AD

TM-68-17

REVISED PERFORMANCE EVALUATION OF THE ARMY VOLUME-CYCLED RESPIRATOR, MODEL 2

by

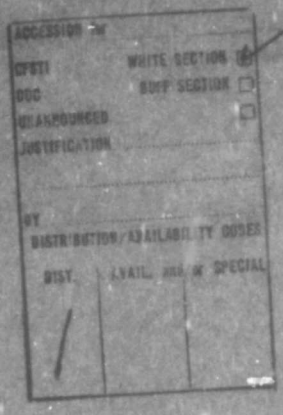
J. W. Joyce, Jr.

July 1968

U.S. ARMY MATERIEL COMMAND

HARRY DIAMOND LABORATORIES

WASHINGTON, D.C. 20438



The findings in this report are not to be construed as an official Department of the Army position.

Destroy this report when it is no longer needed. Do not return it to the originator.

AD

AMCMS Code: 5900.21.11123
HDL Proj: 31000

TM-68-17

REVISED PERFORMANCE EVALUATION OF THE ARMY VOLUME-CYCLED RESPIRATOR, MODEL 2

by

J. W. Joyce, Jr.

July 1968

U.S. ARMY MATERIEL COMMAND
HARRY DIAMOND LABORATORIES
WASHINGTON, D.C. 20438

ABSTRACT

The Army volume-cycled respirator is designed to assist or control the ventilation of patients. A previous analysis of this respirator contained erroneous information that led to quantitatively incorrect performance results. This report describes the errors made and presents both revised and new performance results.

The tests performed indicate that the requirements for cycling rates, pressure-cycling, and assist sensitivity have been satisfied. The requirements for tidal volumes, minute volumes, and inspiratory times have not been fully met. The failure of the respirator to meet tidal volume and minute volume requirements was traced to excessive internal compliance of the respirator. This compliance results in wide variations in delivered tidal volumes with changes in patient load conditions and is considered an undesirable feature for a volume-cycled respirator.

BLANK PAGE

CONTENTS

Illustrations

1. INTRODUCTION

The Army volume-cycled respirator (fig 1) is a joint development of the Harry Diamond Laboratories (HDL) and the Walter Reed Army Institute of Research (WRAIR). It is powered and controlled by two bistable fluid amplifiers. The respirator is designed to either assist or control a patient's ventilation, and it may be pressure-cycled as well as volume-cycled. The power circuit and patient circuit are separated to permit the use of contaminated (not suitable for breathing) compressed gases as the power source.

In an earlier analysis of this respirator (ref 1), some incorrect calculations and measurements led to erroneous test results. The necessary corrections are explained in this report, and revised and additional results are presented.

2. DESIGN REQUIREMENTS

The current functional and packaging requirements for this respirator are based on the premise that the respirator will be used primarily on adult patients. Further, these requirements have been generated with emphasis on the needs of Army medicine.

2.1 Functional Requirements

The functional requirements for this respirator are as follows:

(1) The respirator should be able to assist or control the ventilation of a patient, and the change from one mode of operation to the other should take place automatically.

(2) In the assist mode of operation, the respirator should be able to sense an inspiratory effort by the patient of 1.0 cm H_2O negative pressure. Sensitivity of the inspiration initiation trigger should, in addition, be variable at least to 4 cm H_2O of negative pressure.

(3) The respirator should have pressure-cycling capabilities. The switching pressures for the pressure-cycled mode of operation should be continuously variable between 10 and 60 cm H_2O.

(4) The respirator should have a continuously variable tidal volume capability from 300 to 2500 cc.

(5) The respirator should be able to deliver minute volumes from 5 to 20 ℓ/min.

Ref 1 Joyce, J. W., "Engineering Evaluation of the Army Volume-Cycled Respirator, Model 2," HDL TM-66-20, Harry Diamond Laboratories, Washington, D. C. 20438, December 1966.

(6) When operating in the control mode, the respirator should be able to cycle from 6 to 60 cpm. Requirements 4 through 6 are summarized graphically in figure 2.

(7) The inspiratory period of the respirator should be continuously variable from 0.4 to 1.3 sec.

(8) Expiratory periods achieved by the device should be compatible with the requirements for inspiratory times and cycling rates stated above.

(9) The respirator must be able to administer pure oxygen as well as room air to the patient.

(10) The respirator should have a gas canister to filter contaminated incoming air when room air is used as the breathing gas.

(11) Provisions must be made to humidify the breathing gases being administered to the patient.

2.2 Packaging Requirements

The packaging requirements for this respirator include the following:

(1) The respirator should be constructed so that parts that may become contaminated can be easily cleaned and/or sterilized. In normal use, only the breathing valve and connecting tubing would be sterilized.

(2) The respirator should be capable of operating continuously for three months with a minimum of maintenance.

(3) Interdependence of controls should be minimized or eliminated if possible.

(4) The respirator should be able to use either contaminated or uncontaminated gases as its power source. This necessitates the isolation of the power circuit from the patient circuit.

(5) In general, the respirator should be small, lightweight, portable (carried by one person), and rugged. It should be able to withstand the environmental conditions imposed on it by military applications.

3. OPERATIONAL DESCRIPTION

A schematic diagram of the respirator is shown in figure 3. The two fluid amplifiers are operated through a common pressure regulator. At the beginning of a cycle, the output of amplifier A (located on the front of the respirator) is flowing into the lower piston chamber, pushing the piston upward, while the output of amplifier B (located on the back of the respirator) is exhausting to the surroundings. As the piston is driven upward, the rubber bellows is compressed, and breathing gas in the bellows is forced into the patient's lungs via the breathing valve. In the normal volume-cycled mode of operation, the upward motion of the piston is terminated when a striker attached to the bottom of the bellows activates an adjustable mechanical (upper excursion) trigger. Activation of this trigger opens feedback lines to both fluid amplifiers and switches them simultaneously, so that the output of amplifier B is now flowing into the upper piston chamber, driving the piston down, and the output of amplifier A exhausts to the surroundings. In the pressure-cycled mode of operation, piston upstroke is terminated when a predetermined pressure has been reached in the patient circuit. This pressure activates the pressure-cycle trigger (a spring-loaded diaphragm arrangement), and as in the case of volume-cycled operation, feedback lines to both amplifiers are opened to admit switching air flow, and the amplifiers are switched to begin the downstroke.

As the piston descends, the bellows refills with fresh breathing gas through a one-way valve in the gas intake at the top of the respirator. The patient, meanwhile, exhales to the surroundings through the breathing valve. None of the exhaled gases are allowed to reenter the bellows where they could be breathed again. At the bottom of the downstroke, the bottom of the bellows strikes a fixed mechanical (lower excursion) trigger, and in so doing, opens the left feedback line to amplifier B. This action switches the amplifier's power stream to the right receiver where it exhausts to the surroundings. Amplifier A is not affected by this trigger, and its output continues to exhaust. At the instant the bellows comes to rest at the bottom of the downstroke (but not until such time), the rate of entrainment out of the lower piston chamber produced by the exhausting amplifier A is sufficient to produce negative pressure in this chamber. The negative pressure operates the delay mechanism by exerting a force on one side of a piston. The rate of motion of this piston is controlled by a variable atmospheric bleed (a needle valve) on the opposite side of the piston, which determines the magnitude of the retarding force; as a consequence, a variable time delay is achieved. In the control mode of operation, at the end of its travel, the delay piston strikes a mechanical trigger which opens the right feedback line to amplifier A, thus switching its power

stream back into the piston chamber to begin a new cycle. In the assist mode of operation, an inspiratory attempt by the patient, in the form of a small negative pressure, initiates a new cycle by activating the inspiration initiation trigger before the delay mechanism trigger is activated. The inspiration initiation trigger is also a spring-loaded diaphragm arrangement which, when activated, uncovers the right feedback line to amplifier A to switch it and begin a new cycle.

The functions of the various controls in altering the parameters involved are discussed later.

4. TEST EQUIPMENT AND PROCEDURE

To determine the performance capabilities of the respirator, it was necessary to simulate lung compliance and air resistance. Lung compliance was simulated by fixed-volume tanks. A perforated disk was inserted in the tubing connecting the respirator to the tank (compliance) to simulate airway resistance.

The compliances of the tanks used with the respirator were experimentally determined by injecting known volumes of air into them during a fixed time interval and measuring the peak pressures developed. In the previous analysis, compliances were calculated from equation (1) which is based on the perfect gas law for an isothermal process.

$$C = \frac{V}{P_\infty} \qquad (1)$$

where C = compliance of tank, ℓ/cm H_2O

V = volume of tank, ℓ

P_∞ = ambient pressure, cm H_2O

The average measured values of compliance and the calculated compliances previously used are shown in table I. These values indicate that the measured compliances are less than the previously calculated values and are a function of the injection time. The compliance increases as the injection time increases. The limiting value is the one corresponding to an isothermal process. The reason that the original calculation was incorrect is that the time factor was not considered. The time constant required for the isothermal condition to exist in the two larger tanks is about 1 min. The times involved in the respirator data (inspiratory times) are only a few seconds at most. Actually, the measured compliances in the larger tanks correspond closely to those calculated for an isentropic process. The small tank has a much smaller time constant, as indicated by the fact that the measured compliance is almost the same as that calculated from the isothermal relationship.

The airway resistance was calibrated by measuring pressure drops across the perforated disk for various flow rates. The pressure drop, ΔP, divided by the corresponding flow rate, Q, was plotted versus flow rate and the result was a straight line (fig. 4), which implies that the pressure drop can be generally described by equation (2)

$$\Delta P = k_1 Q + k_2 Q^2 \qquad (2)$$

where k_1 = y-intercept of the straight line

k_2 = slope of the straight line

For the curve in figure 4, $k_1 = 3.8$ and $k_2 = 26.9$. Equation (2) consists of a laminar flow pressure drop ($k_1 Q$) and a turbulent flow pressure drop ($k_2 Q^2$). If airway resistance is defined as the pressure drop per unit of flow produced by a flow rate of 1 ℓ/sec, equation (2) becomes

$$\Delta P = k_1 + k_2 \qquad (3)$$

Using the above definition, the resistance of the perforated disk is 30.7 cm H_2O/ℓ/sec.

A pressure transducer (±1 psid range) was used to sense pressures in the tanks and ahead of the airway resistance. The latter measurement represents face mask pressure -- the pressure monitored by most respirators. A second transducer (50 psia range) was used to monitor pressures in the right feedback line of amplifier B to determine the downstroke time and the delay time. These two durations cannot be determined from tank pressures, which indicate only the total expiratory time (the sum of downstroke time and delay time). The outputs of the transducers were displayed on an oscilloscope, and the data were recorded from photographs of the oscilloscope traces.

5. PERFORMANCE RESULTS

The respirator was tested under conditions of extreme compliances and airway resistance. The four combinations used were as follows:

(1) Low compliance, no resistance

(2) Low compliance, high resistance

(3) High compliance, no resistance

(4) High compliance, high resistance

5.1 Calibration of Tidal Volumes

Before testing the respirator against the extremes of load conditions, it was first necessary to calibrate the tidal volume settings for the respirator. This calibration was done in the following manner. The respirator was connected to the 76-liter tank with no airway resistance. The volumes delivered to the tank were then calculated from the peak pressures developed using the applicable compliance value from table I. In this manner, strokes corresponding to tidal volumes at 100-cc intervals were marked on the bellows housing of the respirator. The maximum tidal volume available was about 1700 cc.

5.2 Inspiratory Times

Inspiratory times for the respirator can be varied by adjusting the input pressure to the fluid amplifiers. Inspiratory time decreases as input pressure increases. To determine operating limits the respirator was tested under each of the four loading conditions described above for two different input pressures at each of the various tidal volume settings. The two input pressures were 30 psig, the upper design limit of input pressure, and the lowest pressure necessary to cycle the unit against the particular load. Figures 5 and 6 indicate that for the low compliance, the respirator cannot be operated at the upper limit of inspiratory time. This effect is due mostly to the high pressures developed in the 0.010-ℓ/cm H_2O compliance, as a result of which the minimum input pressure to cycle the respirator is raised to such a level that the slower inspiratory times cannot be achieved. Figures 7 and 8 show that for the high compliance, both limits of inspiratory time are achieved over the range of tidal volumes. At small tidal volumes, however, the upper limit is not achievable, while at large tidal volumes, the lower limit cannot be reached. These are limitations of the respirator inasmuch as even at the lowest possible input pressures, the strokes for small tidal volumes are so short that they will always be completed in less than 1.3 sec. Similarly, for large tidal volumes, the stroke cannot be completed in 0.4 sec, even for the maximum input pressure of 30 psig. It remains to be seen whether this limitation is, in practice, a serious one, or whether the real problem is a lack of adequate definition of the requirements for inspiratory times. It seems reasonable that the range of inspiratory times would be a function of tidal volume -- the faster times being associated with the smaller tidal volumes, and the slower times with larger tidal volumes. In other words, constant inspiratory times regardless of tidal volumes appear to be unrealistic.

5.3 Cycling Rates

The cycling rates produced by the respirator are shown in figures 9 through 11. As before, the respirator was tested under the extreme loading conditions of lung compliance and airway resistance. The lower limit of cycling rates in all cases was obtained for the lowest input pressure to cycle the respirator (maximum inspiratory time) and maximum delay time (maximum expiratory time). The upper limit in each case was obtained at the upper limit of input pressure (30 psig), corresponding to the minimum inspiratory time. For the small compliance (0.010 ℓ/cm H_2O), the delay control was adjusted for minimum delay time, so that expiratory time was also minimized. The resulting cycling rates for the small compliance (fig. 9) provide a wide enough range of operation to satisfy the imposed requirements. These results also show that the airway resistance has very little effect on the cycling rates for this case.

For the large compliance, the delay control was adjusted so that the total expiratory time was just equal to the time required for the tank pressure to return to atmospheric. The results shown in figure 10 indicate that the range of cycling rates with no airway resistance is not as great for the high compliance as for the small compliance but is still adequate to meet the requirements. For the large compliance with maximum airway resistance (fig. 11), the range of cycling rates is quite narrow. This limited performance is not, however, the fault of the respirator. It is caused by the long expiratory time required for the tank to exhaust completely and return to atmospheric pressure, thereby greatly lowering the upper limit of cycling rates attainable. Shorter expiratory times would result in air "trapping" within the tank; that is, a residual volume of air (and a corresponding residual positive pressure) would remain in the lungs, causing an increase in mean pulmonary pressure and a decrease in effective ventilation capacity. Since this limitation for large compliance is not a respirator deficiency, the cycling requirements may still be considered as having been met.

