been switched to the right wall, the gases flow out the right receiver, and simultaneously the breathing valve allows the patient to exhale to the surroundings (fig. 3b). As the jet flows out the right receiver, it entrains gas from the left receiver. To satisfy this entrainment need, a small flow of gas is allowed to pass back through the breathing valve. Without this flow back through the breathing valve, the pressure in the left receiver would drop too fast, and the expiratory phase of the cycle would be inadequate. The entrainment process continues to drop the pressure in the left receiver (and in the patient's lungs) until the pressure reaches a given negative (below ambient) value. At this point, entrained flow through the right control nozzle is sufficient to switch the jet back to the left side, thus ending the expiratory phase and beginning a new cycle. The negative pressure at which switching occurs is determined by the setting of the calibration setscrew in the right control line.

As previously explained, the flow control valve regulates the input pressure to the fluid amplifier and in so doing determines the output of the respirator. Increasing input pressure causes both switching pressures to increase in magnitude and also increases inspiratory flow.

The breathing valve was added to the respirator as a result of preliminary medical evaluation at WRAIR of an earlier prototype that did not include such a valve. These prototypes exhibited high expiratory resistance, which was considered undesirable by the WRAIR medical team. The addition of the breathing valve allows the patient to exhale freely to the surroundings until the face mask pressure returns to ambient. Once ambient pressure is reached, the entrainment through the breathing valve produced by the fluid amplifier causes the pressure to continue to decrease but at a slower rate. This is illustrated by the pressure curve schematically represented in figure 4, which shows that the expiratory phase consists of a rapid decay to near ambient, followed by a slower decay to reach the final switching point.

3. TEST EQUIPMENT AND PROCEDURE

To evaluate this respirator, at least on an engineering basis, it is necessary to determine its performance capabilities. These capabilities include switching pressures, cycling rates, tidal and minute volumes, and gas flow consumption. These are the parameters of interest to personnel involved in using respirators. To determine

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these parameters, it was necessary to simulate lung compliance and airway resistance. Lung compliance was simulated by tanks for which the pressure-volume relationships were experimentally established by injecting known volumes of air into tanks and measuring the corresponding pressure increases. Compliances used ranged from 0.010 to 0.164 ·/cm H_2O.

Airway resistance was simulated by a perforated disk that was experimentally calibrated. It has been shown [2] that the pressure drop across the disk can be expressed as

$$\Delta P = k_1 Q + k_2 Q^2 \tag{1}$$

where

ΔP = pressure drop across disk, cm H_2O

Q = flow through disk, ℓ/sec

k_1 and k_2 are constants

Equation (1) indicates that the pressure drop consists of a laminar loss ($k_1 Q$) and a turbulent loss ($k_2 Q^2$). Dividing equation (1) by the flow rate Q yields

$$\frac{\Delta P}{Q} = k_1 + k_2 Q \tag{1a}$$

If airway resistance is defined as pressure drop per unit of flow, calibrated at a flow of 1 ℓ/sec, then from equation (1a) for $Q = 1$ ℓ/sec,

$$R = k_1 + k_2 \tag{2}$$

where

R = airway resistance, cm H_2O/ℓ/sec

For the perforated disk used, $k_1 = 3.8$ and $k_2 = 26.9$; therefore the disk represents an airway resistance of 30.7 cm H_2O/ℓ/sec.

A transducer (±1psi range) was used to sense pressure in the tanks. The output of the transducer was displayed on a storage oscilloscope. Because of slight variations in performance from cycle to cycle, six or more cycles were displayed for any given set of operating conditions, and their average used for the data point in question.

4. CALIBRATION PROCEDURE

As mentioned in the operational description, the settings of the calibration setscrews determine the positive and negative switching pressures. In addition, there is a third setscrew (not shown in

2. Joyce, J.W., "Revised Performance Evaluation of the Army Volume-Cycled Respirator, Model 2," TM-68-17, Harry Diamond Laboratories, July 1968.

fig. 3) located in the right receiver to regulate the duration of the expiratory portion of the breathing cycle. Each respirator was calibrated in the following manner.

The respirator is connected to a 76-l tank (representing a compliance of 0.053 l/cm H_2O). The setscrew in the left feedback line is adjusted first to obtain maximum positive switching pressure. This adjustment is made at an input pressure of 140 cm H_2O (2 psig).

The setscrew in the right feedback line is then adjusted so that negative switching pressures of 9 to 10 cm H_2O are obtained for an input pressure of 280 cm H_2O (4 psig).

Finally the setscrew in the right receiver is adjusted to produce an expiratory time approximately twice the inspiratory time. This adjustment is made at an input pressure of 210 cm H_2O (3 psig).

The entire calibration procedure was based on experimental performance data from earlier prototypes of the respirator. Units that could not have their setscrews adjusted to meet the above operating conditions were rejected. A total of 25 respirators were successfully calibrated; three rejects were encountered. After each unit was properly calibrated, the setscrews were sealed in place with a potting compound to prevent subsequent readjustment.

5. PERFORMANCE RESULTS

To obtain representative data, six respirators were selected at random from the 25 successfully calibrated units. Each of these six respirators was tested against eight different combinations of compliance and airway resistance. The compliances used had values of 0.010, 0.026, 0.053, and 0.164 l/cm H_2O. The two values of resistance used were 30.7 cm H_2O/l/sec and zero. Input pressure was varied from 70 to 350 cm H_2O (2 to 5 psig) for each load condition and respirator.

5.1 Switching Pressures

The switching pressure characteristics of the Army Emergency Respirator are shown in figure 5. These curves represent the average performance for the six respirators operating against the four different compliances. The results show that the positive and negative switching pressures vary linearly (or very nearly so) with input pressure over the range of pressures tested. The positive switching pressures increase in magnitude more rapidly than do the negative switching pressures.

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Table I shows the maximum differences in switching pressures for a given input pressure. obtained by testing with various values of compliance. The maximum differences generally occurred using the extremes of compliance, with the greater magnitude values occurring in the smallest compliance.

TABLE I. Maximum difference in switching pressures due to compliance (cm H_2O)

P_{in}	ΔP^+	ΔP^-
140	1.5	1.0
175	2.0	2.5
210	3.0	1.5
245	3.0	2.5
280	3.0	3.0
315	3.5	3.0
350	3.5	4.0

P_{in} = input pressure

ΔP^+ = maximum difference in positive switching pressure

ΔP^- = maximum difference in negative switching pressure

All pressures rounded to nearest 0.5 cm H_2O

Table II shows the maximum differences in switching pressures occurring in the six respirators tested for each compliance. These differences are of the same order of magnitude as those in table I. The smallest values (table II) were observed for the 0.053-l/cm H_2O compliance--the compliance used to calibrate the respirators.

5.2 Cycling Rates

The average cycling rates for the six respirators tested are shown in figures 6 and 7. The results show that cycling rate decreases as (1) compliance increases; (2) input pressure increases; and (3) airway resistance increases (at least for values of resistance used). Cycling rate decreases as compliance increases because the pressure builds up slower in the larger compliances than in the small ones. Also the time required to reach the negative switching pressure is increased with increasing compliance.

Cycling rate is decreased by increasing input pressure to the respirator. As input pressure increases, both the flow rate and magnitude of switching pressures are increased. The cycling rate curves

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TABLE II. Maximum differences in switching pressures among respirators

Compliance (ℓ/cm H₂O)	Switching Pressure	Input pressure (cm H₂O)						
		140	175	210	245	280	315	350
.010	ΔP^+	2.0	2.5	2.5	3.0	3.5	3.5	4.0
	ΔP^-	1.0	1.5	1.5	2.0	2.0	4.0	4.0
.026	ΔP^+	1.5	1.5	2.0	2.5	2.5	3.5	3.5
	ΔP^-	1.0	0.5	1.0	1.5	1.5	3.5	5.0
.053	ΔP^+	1.0	2.0	2.0	2.5	2.5	3.0	4.5
	ΔP^-	0.5	1.5	1.0	0.5	1.5	3.0	3.5
.164	ΔP^+	2.0	2.0	3.0	3.0	4.0	4.5	5.0
	ΔP^-	1.0	1.0	0.5	1.5	2.5	3.5	5.5

ΔP^+ = maximum difference in positive switching pressures (cm H₂O)
ΔP^- = maximum difference in negative switching pressures (cm H₂O)
All pressures are rounded to nearest 0.5 cm H₂O

indicate that switching pressures are increasing faster than flow rate.

Finally, a comparison of figures 6 and 7 shows that cycling rate increases as airway resistance decreases for the two values of resistance used. The dominant factor in this increase in cycling rate is that the exhalation through the breathing valve to ambient conditions (the steep part of the expiratory phase, fig. 4) takes place more rapidly as airway resistance decreases.

5.3 Delivered Tidal Volumes

The average tidal volumes (calculated from pressures developed in the tanks) delivered to the two smallest compliances are shown in figure 8; the volumes delivered to the two largest compliances are shown in figure 9. These results show that tidal volume increases with increasing compliance and input pressure, and decreases with airway resistance.

The increase in tidal volume with increasing compliance and input pressure can best be explained by considering the definition of compliance as shown below.

$$C = \frac{\Delta V}{\Delta p}$$

where   (3)

C = compliance, ℓ/cm H_2O

ΔV = tidal volume, ℓ

Δp = change in pressure during inspiratory phase, cm H_2O

From equation (3), the delivered tidal volume is the product of compliance and pressure difference. Since the switching pressures remain virtually constant as compliance changes, Δp is constant and tidal volume varies directly with compliance. Similarly, for a given compliance, tidal volume varies directly with Δp. Since Δp increases with increasing input pressure, tidal volume likewise increases with increasing input pressure.

The increase in tidal volume with decreasing airway resistance occurs because the pressure drop across the resistance is smaller for the small resistance than for the large resistance. The respirator sees only the pressure ahead of the resistance. For the high resistance the pressure in the tank (which determines tidal volume delivered) is considerably less than the pressure ahead of the resistance (the switching pressure). For the small resistance the tank pressure is equal to the switching pressure, and therefore the delivered tidal volume is greater than for the case of the large resistance.

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5.4 Minute Volumes

The average minute volumes delivered to the various compliances are presented in figures 10 (no airway resistance) and 11 (maximum resistance). Minute volume is shown to increase with increasing input pressure and decrease with increasing airway resistance. Minute volume does not appear to change much with compliance, except in the case of the lowest compliance (0.010 ℓ/cm H_2O), for which minute volumes are less than for the other compliances.

The effects of input pressure and airway resistance on minute volume are determined by their effects on tidal volume and cycling rate, since minute volume is the product of these two parameters. For increasing airway resistance, both the tidal volume and cycling rate decrease for the resistances used in these tests so that minute volume must also decrease. As input pressure increases, tidal volume increases while cycling rate decreases. The minute volume results imply that tidal volume increases faster than cycling rate as input pressure is increased for the two values of resistance tested.

5.5 Air Consumption Rate

The air flows required by the respirator are shown in figure 12. These results represent the average for data taken from all 25 successfully calibrated respirators. The maximum flow consumed by the respirator is 45 ℓ/min at the upper-limit input pressure of 350 cm H_2O (5 psig). These flow requirements are determined by the area of the power nozzle of the fluid amplifier. A smaller nozzle area would reduce air consumption but would also reduce inspiratory flows to the patient. Similarly, if higher inspiratory flows were required, the air consumption would increase. Therefore, a balance between inspiratory flows desired and efficiency that can be tolerated must be achieved.

6. DISCUSSION AND CONCLUSION

Twenty-five Army Emergency Respirators have been fabricated and calibrated at HDL. Six of the units were tested extensively to determine the performance capabilities. The results indicate that the respirator can produce cycling rates and minute volumes suitable for patient use.[3] Medical tests on some earlier prototypes showed their ability to ventilate dogs adequately.[4] In these tests, the respirator was operated as both an assistor and a controller.

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3. Ibid

4. Straub, H. H. and Mosley, E., "A Respirator without Moving Parts," Proceedings of the 17th Annual Conference of Engineering in Medicine and Biology, 1964.

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In its present form, the Army Emergency Respirator offers small size, potential low cost, simplicity of design and operation, and reliability. Some of these respirators have been distributed for medical evaluation, although very little has been reported from the evaluators. One evaluator felt that the inspiratory flow rates might be a little low for assist applications, and that the unit was noisy. Other comments from evaluators have related to high oxygen comsumption, the lack of a pressure gauge on the respirator to tell the user what is happening, the lack of a suitable safety relief valve to preclude administering excessively high pressures to the patient, and the fact that a small leak in the patient circuit causes the respirator to stop cycling. This last problem is inherent in any pressure-cycled respirator, since it cannot cycle if it cannot build up enough pressure to reach the switching point. These and other comments bear serious consideration in the design of any subsequent prototypes of the respirator.

GLOSSARY OF MEDICAL TERMS

Airway resistance

The pressure difference across the airway (between the mouth and the lungs) per unit flow, usually expressed as centimeters of water per liter of flow per second (cm H_2O/l/sec). Since the resistance is nonlinear, the value obtained depends on the flow rate; consequently the resistance is usually given for a particular flow rate. In this report, the airway resistance is defined for a flow of 1 l/sec.

Assisted respiration

Artificial respiration which is synchronized with the inspiratory effort of the patient.

Compliance

The volume increase produced by a unit pressure increase in the lungs, usually expressed as l/cm H_2O.

Controlled respiration

Any form of intermittent artificial inflation of the lungs, but not necessarily synchronous with any respiratory effort of the patient.

Expiratory phase

That part of the respiratory cycle which includes exhalation and expiratory pause (period of complete respiratory inactivity).

Inspiratory phase

That part of the respiratory cycle during which the lungs are inflated.

Minute volume

The total volume of air delivered to the patient circuit per minute. It equals the product of tidal volume and breathing rate.

Patient circuit

That part of the respiratory circuit in communication with the patient's lungs, here considered to be everything to the patient side of the breathing valve.

Tidal volume

The amount of gas delivered to or breathed by the patient per breath.