5.4 Pressures Developed

The pressures developed in the tank and ahead of the airway resistance for the extreme conditions of operation are shown in figures 12 through 16. For the low compliance and no airway resistance (fig. 12), the pressures in the tank vary slightly with input pressure. This effect is probably caused by losses through the breathing valve during the inspiratory phase. At the lower input pressures, the valve does not completely seal as rapidly, and some of the breathing gases are lost to the surroundings. The differences produced are not great enough to

cause concern. The results for the low compliance with maximum airway resistance (fig. 13) show that the airway pressure (ahead of the resistance) is higher at the 30 psig input pressure than at the lower input pressures. This effect is to be expected because the higher input pressure is associated with faster inspiratory times, and hence, greater inspiratory flow rates that result in increased pressure drops across the resistance. The tank pressures in this case are relatively independent of input pressure. For the case of low compliance with or without airway resistance, the peak tank pressures are less than those that would be developed if all the preset tidal volume were delivered to the tank. The loss of tidal volume and its cause are discussed later.

For the large compliance, the results for the case of no airway resistance (fig. 14) show no significant variation of tank pressure with changes in input pressure. For maximum resistance, the pressures ahead of the resistance (fig. 15) become quite large at the upper input pressure. The pressure drops across the resistance for 30-psig input pressure are up to three times the corresponding tank pressure developed (fig. 16). Tank pressures actually developed in the high compliance tank, compared with those that would occur if the preset volume were delivered, were slightly high with no resistance present, and generally a little low for high airway resistance.

5.5 Tidal Volumes Delivered

Using the data for the tank pressures developed, the tidal volumes actually delivered to the tanks can be calculated and compared with the calibrated or preset values. In this manner, the effects of patient load on respirator performance can be ascertained. It is desirable for a volume-cycled respirator to deliver a constant tidal volume regardless of changes in patient loads. Figures 17 and 18 show the effects of the extremes of compliance and airway resistance on tidal volume, based on the calibrated volume settings. These results show that appreciable losses in tidal volume occur when the respirator is operated against the low compliance or the high resistance, or both. There is also a slight increase in tidal volume (about 10 percent on the average) when operating the respirator against the high compliance with no airway resistance.

The reason for these variations in delivered tidal volumes produced by different patient loads is that the rubber bellows can expand radially as patient circuit pressures increase. Clearly, the amount of expansion will be a function of the pressure inside the bellows. Thus, for low compliance or high airway resistance, the pressure developed in the bellows is high, and the expansion is large compared with the expansion produced by the lower pressures associated with the compliance used to calibrate tidal volumes. The larger expansion at

high loads means that less volume is displaced from the bellows, and the delivered tidal volume is thereby reduced. For the large compliance, bellows pressures are somewhat lower than those associated with calibration, and the resulting lesser degree of expansion allows the delivered volume to slightly exceed the corresponding calibrated volume. This effect can be illustrated by equation (4).

$$V_{T_2} = \frac{C_2 (C_1 + C_R)}{C_1 (C_2 + C_R)} V_{T_1} \tag{4}$$

where

V_{T_2} = volume delivered to patient, cc

V_{T_1} = calibrated tidal volume, cc

C_1 = compliance used to calibrate volumes, ℓ/cm H_2O

C_2 = patient compliance, ℓ/cm H_2O

C_R = compliance of respirator, ℓ/cm H_2O

Equation (4) shows that delivered volume is a function of respirator compliance, which in this case is due primarily to the expansion of the bellows (as opposed to compression effects caused by dead space volume). As respirator compliance increases, the discrepancy between V_{T_2} and V_{T_1} likewise increases. The only way to eliminate any difference between V_{T_1} and V_{T_2} is to have a respirator with no internal compliance ($C_R = 0$). Although such a respirator is not likely to be found, a good volume-cycled respirator should at least have minimal internal compliance.

5.6 Minute Volumes

The actual minute volumes delivered to the tanks are shown in figures 19 through 21. The minute volumes were determined by first calculating the tidal volumes delivered (from tank pressure measurements) and then multiplying these volumes by the corresponding respirator cycling rates. The minute volumes for the low compliance are shown in figure 19, which indicates that the upper limit cannot always be reached, especially for the case of maximum airway resistance. The reason for this is twofold. First, the actual tidal volumes are considerably less than calibrated values (fig. 17). Second, the high pressures developed in the patient circuit (tank and bellows) create a severe back load on the driving piston of the respirator, and thereby restrict the upper limit of cycling rates that can be achieved. This effect is most noticeable for the case of high airway resistance.

Figure 20 shows that the minute volume requirements are satisfied for all operating conditions for the case of high compliance and no airway resistance. For high compliance and maximum airway resistance, figure 21 shows that once again the upper limit of minute volumes cannot be reached. This result is to be expected because of the limitations imposed on the cycling rates for this loading condition (sect 5.3). Therefore, the failure of the respirator to meet the upper limit of minute volumes under these conditions, as in the case of its failure to meet the cycling rate requirements, is not a shortcoming of the respirator, but is instead a limitation imposed by the load itself as explained in section 5.3.

5.7 Pressure Cycling

The switching pressure for pressure-cycled operation is controlled by the spring force acting on the diaphragm of the pressure-cycle trigger as described earlier. Figure 22 shows the effects of input pressure to the respirator and compliance on the switching pressure for a given spring setting. Even for the extremes of loading and operating conditions involved, the range of switching pressure variation is only 5 cm H_2O. The same range of variation occurs at higher switching pressures (higher spring forces) also. Therefore, once the spring force has been set, the switching pressure will not change by more than 5 cm H_2O with a change in lung compliance or input pressure to the respirator. The spring was selected so that switching pressures from 10 to 60 cm H_2O could be achieved. This range of pressures is in accord with the requirements for pressure-cycled operation.

5.8 Assist Sensitivity

The sensitivity of the inspiration-initiation trigger to inspiratory efforts by the patient is also controlled by a spring force adjustment. Increasing the spring force by rotating the control knob in a clockwise direction decreases the sensitivity; i.e., the patient must exert a stronger inspiratory effort to initiate a new cycle. Because of the small pressures involved, this trigger is somewhat dependent upon input pressures to the respirator. Therefore, the maximum sensitivity is obtained as follows: Once the input pressure has been selected, the sensitivity control knob is turned counterclockwise until the respirator begins to free run; i.e., when it starts to cycle without an effort from the patient (and before the delay time has elapsed). The knob is then turned clockwise until the free run is just eliminated. Under these conditions, the sensitivity is about 0.5 cm H_2O of negative pressure. With the control knob turned clockwise until it bottoms, the sensitivity is about 4 cm H_2O of negative pressure. These sensitivities are within the limits set forth in the requirements.

5.9 Air Consumption

The input air flows to the respirator as a function of input air pressure are shown in figure 23. This curve sums the input flows to both fluid amplifiers. The peak consumption rate is 61 ℓ/min of free air at an input pressure of 30 psig. In normal use, it is anticipated that the respirator would be operated at input pressures between 10 and 20 psig; the corresponding flows under these conditions are 33 to 47 ℓ/min of free air.

The maximum pressure developed in the piston chamber (the pressure recovery of either fluid amplifier) is 30 to 35 percent of the input pressure to the respirator.

6. DISCUSSION AND CONCLUSION

The performance results indicate that the respirator satisfies some, but unfortunately not all, of the functional requirements imposed on it. The primary cause for deficiencies in performance is the internal compliance of the respirator, such compliance resulting mostly from expansion of the bellows. This compliance produces large discrepancies between the calibrated and delivered tidal volumes under severe patient load conditions. This in turn limits the minute volume delivered to the patient. In addition, the maximum available tidal volume (1700 cc) is less than the 2500 cc set forth in the requirements. The only performance deficiency not attributed to the internal compliance problem is that concerning inspiratory times, and this deficiency is probably not too serious.

Five prototypes of this respirator have been completed. One is located at HDL and another at WRAIR. The remaining three units were distributed for further medical evaluation. The findings of one of these evaluators (Dr. Meyer Saklad and his associates at the Rhode Island Hospital, Providence, R. I.) confirms that tidal volumes are very much affected by patient loading; that is, tidal volume decreases sharply as patient load is increased to a maximum value.

The internal compliance problem strongly suggests that steps should be taken to minimize the variations in tidal volume produced by changes in patient loads. Potential solutions to the problem include constraining the bellows so as to minimize radial expansion at the end of the inspiratory phase or eliminating the bellows completely. In any event, some major design changes appear to be in order.

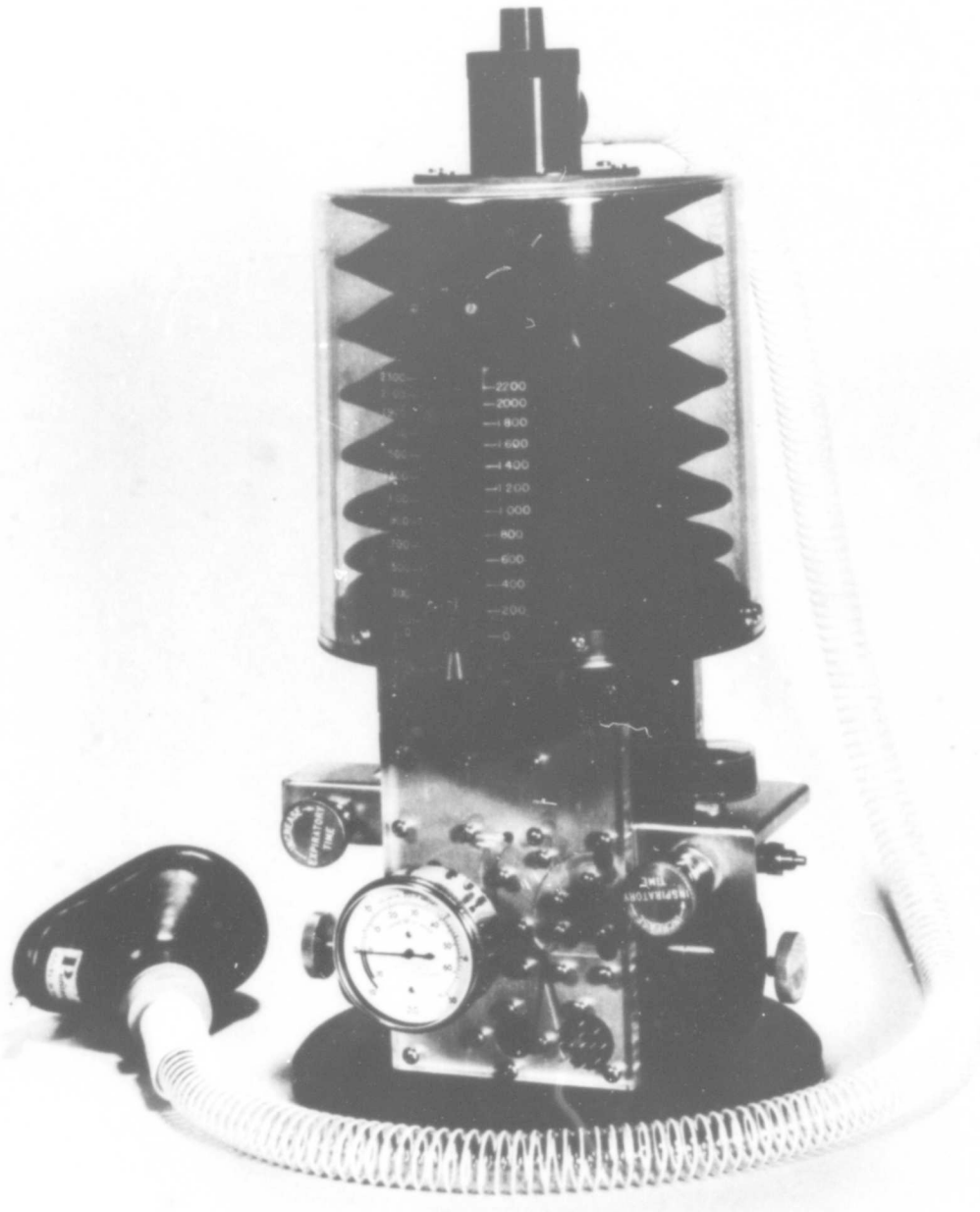


Figure 1. Army volume-cycled respirator.

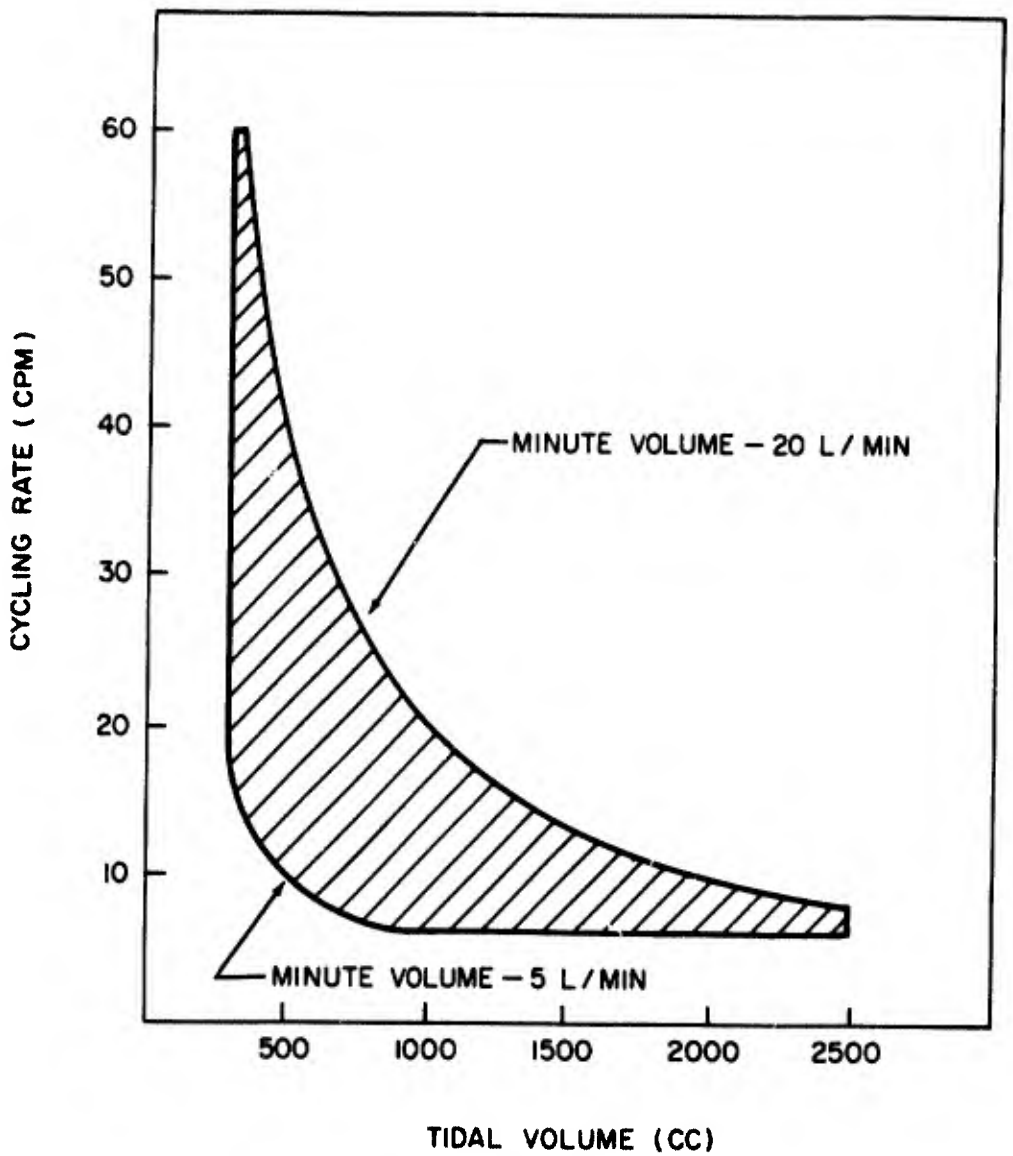


Figure 2. Requirements for cycling rates as a function of tidal volume.

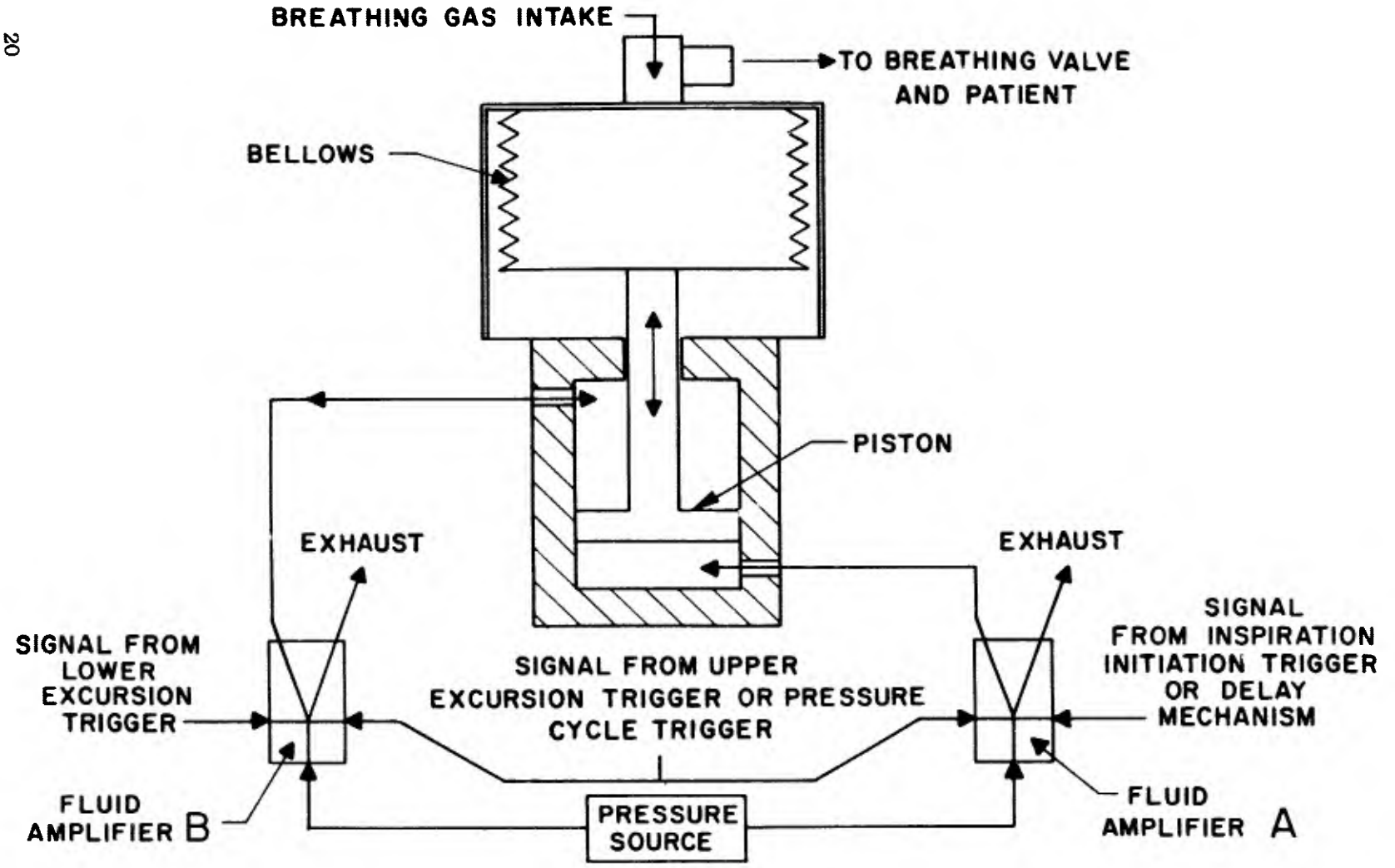


Figure 3. Volume-cycled respirator schematic.

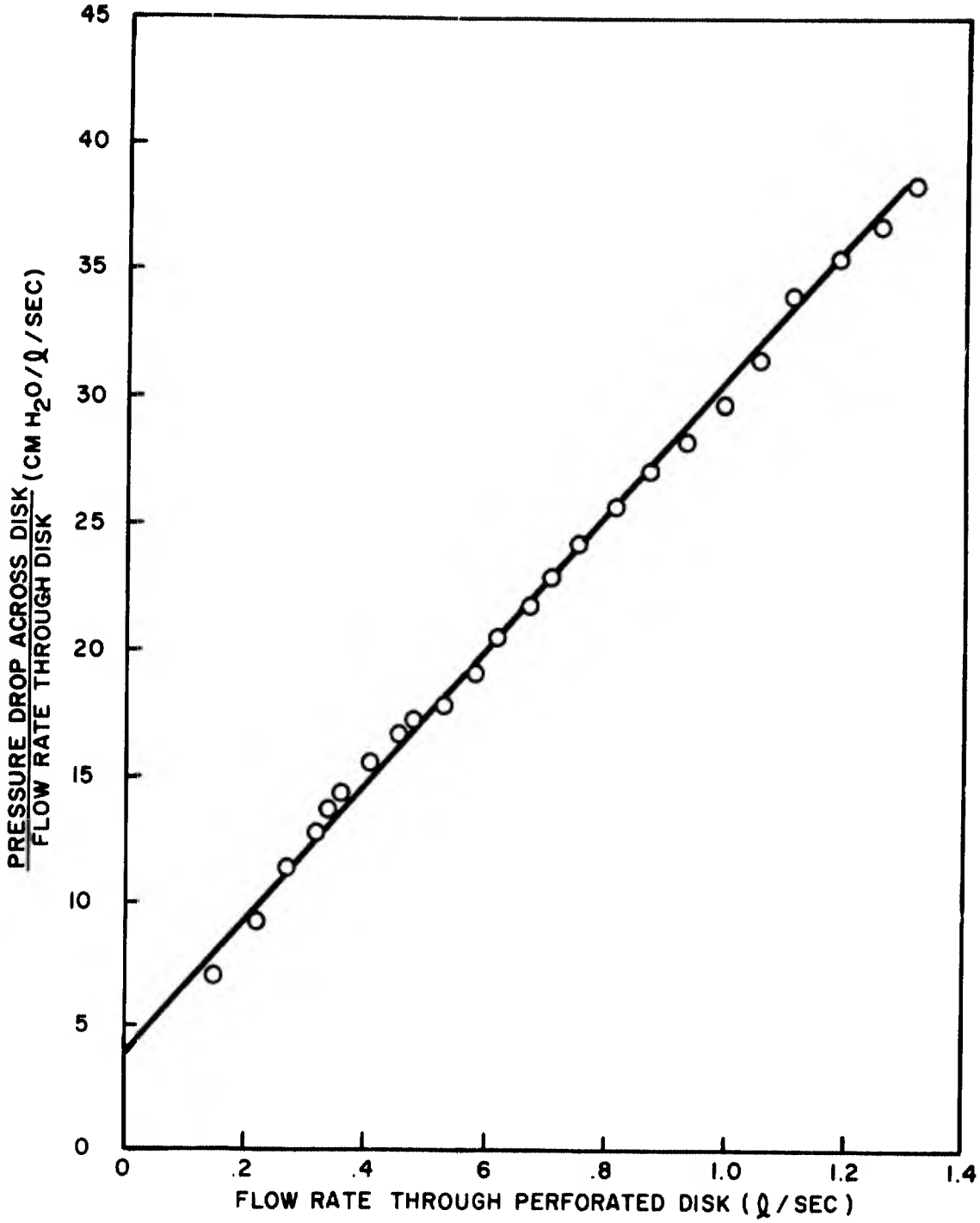


Figure 4. Airway resistance calibration curve.

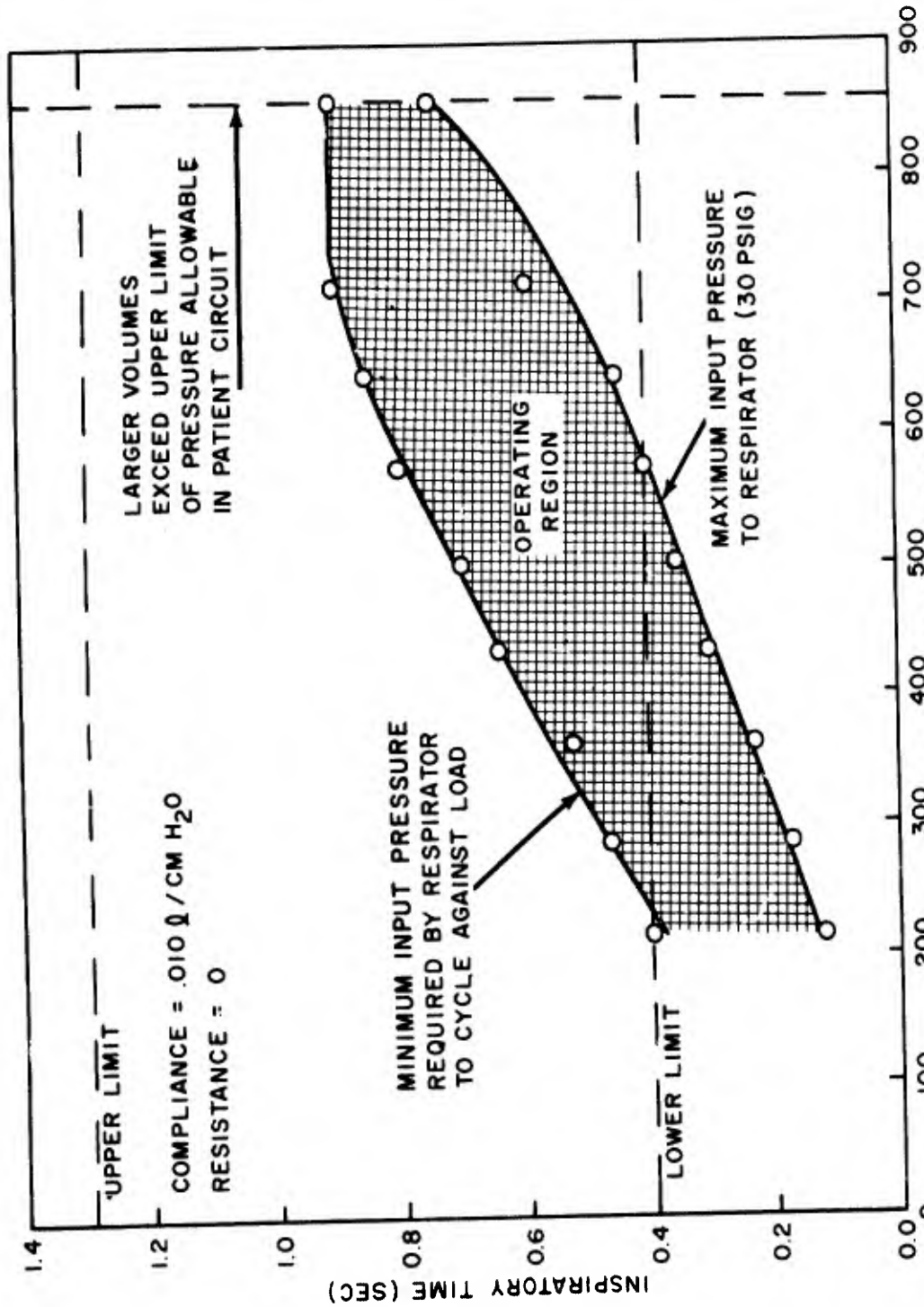


Figure 5. Inspiratory times - low compliance, no resistance.