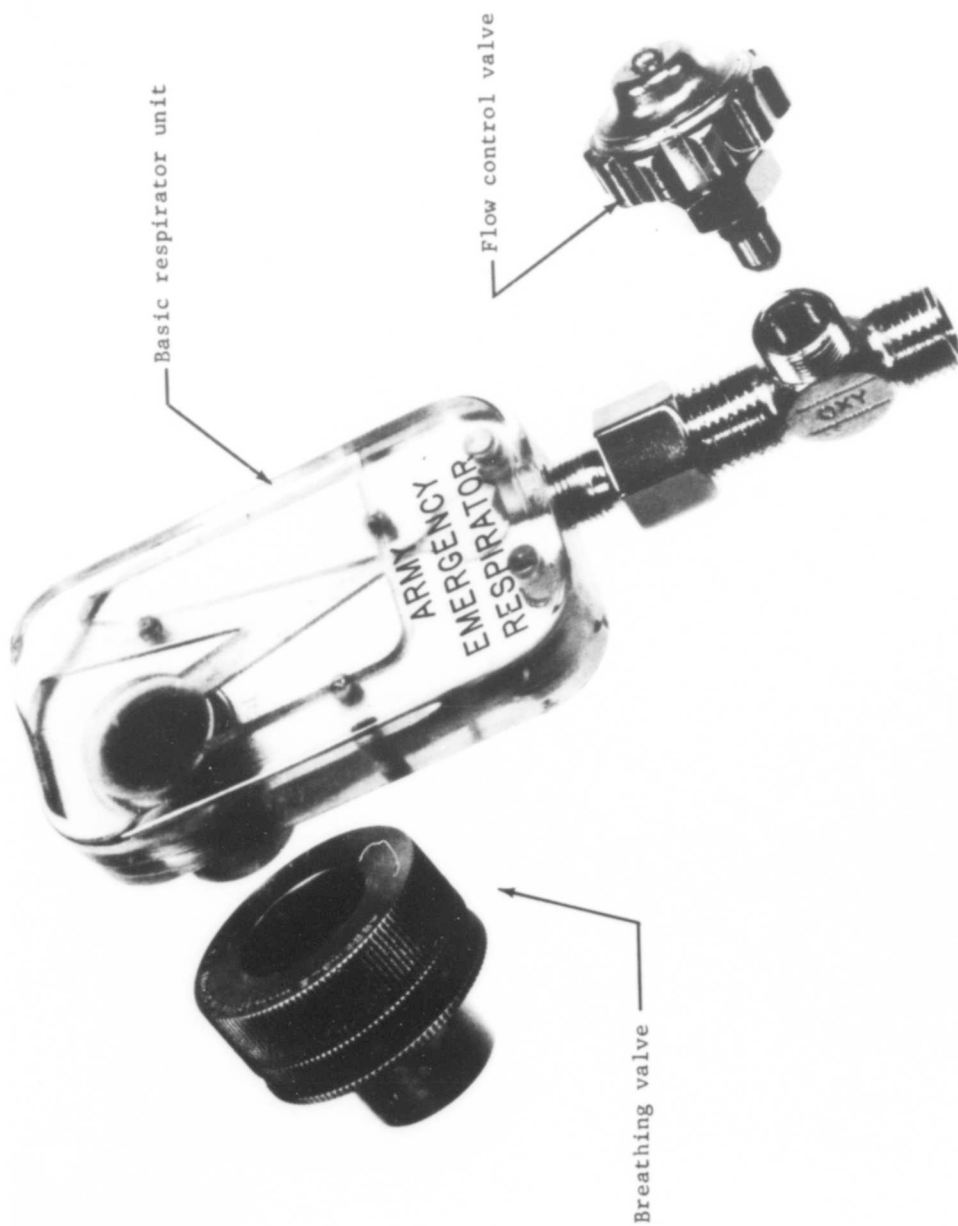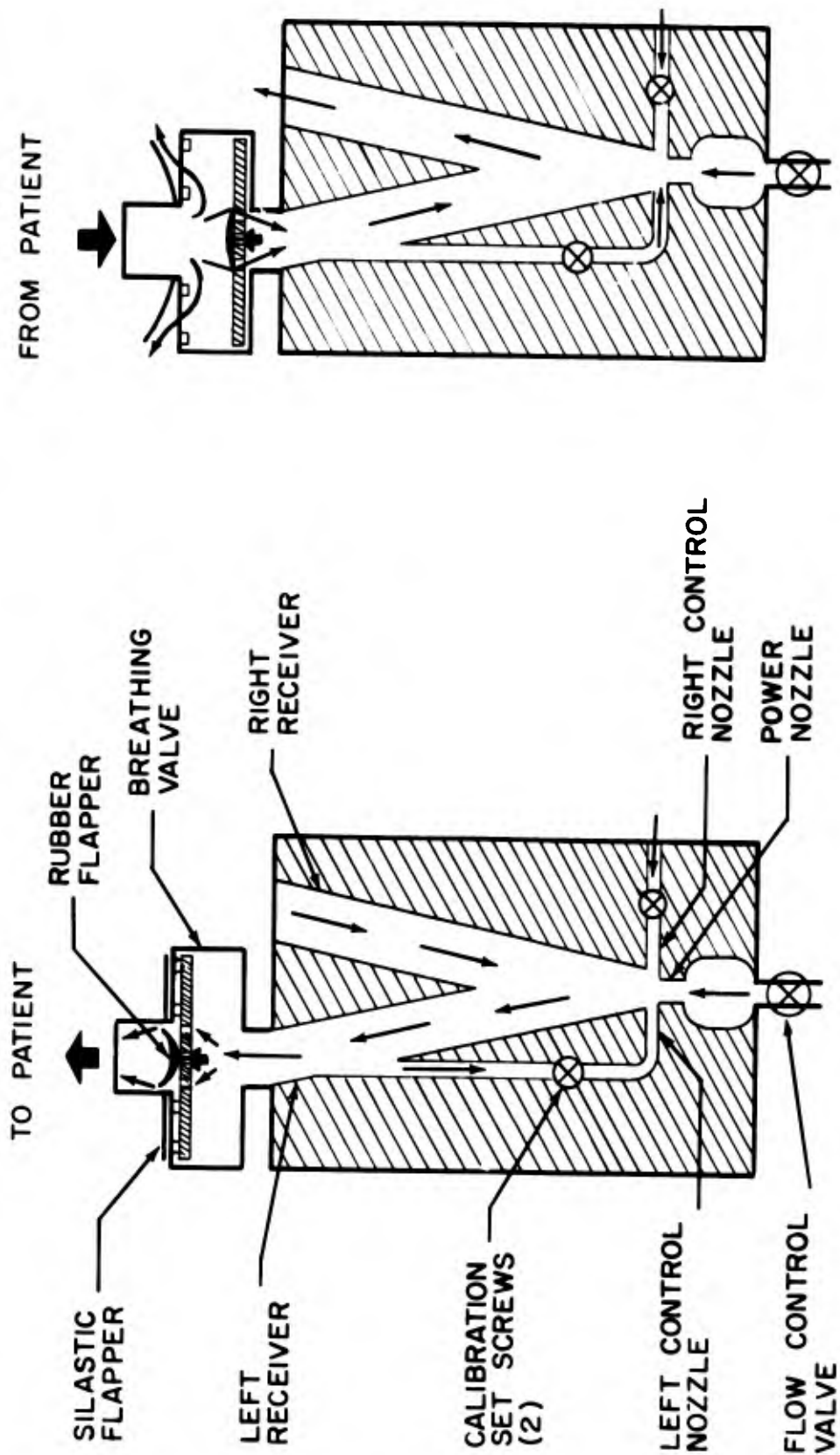631-66

Figure 1. Army Emergency Respirator.