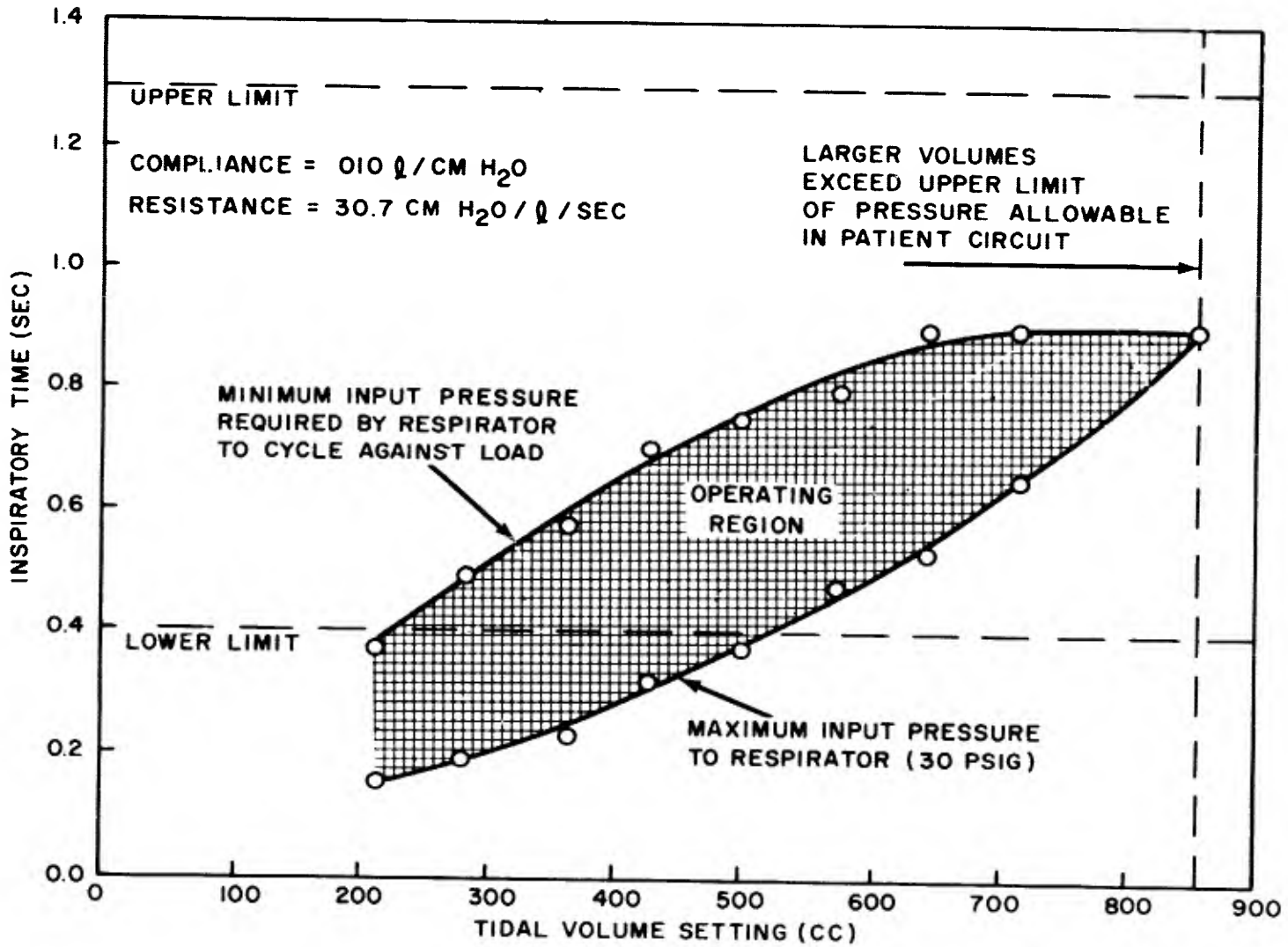


Figure 6. Inspiratory times - low compliance, high resistance.

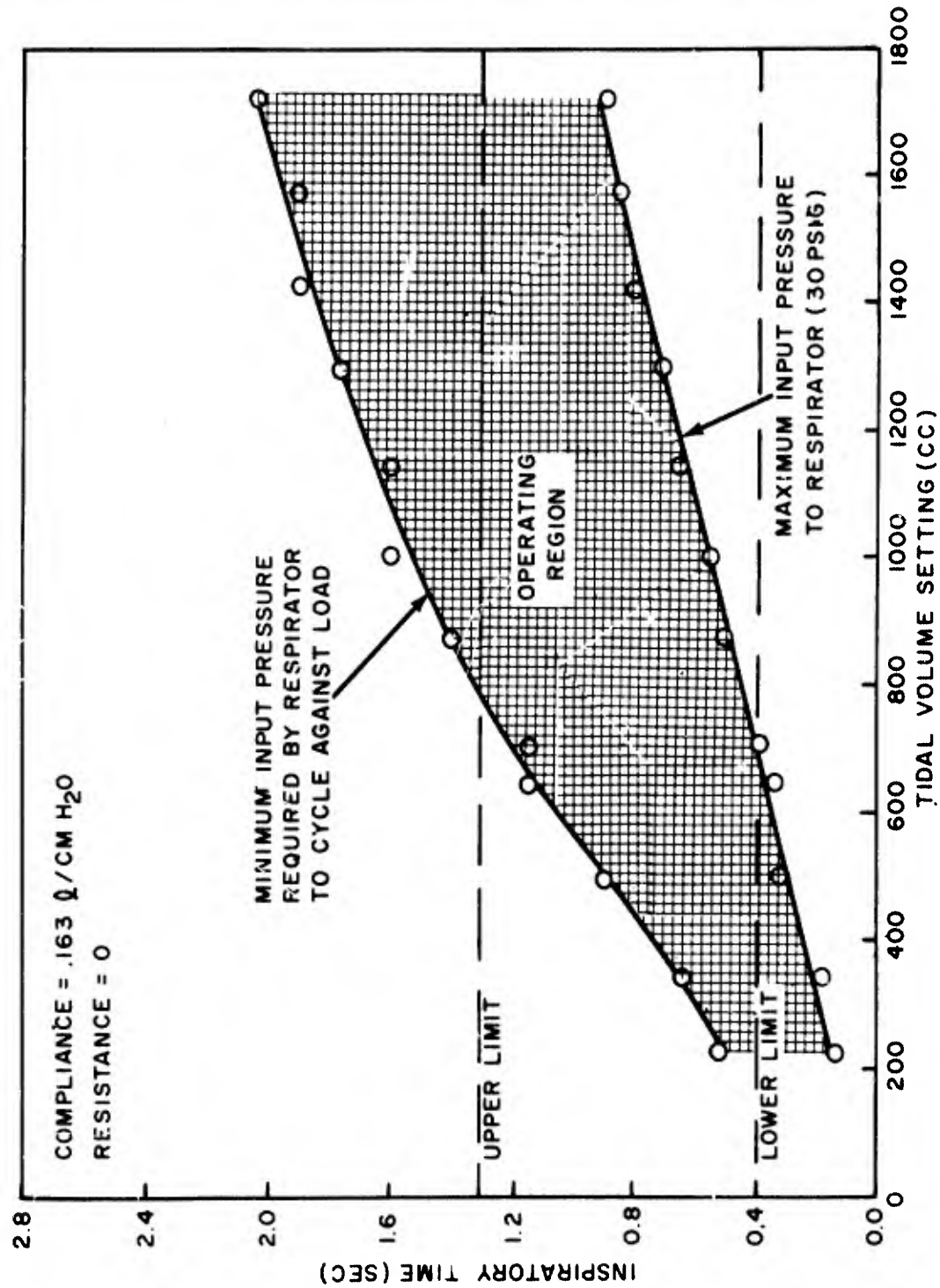


Figure 7. Inspiratory times - high compliance, no resistance.

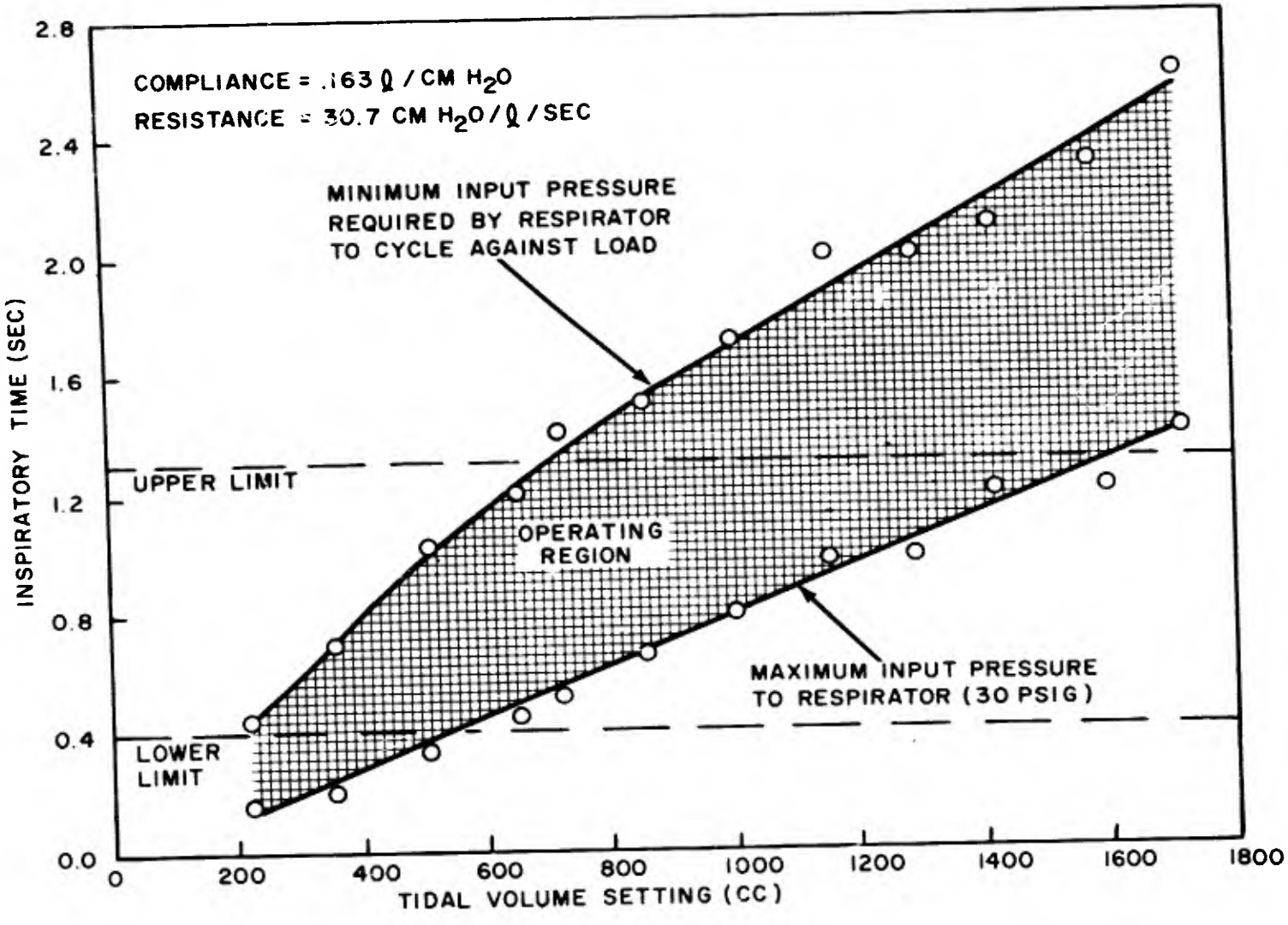


Figure 8. Inspiratory times - high compliance, high resistance.

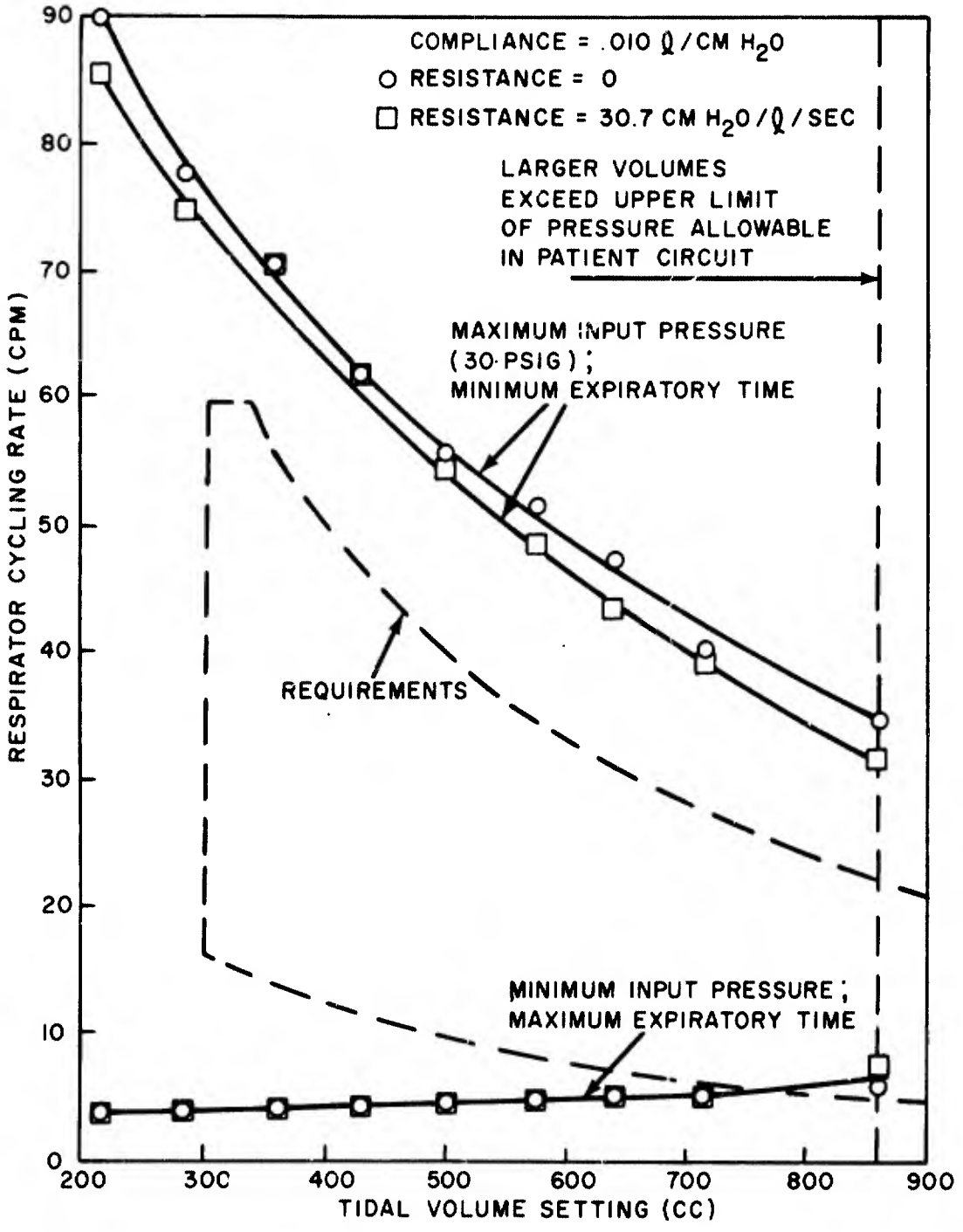


Figure 9. Cycling rates - low compliance.

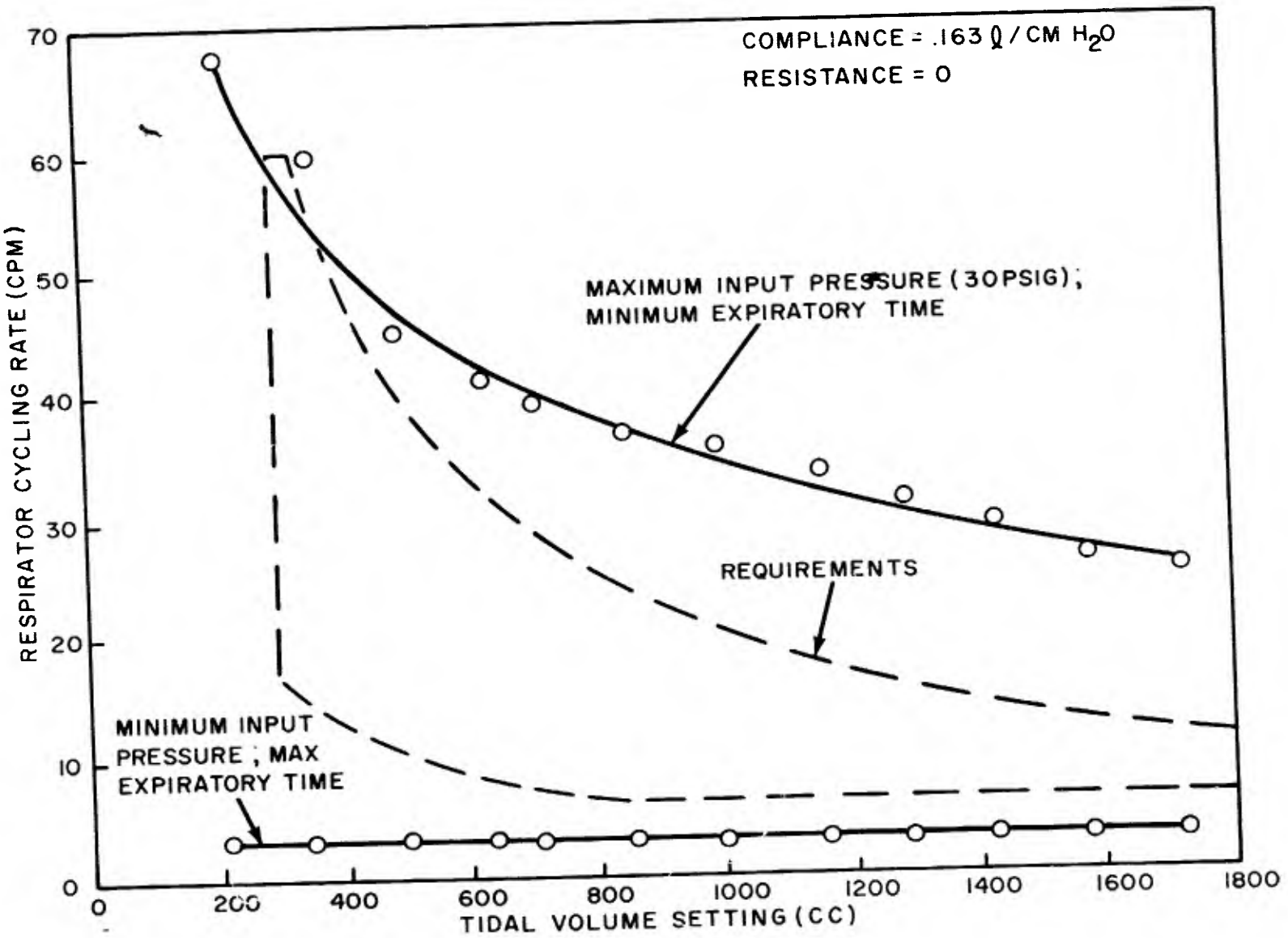


Figure 10. Cycling rates - high compliance, no resistance.

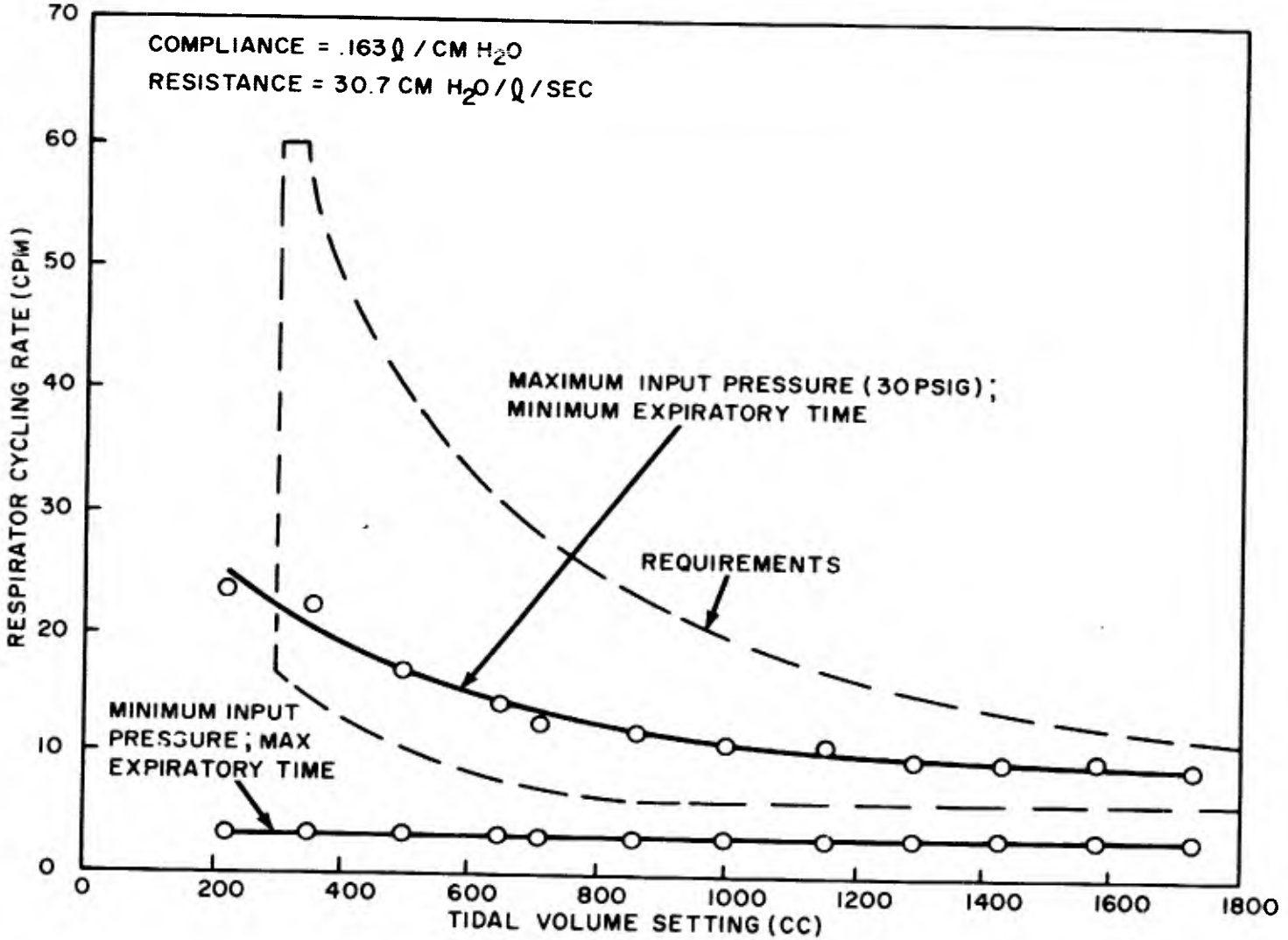


Figure 11. Cycling rates - high compliance, high resistance.

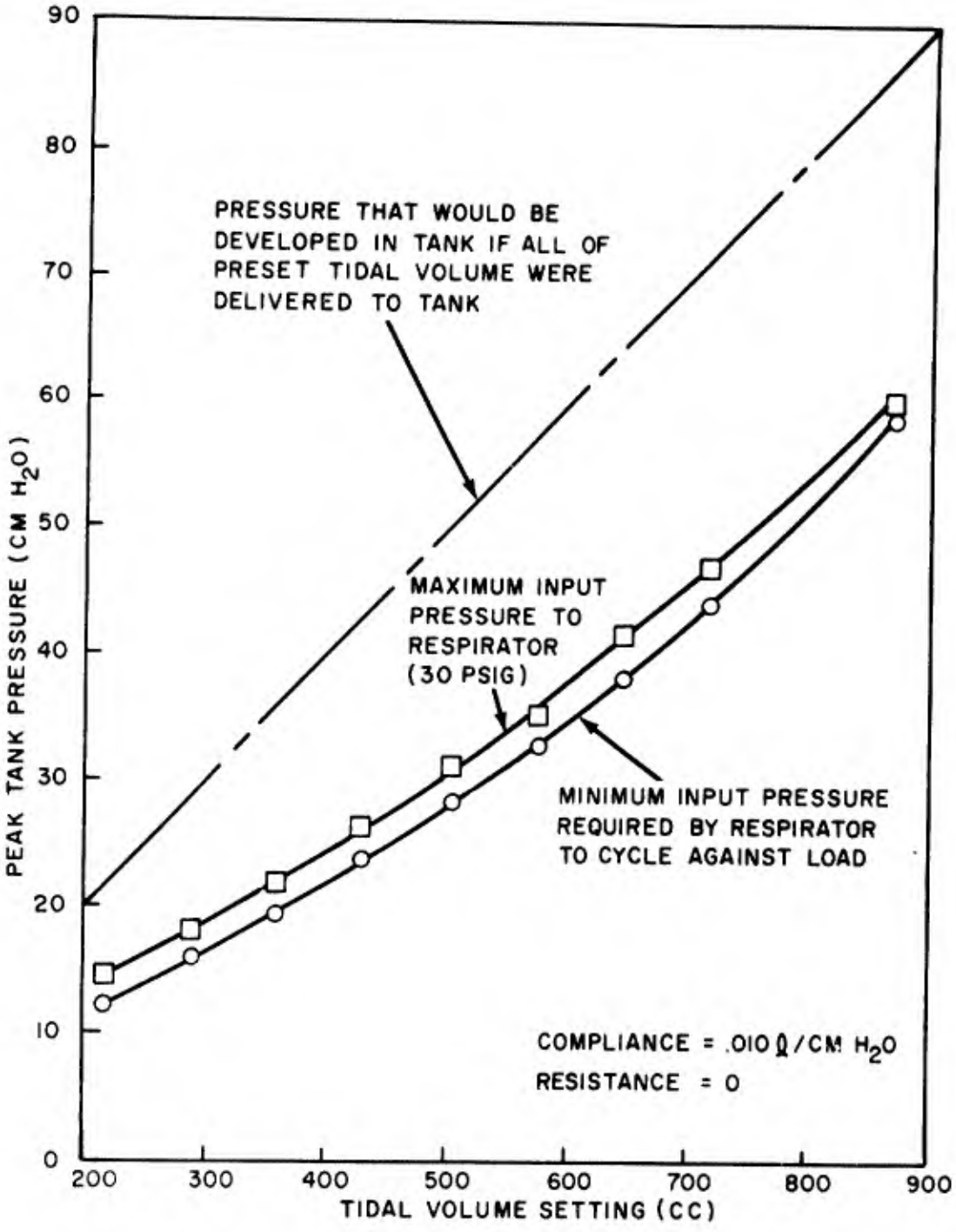


Figure 12. Pressures developed - low compliance, no resistance.

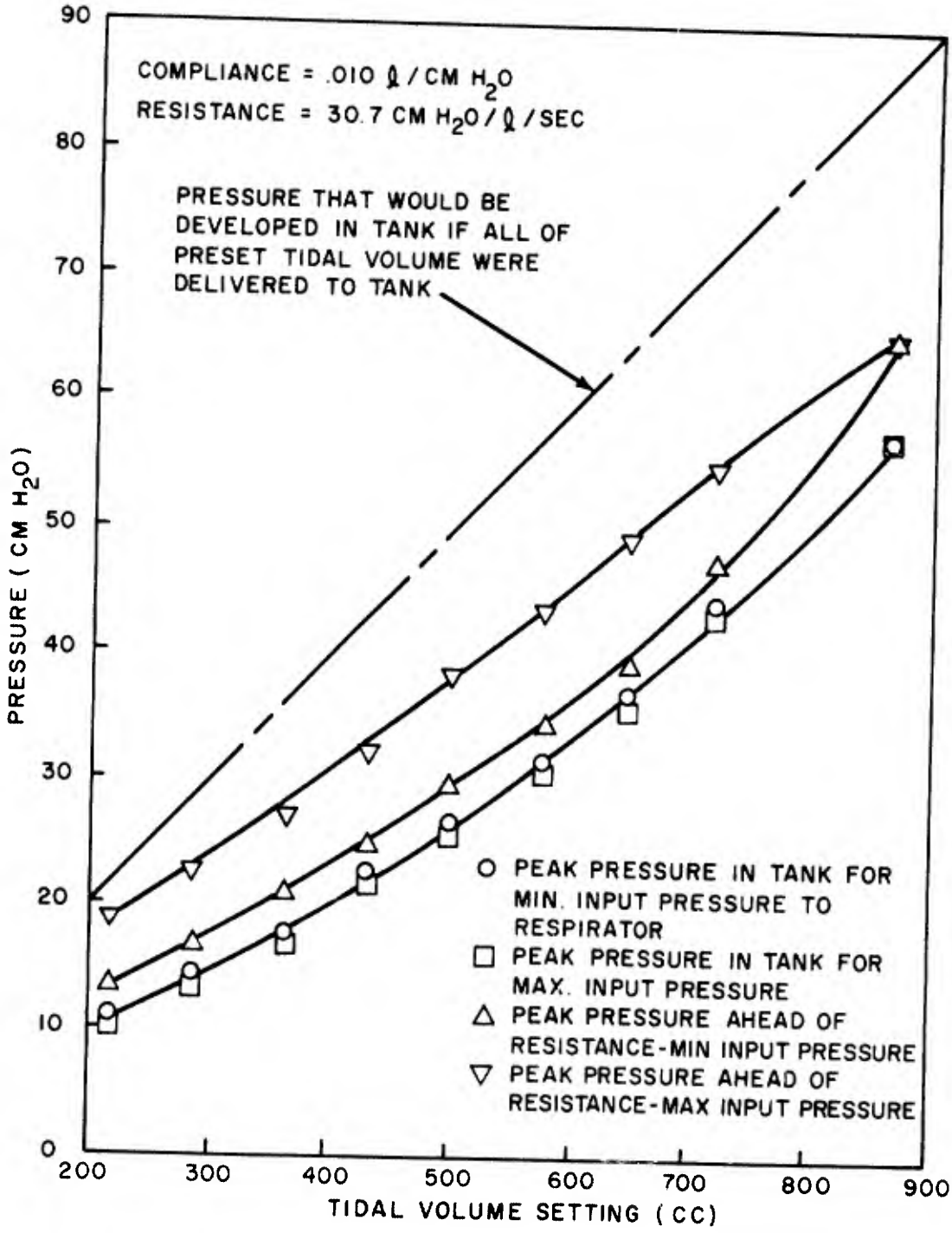


Figure 13. Pressures developed - low compliance, high resistance.