Basic respirator unit

Flow control valve

ARMY EMERGENCY RESPIRATOR

OXY

Breathing valve

Figure 2. Army Emergency Respirator – exploded view.

FROM PATIENT

B. EXPIRATION

1154-66

RUBBER
FLAPPER

BREATHING
VALVE

RIGHT
RECEIVER

RIGHT CONTROL
NOZZLE

POWER
NOZZLE

TO PATIENT

SILASTIC
FLAPPER

LEFT
RECEIVER

CALIBRATION
SET SCREWS
(2)

LEFT CONTROL
NOZZLE

FLOW CONTROL
VALVE

A. INSPIRATION

Figure 3. Respirator schematic.

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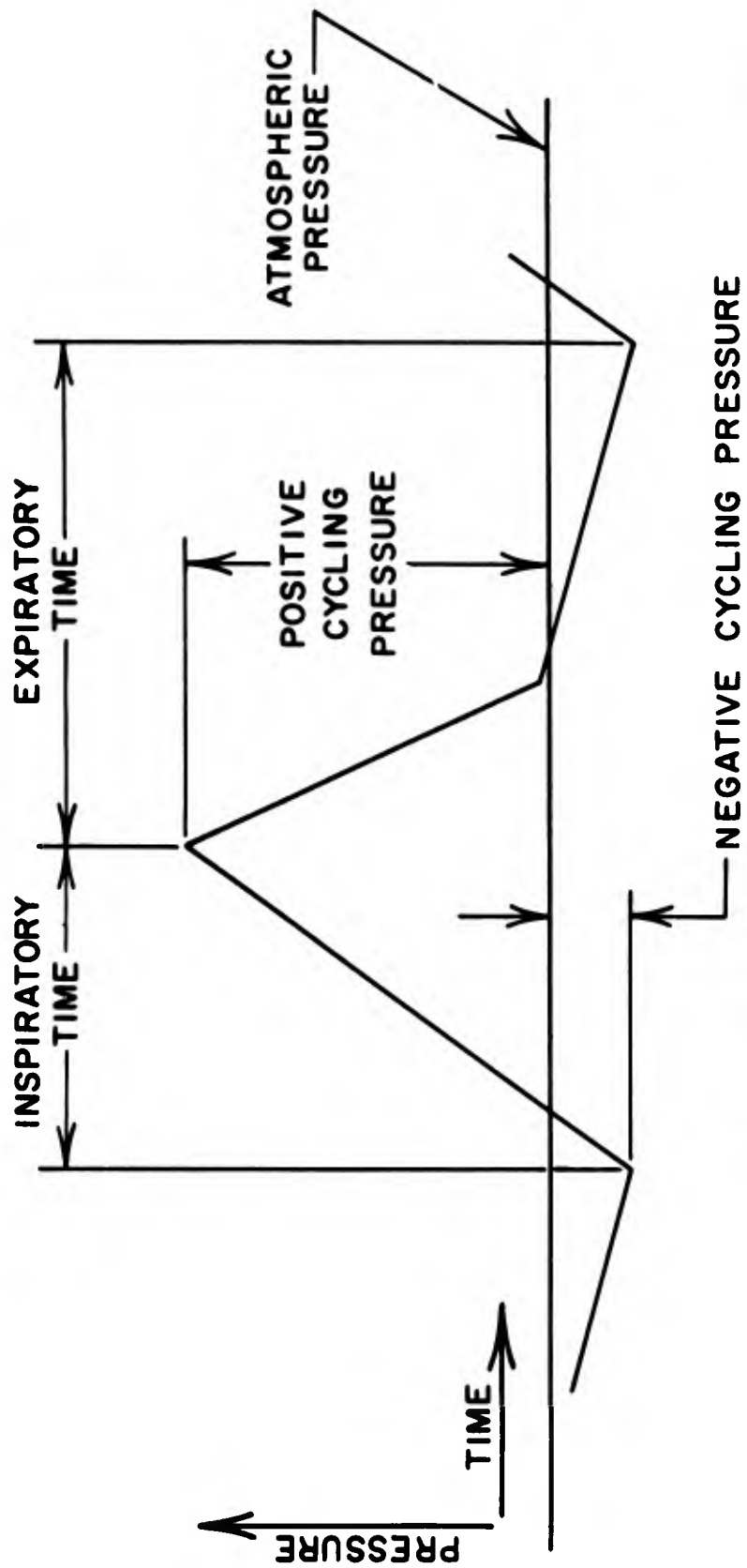
Figure 4. Typical pressure curve schematic.