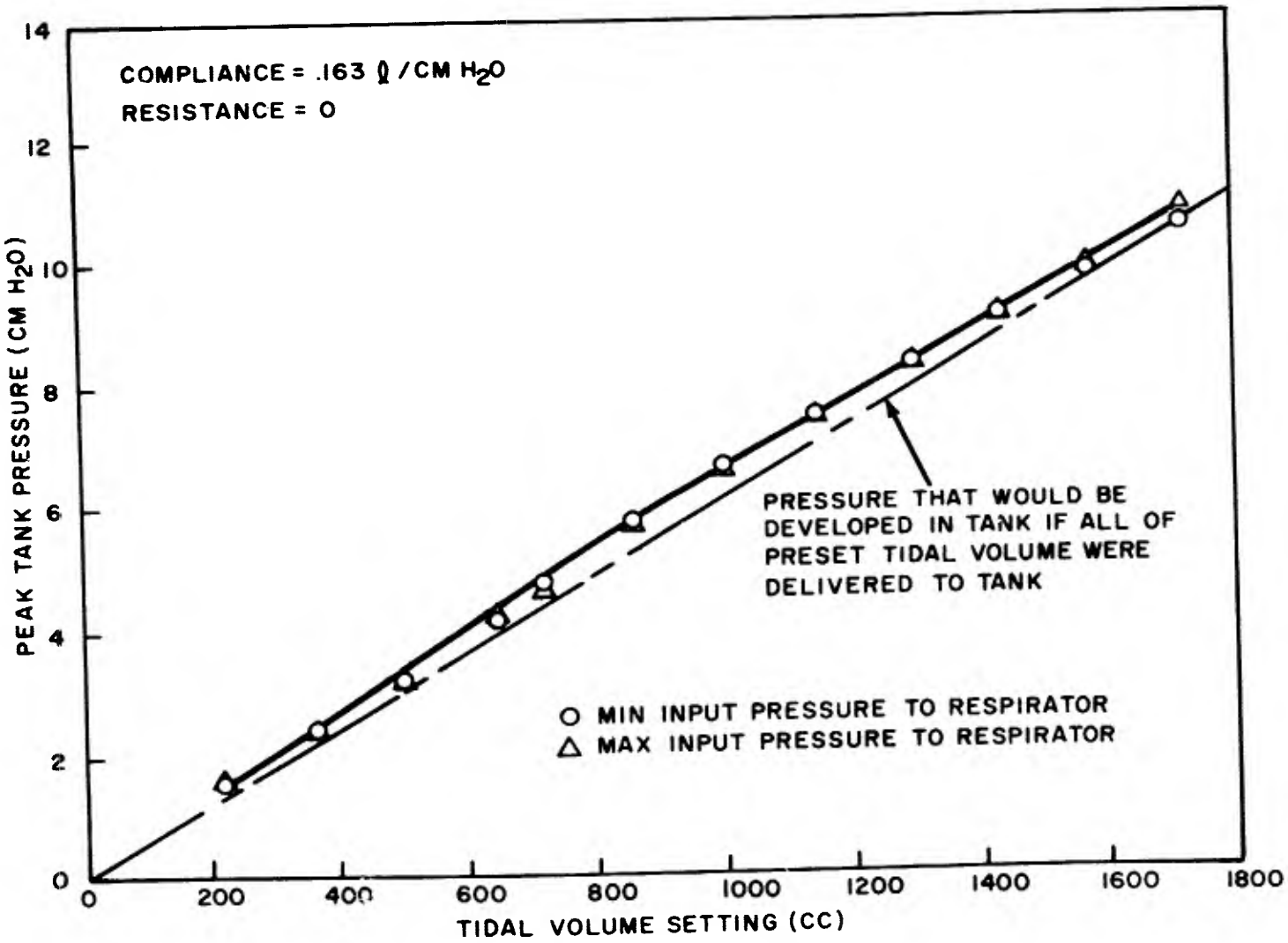


Figure 14. Pressures developed - high compliance, no resistance.

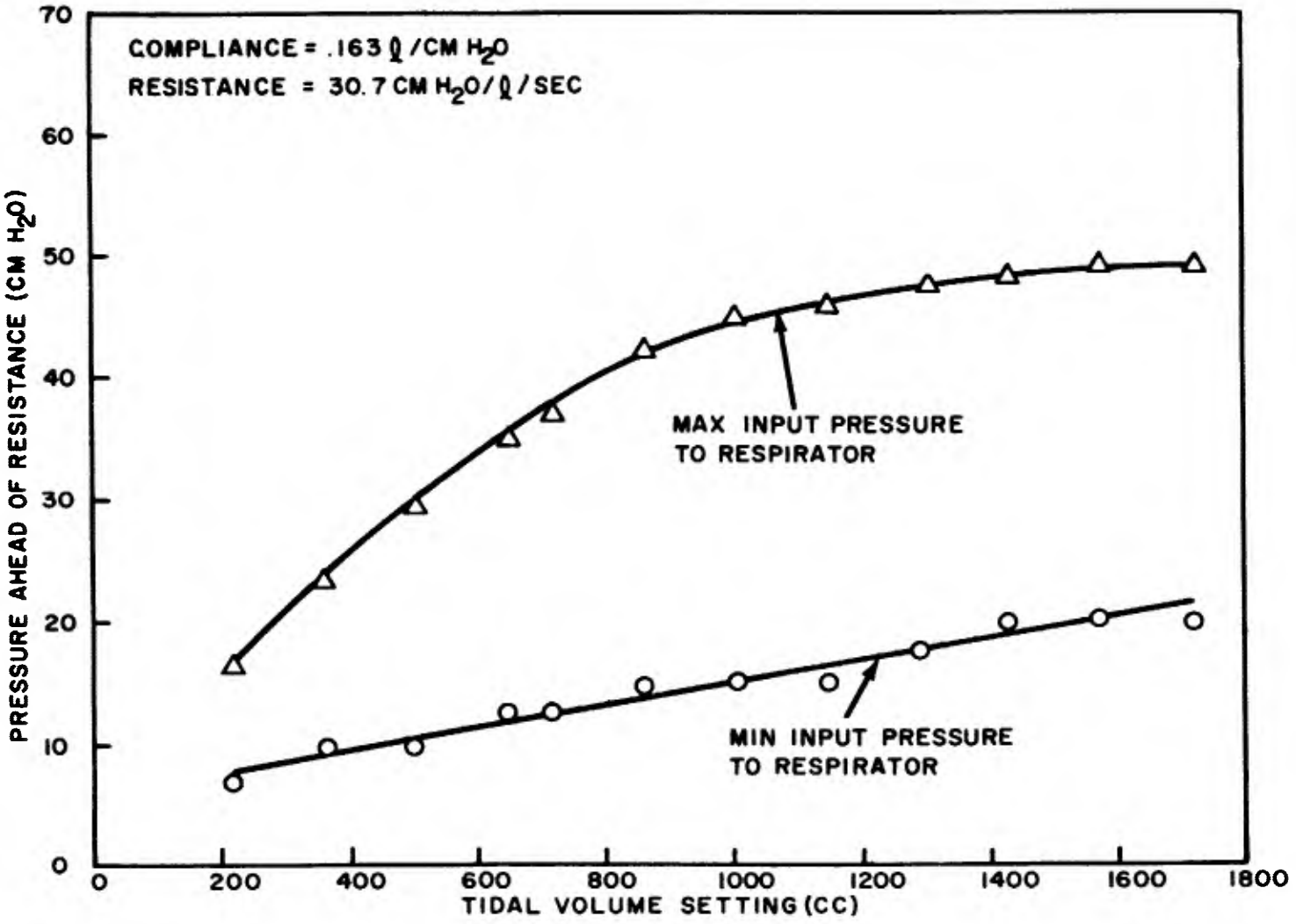


Figure 15. Pressures ahead of resistance - high compliance.

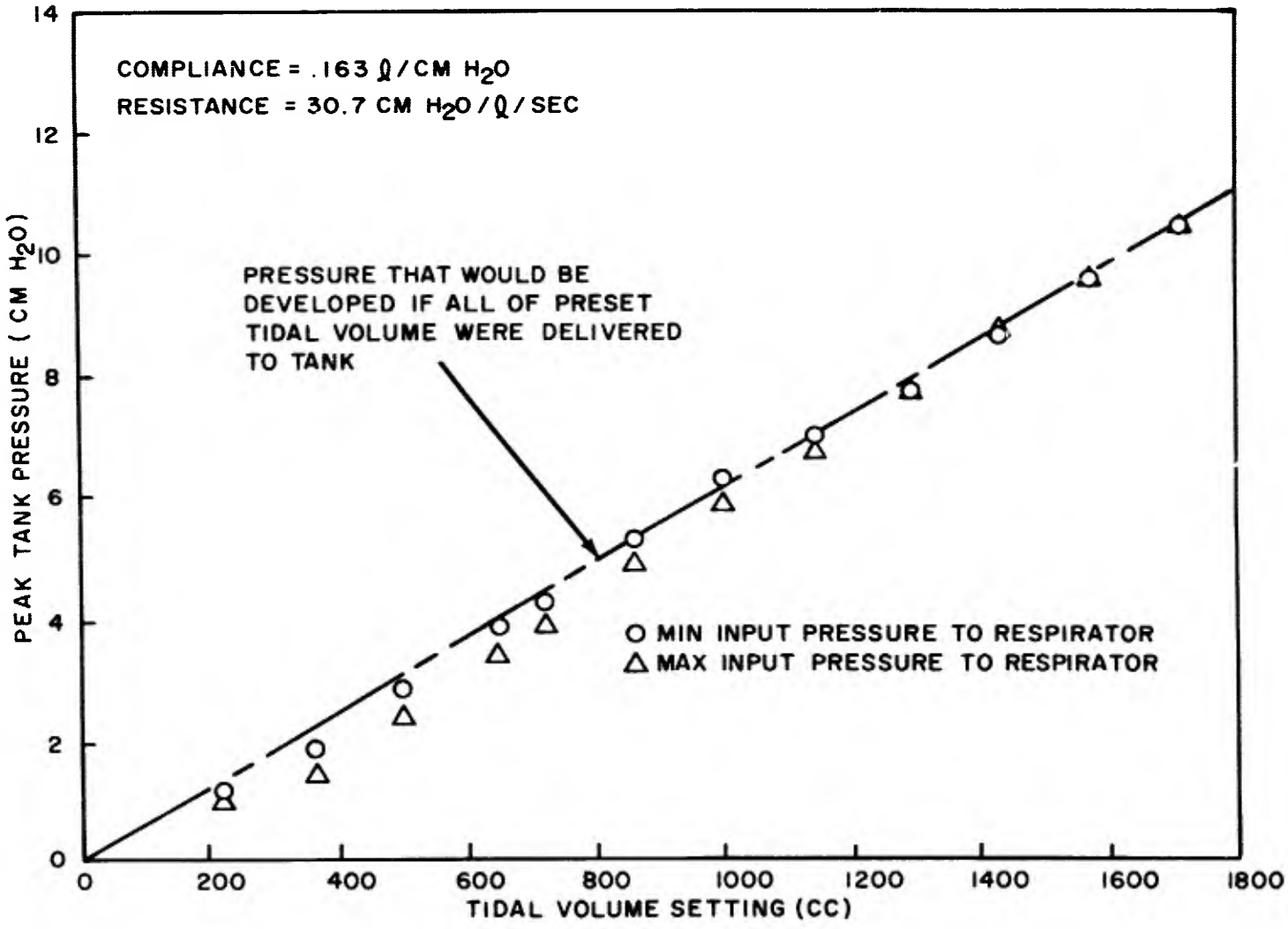


Figure 16. Pressures in tank - high compliance, high resistance.

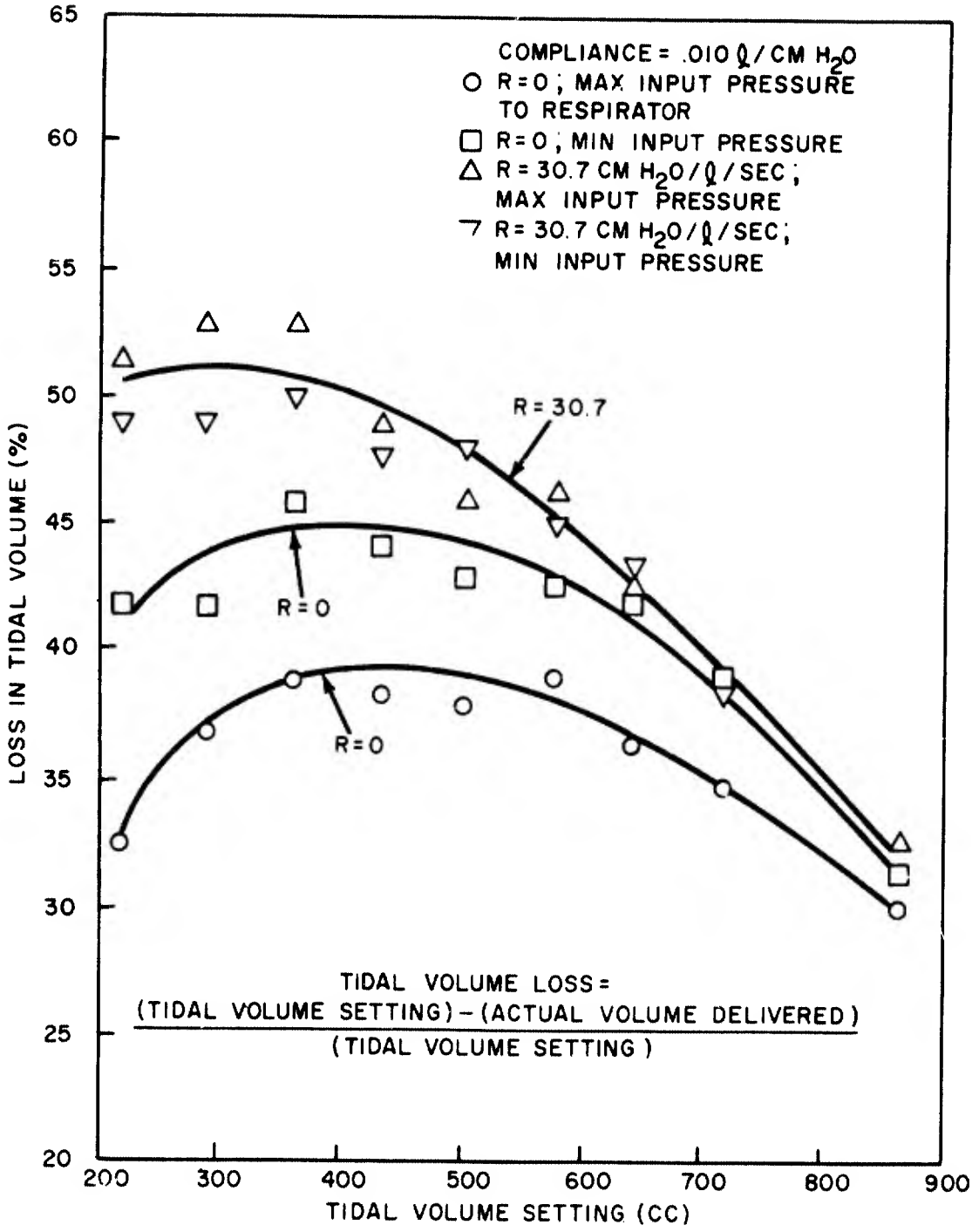


Figure 17. Loss in tidal volume - low compliance.

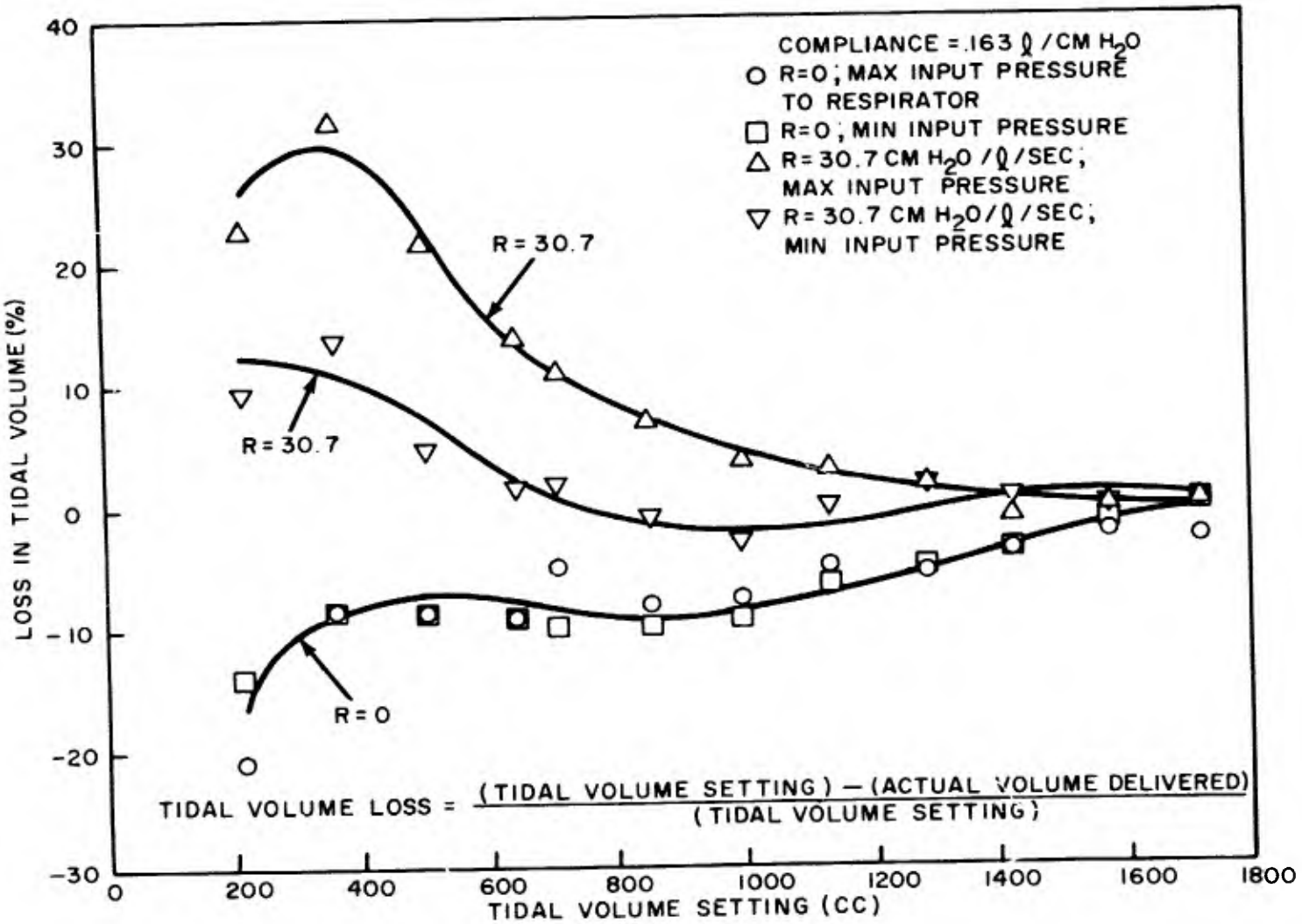


Figure 18. Loss in tidal volume - high compliance.

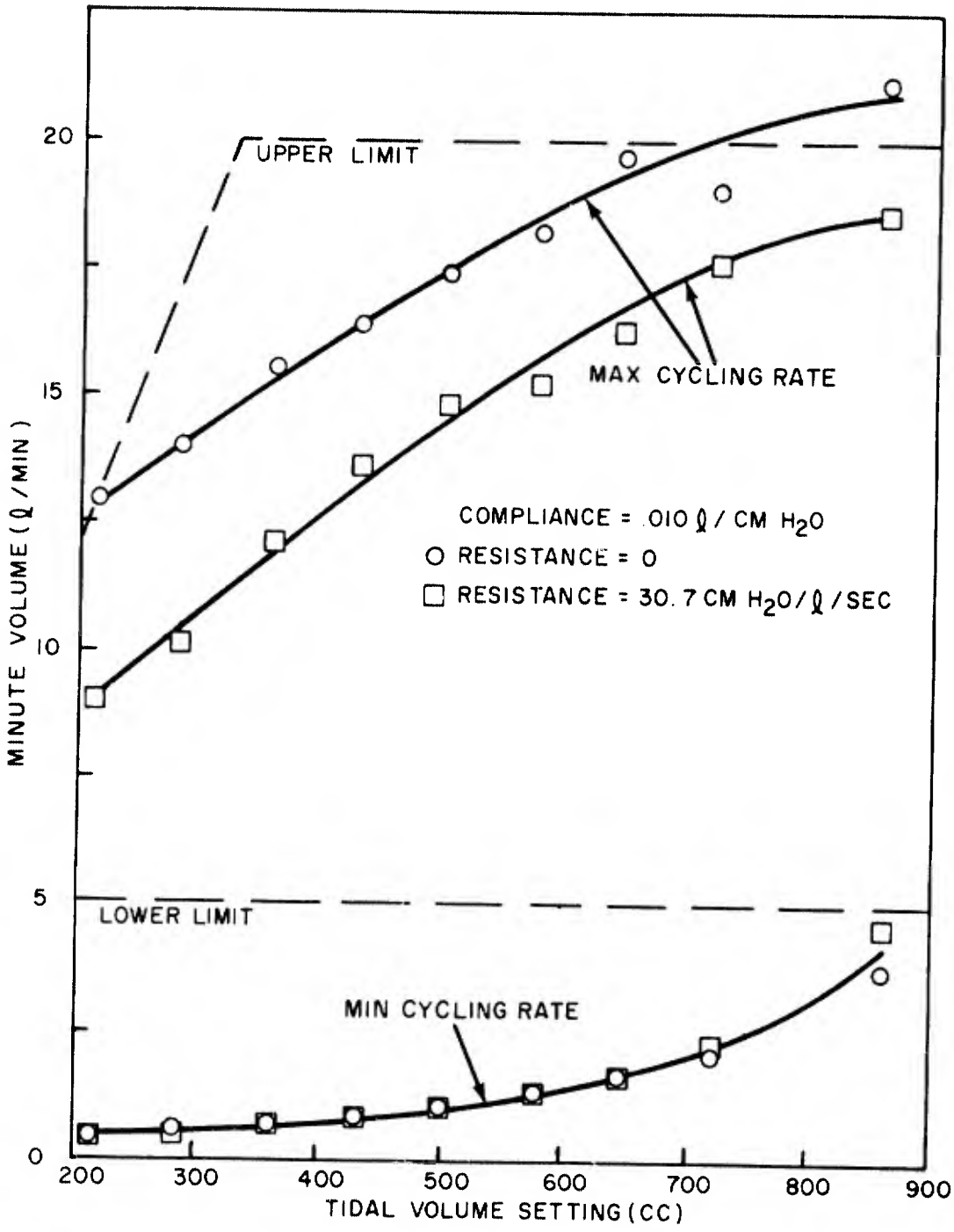


Figure 19. Minute volumes - low compliance.

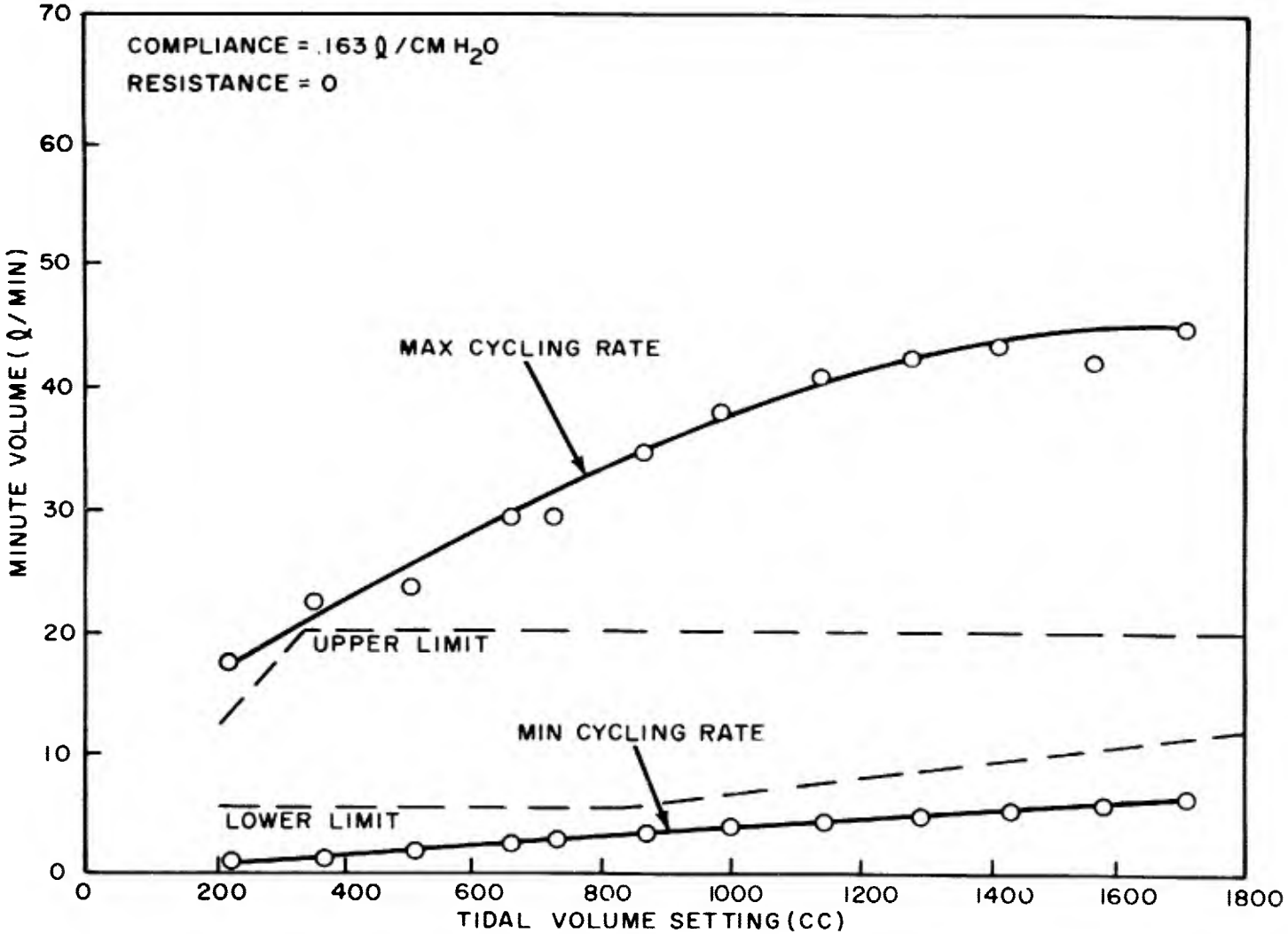


Figure 20. Minute volumes - high compliance, no resistance.