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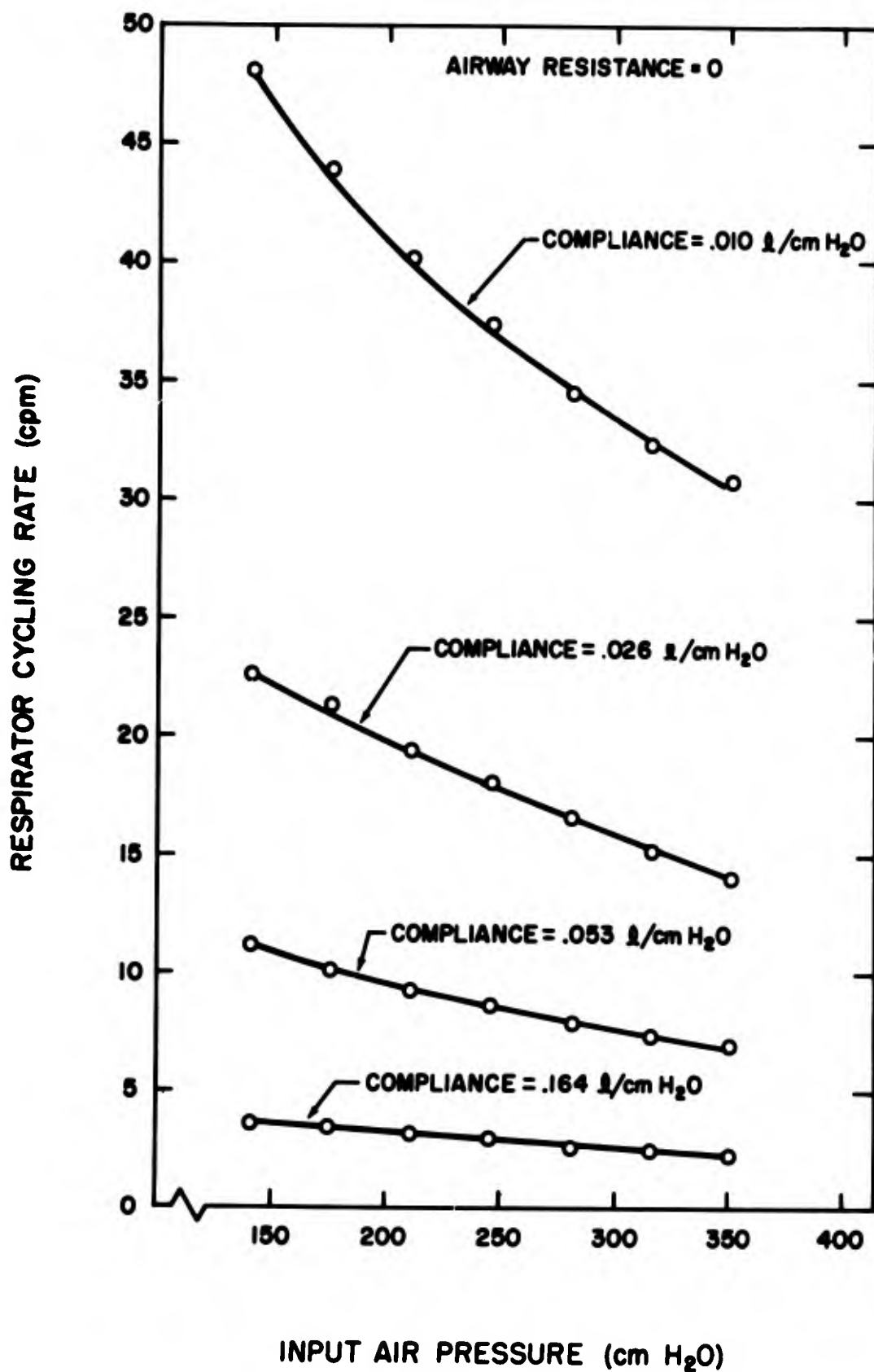20
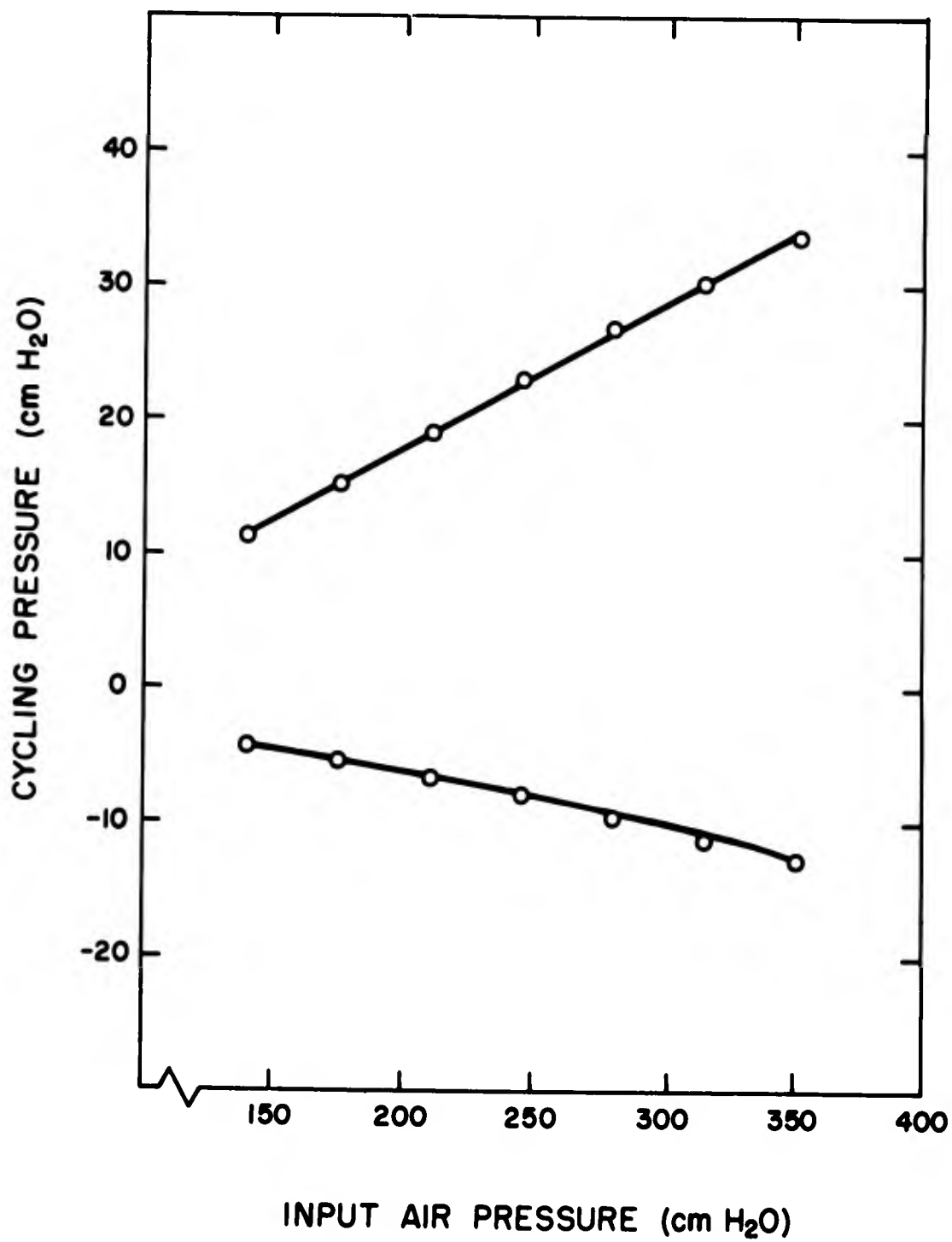
Figure 5. Switching pressure characteristics.   2045-68

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Figure 6. Cycling rates—no resistance.   2052-68

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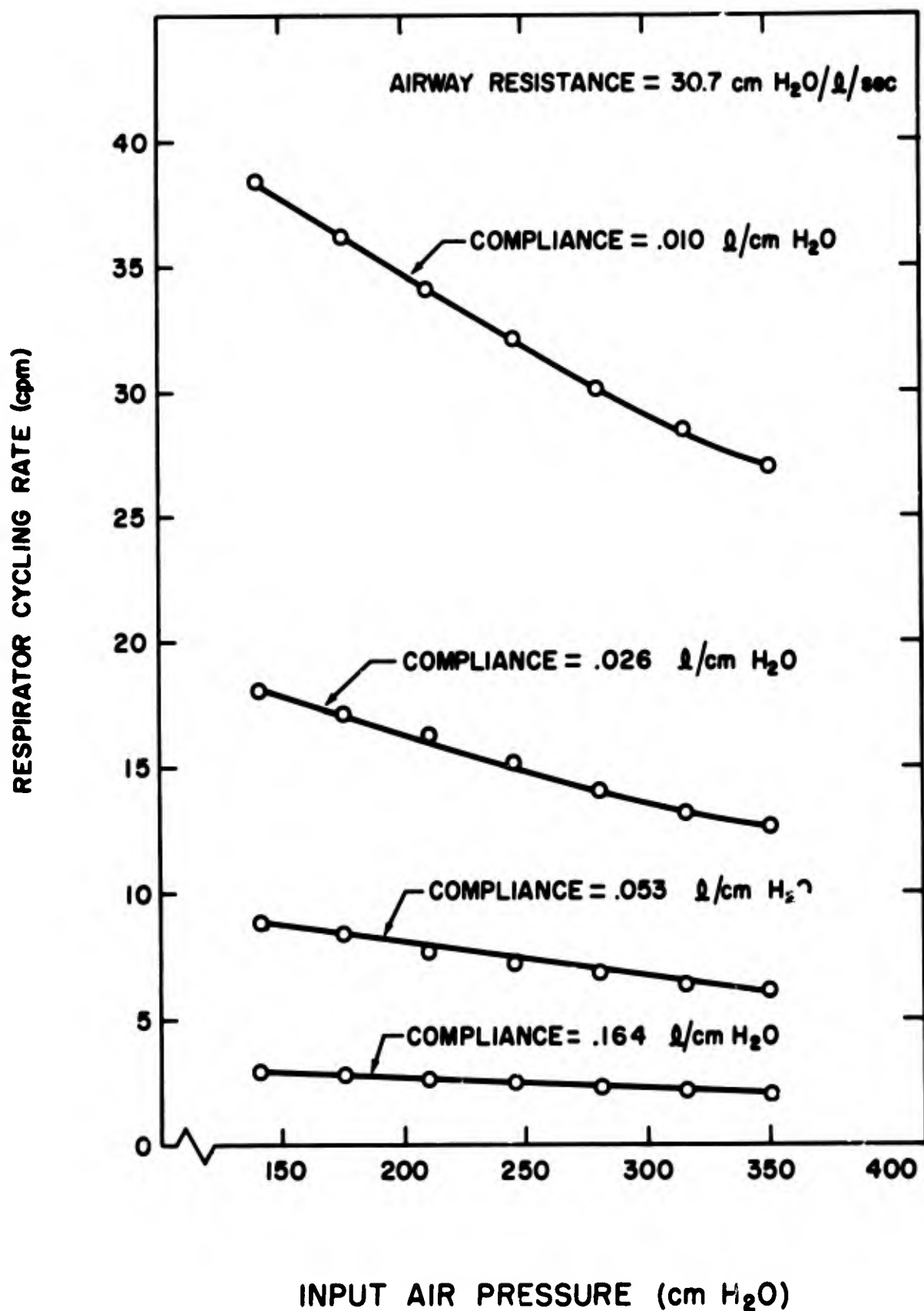**INPUT AIR PRESSURE (cm H₂O)**