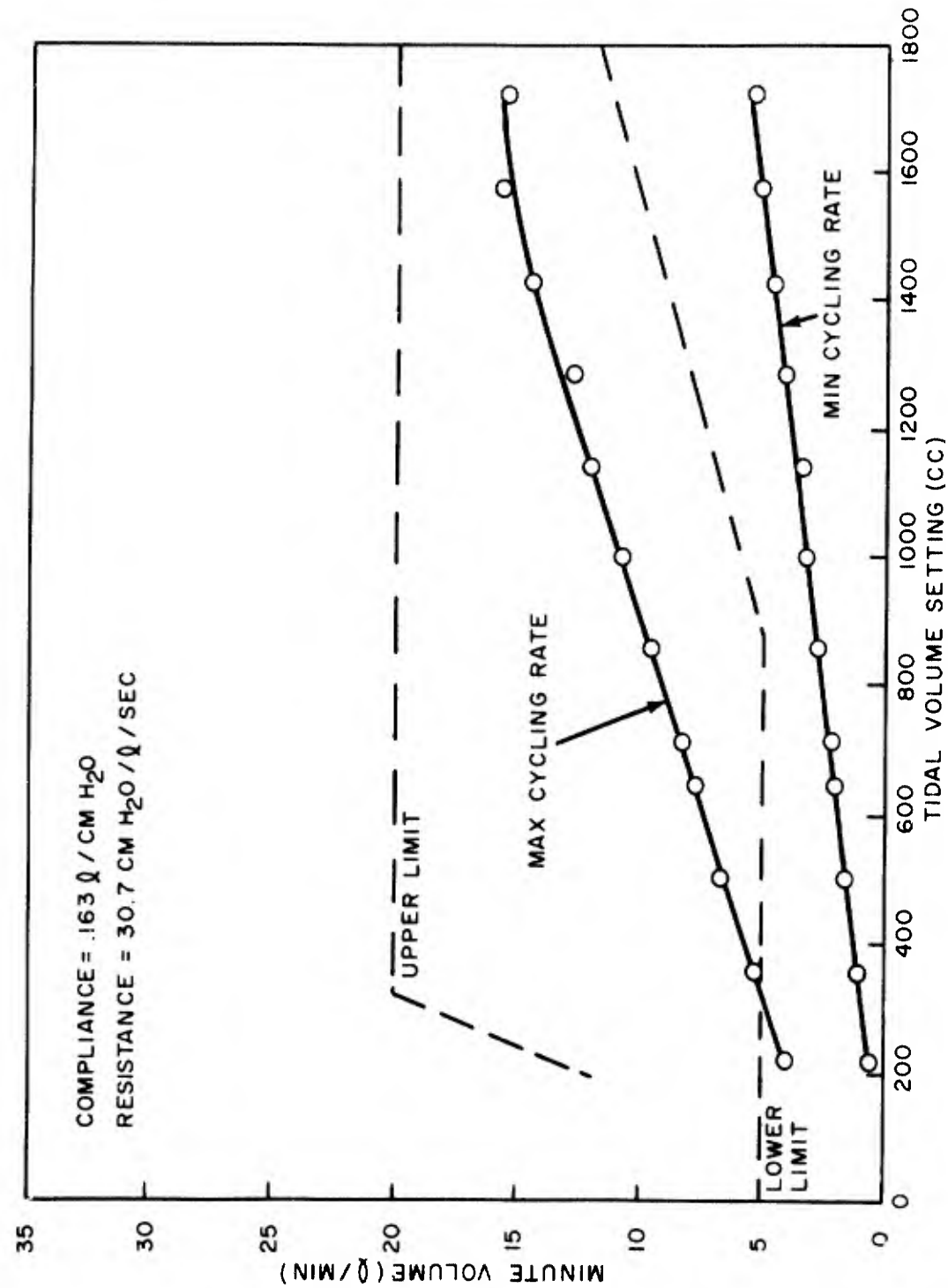


Figure 21. Minute volumes - high compliance, high resistance.

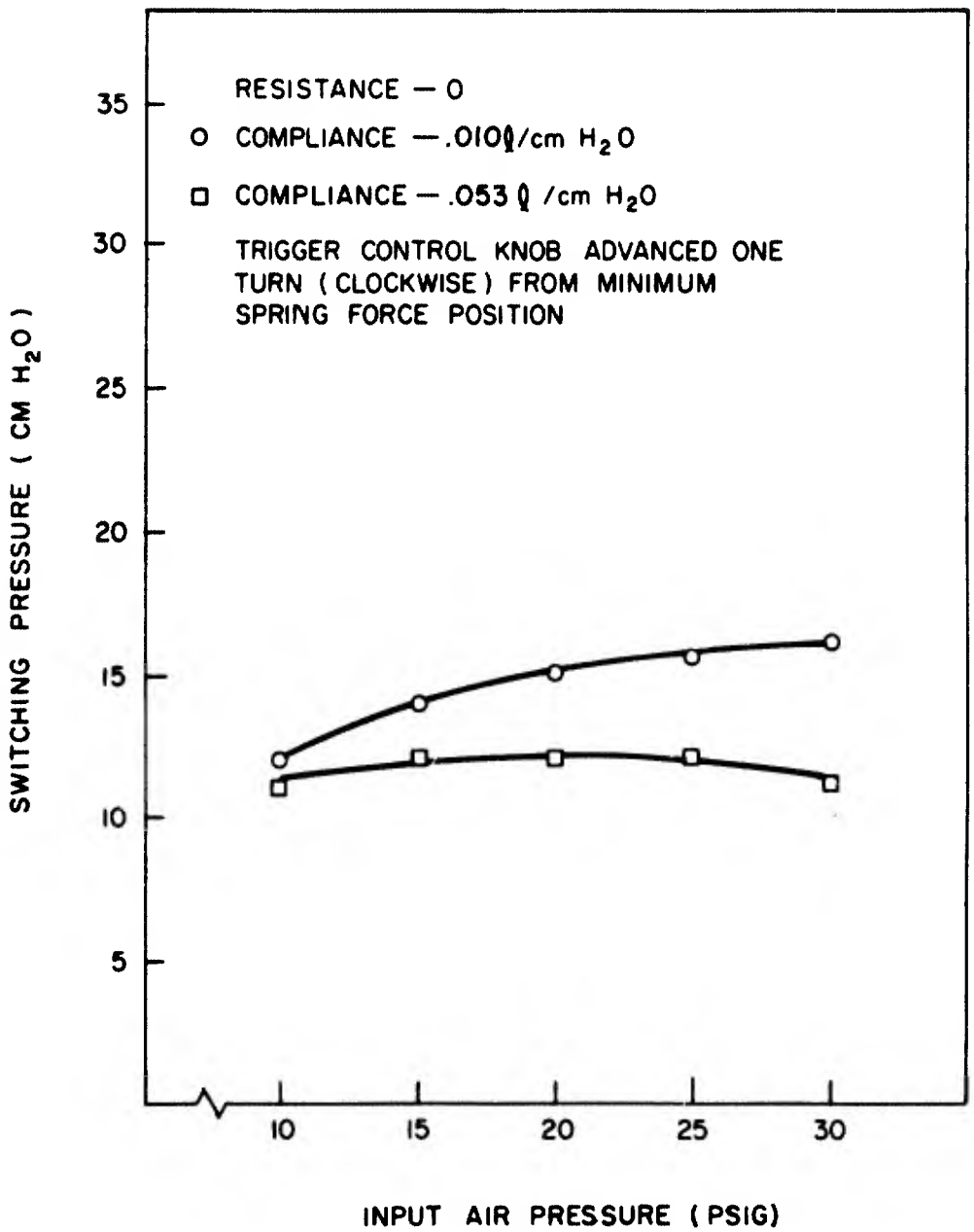


Figure 22. Dependence of pressure-cycling on input pressure and loading.

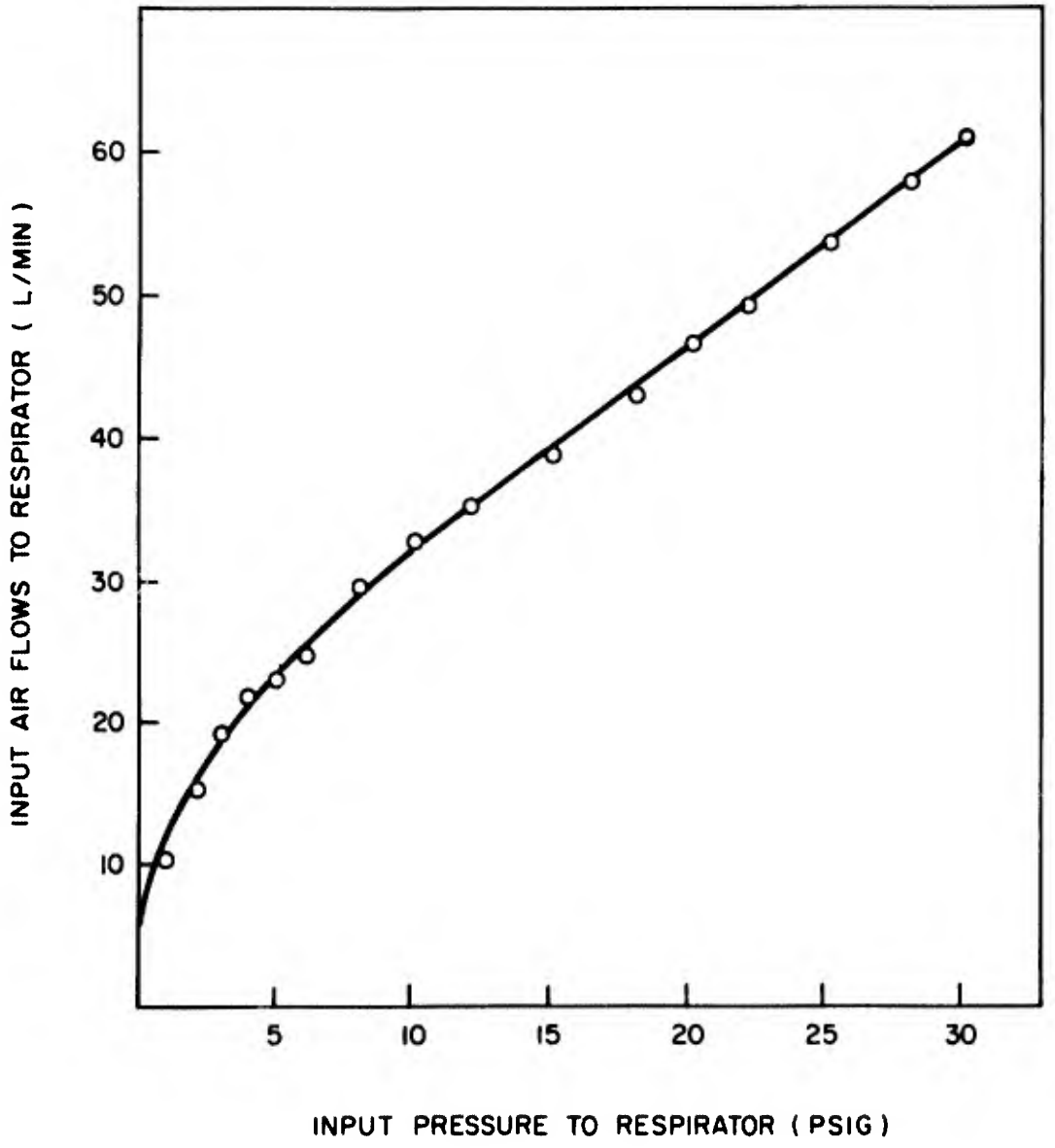


Figure 23. Air consumption rate.

Table I. Tank Compliances (ℓ/cm H_2O)

Tank volume (ℓ)	C_1	C_2	C'
11	0.010	0.010	0.011
76	0.052	0.053	0.074
236	0.161	0.164	0.230

C_1 = measured compliance for injection time of 0.5 sec

C_2 = measured compliance for injection time of 2.0 sec

C' = calculated compliance from previous analysis (using isothermal relationship)

BLANK PAGE

UNCLASSIFIED

Security Classification

DOCUMENT CONTROL DATA - R & D

(Security classification of title, body of abstract and indexing annotation must be entered when the overall report is classified)

1. ORIGINATING ACTIVITY *(Corporate author)*

Harry Diamond Laboratories
Washington, D.C. 20438

2a. REPORT SECURITY CLASSIFICATION

Unclassified

2b. GROUP

3. REPORT TITLE

REVISED PERFORMANCE EVALUATION OF THE ARMY VOLUME-CYCLED RESPIRATOR, MODEL 2

4. DESCRIPTIVE NOTES *(Type of report and inclusive dates)*

5. AUTHOR(S) *(First name, middle initial, last name)*

J. W. Joyce

6. REPORT DATE	7a. TOTAL NO. OF PAGES	7b. NO. OF REFS
July 1968	46	1

8a. CONTRACT OR GRANT NO.	9a. ORIGINATOR'S REPORT NUMBER(S)
b. PROJECT NO.	TM-68-17
c. AMCMS Code: 5900.21.11123 d. HDL Project No. 31000	9b. OTHER REPORT NO(S) *(Any other numbers that may be assigned this report)*

10. DISTRIBUTION STATEMENT

This document has been approved for public release and sale; its distribution is unlimited.

11. SUPPLEMENTARY NOTES	12. SPONSORING MILITARY ACTIVITY
	Army Materiel Command Walter Reed Army Institute of Research

13. ABSTRACT

The Army volume-cycled respirator is designed to assist or control the ventilation of patients. A previous analysis of this respirator contained erroneous information that led to quantitatively incorrect performance results. This report describes the errors made and presents both revised and new performance results.

The tests performed indicate that the requirements for cycling rates, pressure-cycling, and assist sensitivity have been satisfied. The requirements for tidal volumes, minute volumes, and inspiratory times have not been fully met. The failure of the respirator to meet tidal volume and minute volume requirements was traced to excessive internal compliance of the respirator. This compliance results in wide variations in delivered tidal volumes with changes in patient load conditions and is considered an undesirable feature for a volume-cycled respirator.

DD FORM 1 NOV 68 1473 REPLACES DD FORM 1473, 1 JAN 64, WHICH IS OBSOLETE FOR ARMY USE.

UNCLASSIFIED

Security Classification

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