Figure 7. Cycling rates—maximum resistance.
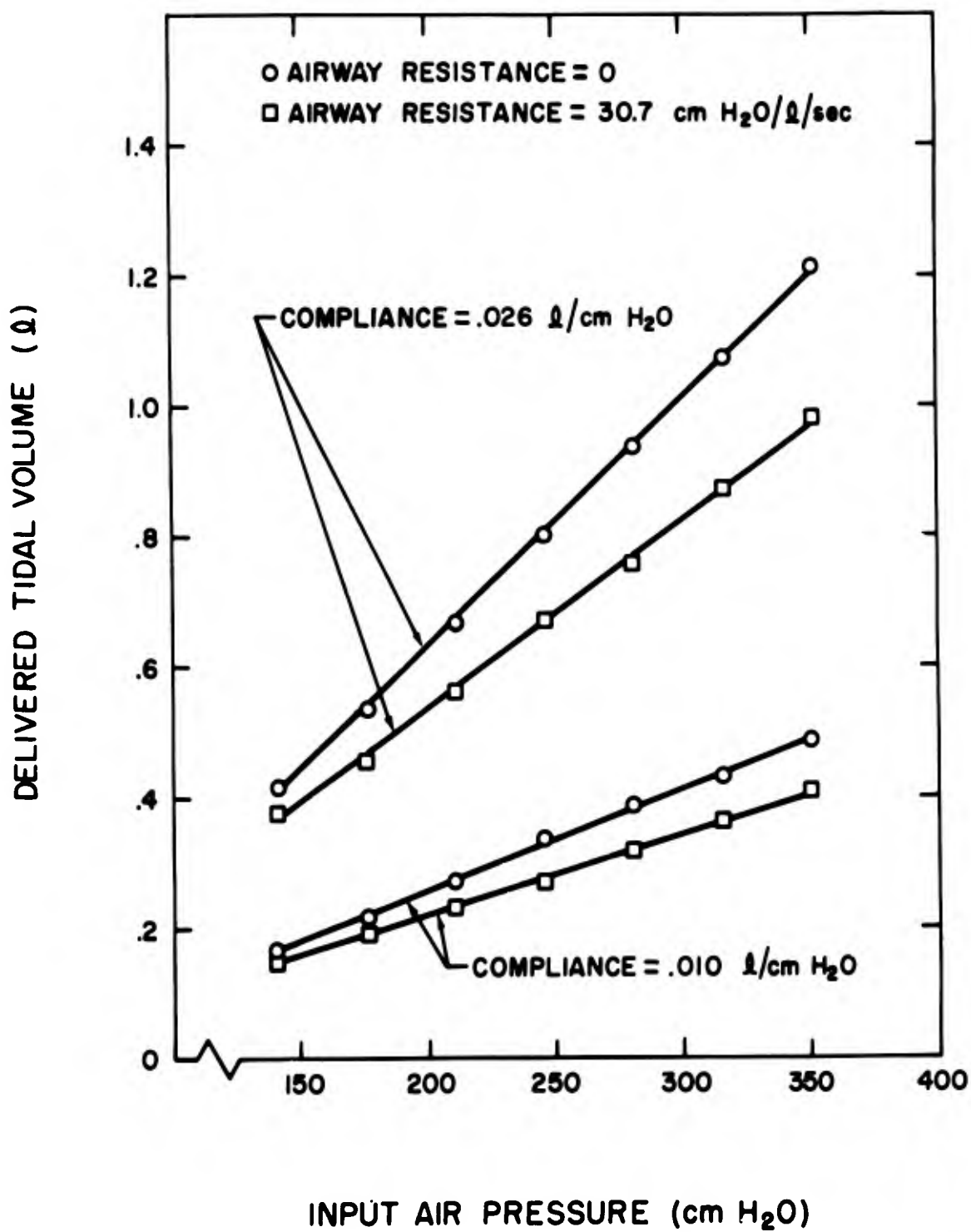
2048-68

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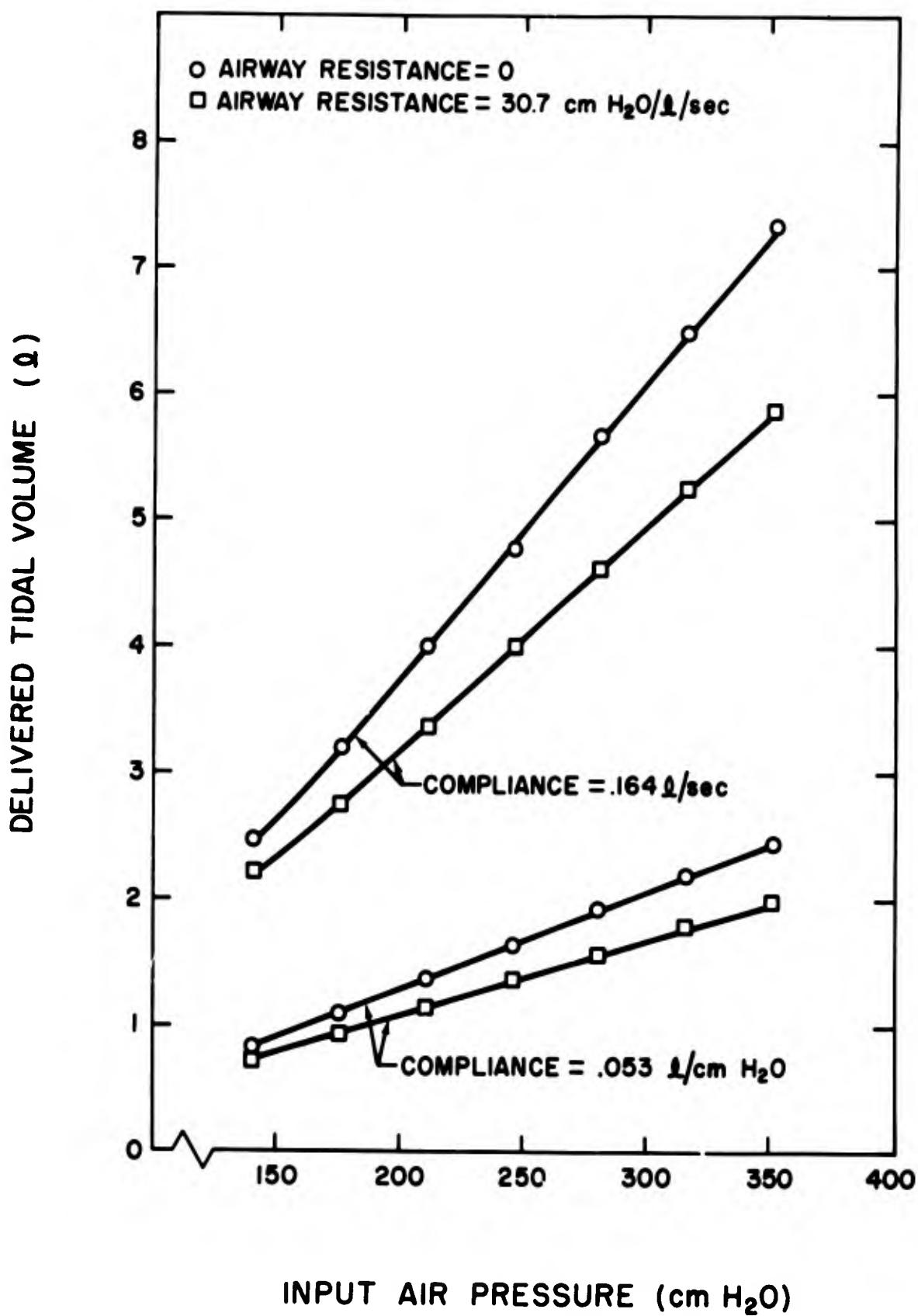Figure 8. Delivered tidal volumes—small compliances.

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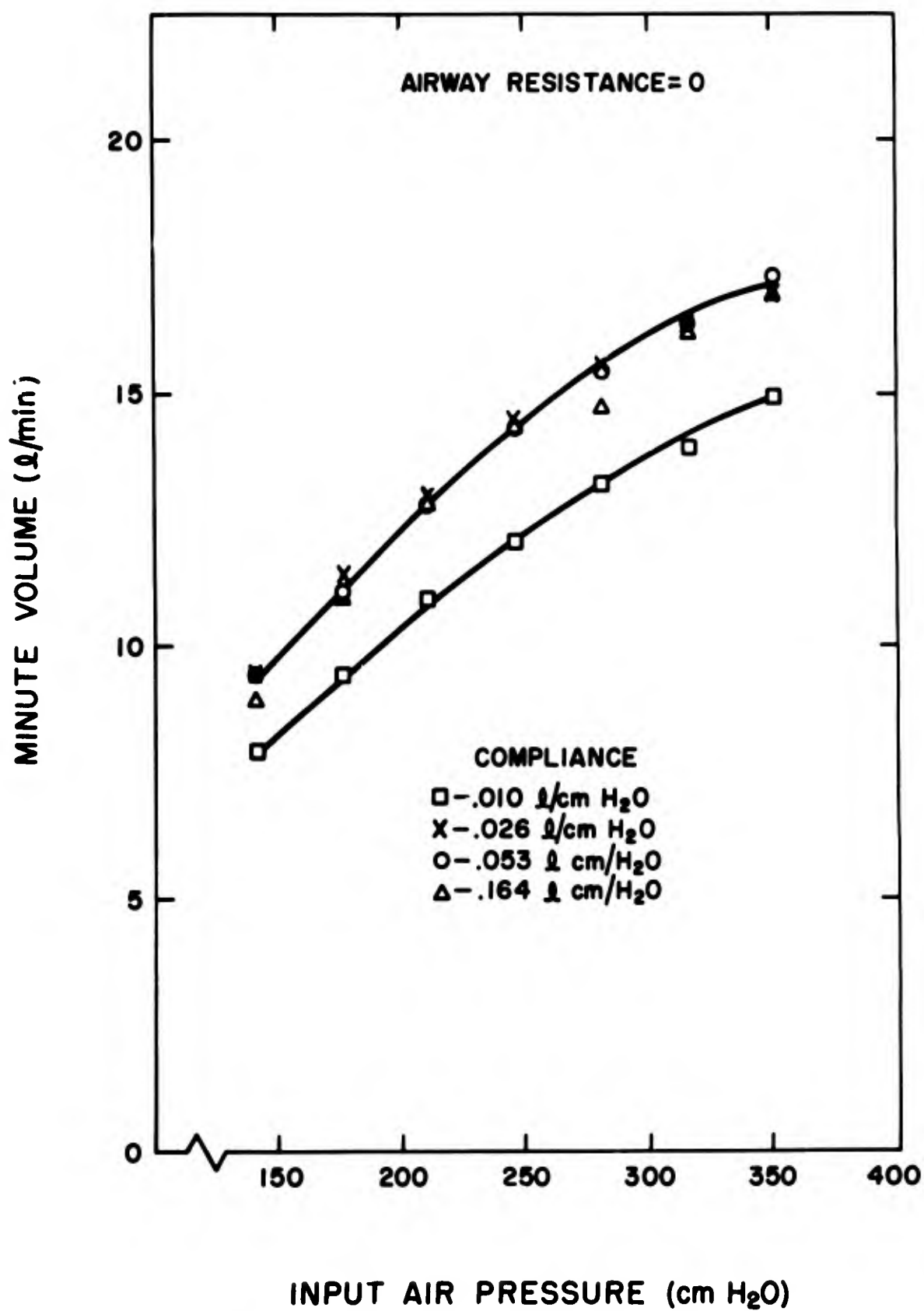Figure 9. Delivered tidal volumes—large tidal volumes.

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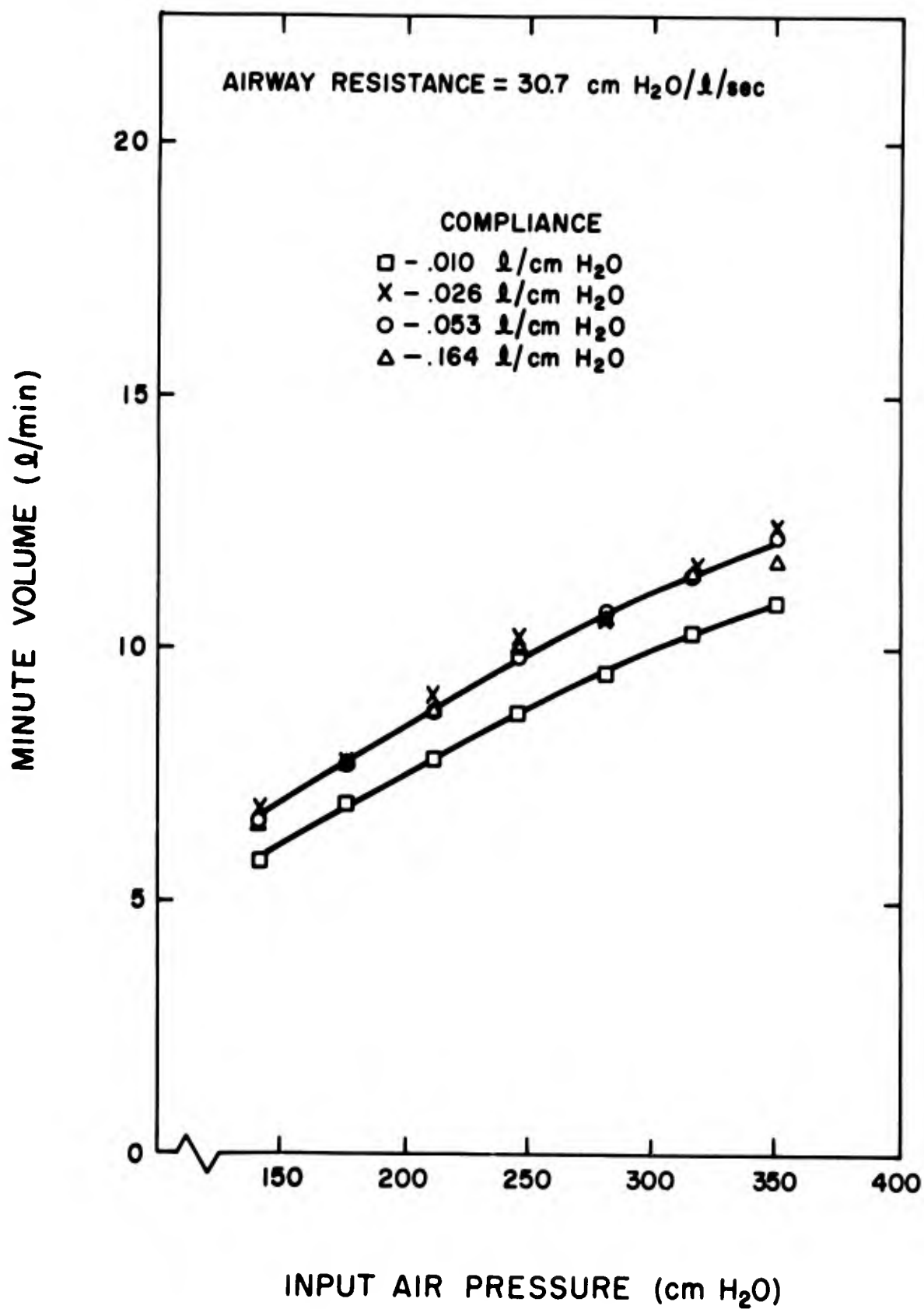Figure 10. Minute volumes—no resistance.

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AIRWAY RESISTANCE = 30.7 cm H_2O/ℓ/sec

COMPLIANCE
□ — .010 ℓ/cm H_2O
X — .026 ℓ/cm H_2O
o — .053 ℓ/cm H_2O
△ — .164 ℓ/cm H_2O
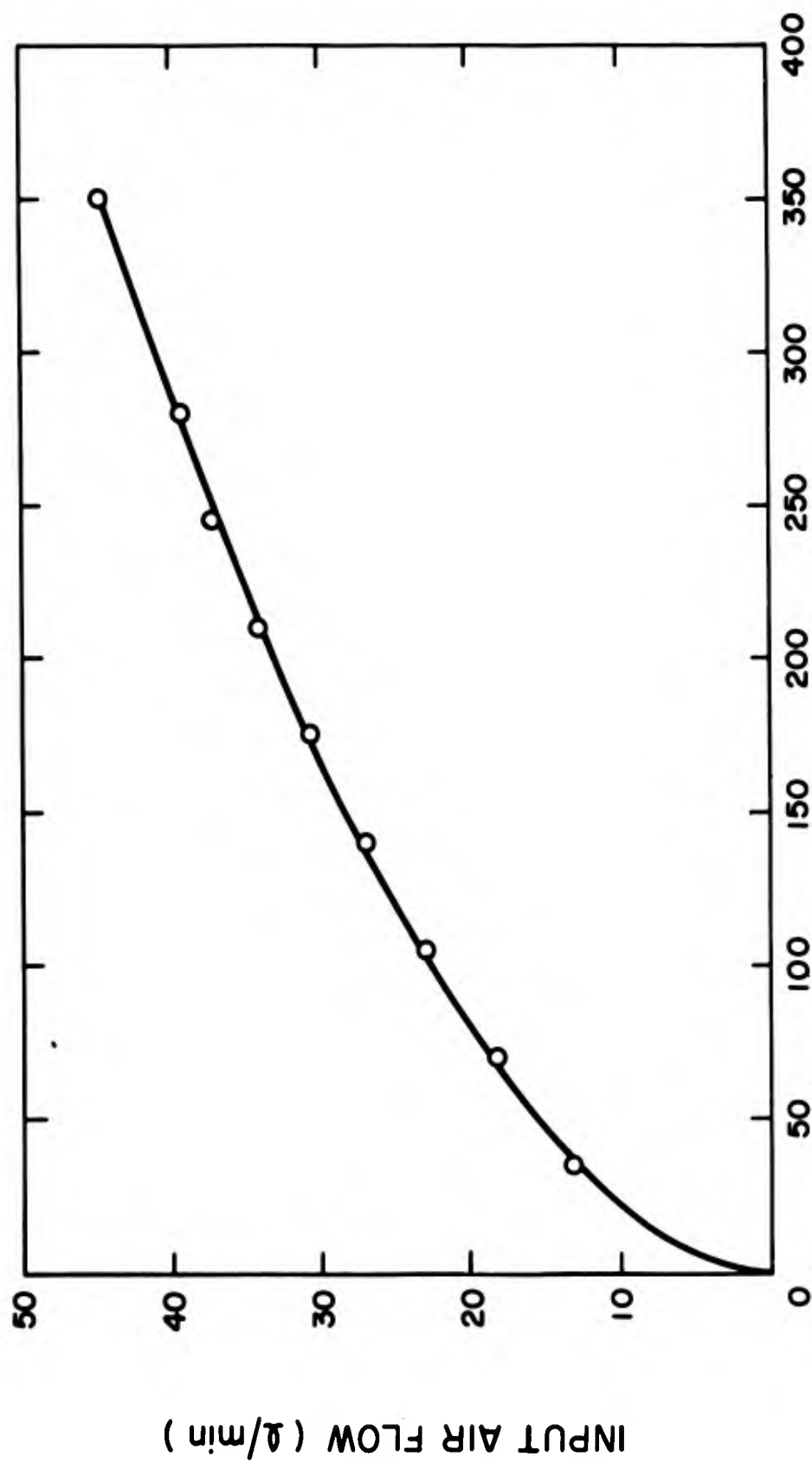
MINUTE VOLUME (ℓ/min)

INPUT AIR PRESSURE (cm H_2O)

Figure 11. Minute volumes—maximum resistance.

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INPUT AIR PRESSURE (cm H₂O)

Figure 12. Air consumption rate.   2047-68

INPUT AIR FLOW (ℓ/min)

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14. KEY WORDS	LINK A		LINK B		LINK C	
	ROLE	WT	ROLE	WT	ROLE	WT
Respirator						
Fluidics						