The following is the abstract of the article discussed in the subsequent letter:

**Narusawa, Uichiro.** General characteristics of the sigmoidal model equation representing quasi-static pulmonary P-V curves. *J Appl Physiol* 91: 201–210, 2001—A pulmonary pressure-volume (P-V) curve represented by a sigmoidal model equation with four parameters, \( V(P) = a + b(1 + \exp[-(P - c)/d])^{-1} \), has been demonstrated to fit inflation and deflation data obtained under a variety of conditions extremely well. In the present report, a differential equation on \( V(P) \) is identified, thus relating the fourth parameter, \( d \), to the difference between the upper and the lower asymptotes of the volume, \( b \), through a proportionality constant, \( \alpha \), with its order of magnitude of \( 10^{-4} \) to \( 10^{-5} \) (in \( \text{ml}^{-1} \cdot \text{cmH}_2\text{O}^{-1} \)). When the model equation is normalized using a nondimensional volume, \( \hat{V} = (1 < V < 1) \), and a nondimensional pressure, \( \hat{P} = (P/c - 1) \), the resulting P-V curve depends on a single nondimensional parameter, \( \Lambda = abc \). A nondimensional work of expansion/compression, \( \hat{W}_{1-2} \), is also obtained along the quasi-static sigmoidal P-V curve between an initial volume (at 1) and a final volume (at 2). Six sets of P-V data available in the literature are used to show the changes that occur in these two parameters \( \Lambda \) (defining the shape of the sigmoidal curve and \( \hat{W}_{1-2} \) accounting for the range of clinical data) with different conditions of the total respiratory system. The clinical usefulness of these parameters requires further study.

**Development of an Algorithm for Improving the Description of the Pulmonary Pressure-Volume Curve**

*To the Editor:* Narusawa (2) reported on a carefully designed study performed to examine the mathematical basis of a sigmoidal pulmonary pressure-volume (P-V) curve as introduced by Venegas et al. (3). The author applied a dimensional analysis in order to relate clinical data to the corresponding shape and range of the pulmonary P-V curves. He concluded that his analysis provides a generalized understanding of respiratory system mechanics.

In the study of Narusawa (2), the symmetric Venegas algorithm

\[
V = a + b(1 + \exp[-(P - c)/d])^{-1}
\]  

was used, where \( V \) is the lung volume and \( P \) is the respective pressure of the respiratory system. The parameter \( a \) denotes the lower asymptote volume, \( b \) represents the vital capacity, \( c \) is the point of maximal compliance (true inflection point), and \( d \) reflects the pressure range that includes most of the volume change. Until now, it has been commonly supposed that the above model equation is in exact correspondence with data obtained from healthy humans, patients suffering from lung injury, or animals.

However, we recently demonstrated (1) that a modification of the equation leads to a further improvement of the P-V algorithm

\[
V = a + b(1 + \exp[-(P - c)/d])^{-S}
\]  

by using the additional exponent \( S \) to ameliorate the smoothness of the sigmoid. In our animal study, we evaluated the applicability of both algorithms to P-V data obtained in 15 artificially ventilated rabbits. The P-V data were recorded by inflating and deflating rabbit lungs in discrete volume steps (4 ml BTPS) with a home-made computerized ventilatory servo system. The pressure of the respiratory system was determined during breath-holds (1 s).

The additional use of the exponent \( S \) slightly improved the mean goodness-of-fit coefficients \( R^2 \) to \( R^2 > 0.997 \). Although both terms revealed almost identical mean values of vital capacity and maximal compliance, we found a significantly higher mean pressure at the true inflection point obtained from the Venegas equation \((3.5 \pm 0.6 \text{cmH}_2\text{O})\) than by using our own modification \((2.9 \pm 0.6 \text{cmH}_2\text{O})\). The corresponding mean values of lung volume were also different.

Because both algorithms were applied to identical data recorded in healthy animals, the precision of the Venegas model is called into question. It remains to be seen whether our modification proves an adequate equation to fit P-V curves to patients’ data.

**REFERENCES**


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**REPLY**

*To the Editor:* Venegas and co-workers showed that the quasi-static P-V curve may be represented by a continuous function (vs. the piecewise continuous functions of previous efforts), which led to my dimensional analyses.
aimed at quantifying various characteristics of P-V curves. Similar analyses may be applied to the modified equation proposed by the authors of the letter.

The modified P-V equation (Eq. 2 in the letter), if normalized, yields two nondimensional parameters, \( c/d \) (the only parameter in the normalized Venegas model equation) and \( S \). Also Eq. 2, unlike the Venegas model equation, is not symmetric with respect to \( P = c \). These characteristics of Eq. 2 provide greater flexibility for curve fitting.

The P-V data sets used in the letter are obtained for artificially ventilated rabbits, and the reported pressure at the true inflection point \( P_0 \) in Ref. 1 is of the order of 3 cmH\(_2\)O. On the other hand, the accuracy of the Venegas equation has been shown, from extensive P-V data analyses reported in Refs. 8 and 15 of my original paper (1), to be more comprehensive. In terms of the magnitude of \( P_0 \), their analyses covered a range of 7 (for dog lungs) to 36 cmH\(_2\)O (for a patient with acute respiratory distress syndrome). Therefore, as pointed out by the authors of the letter, further examinations of various P-V data are required to confirm the accuracy of the proposed modification to the Venegas model equation.

Ideally, changes in the intrarespiratory system conditions should be examined along a specified P-V curve and related to parameters of the model equation, which would provide the physiological meaning of each parameter, thus elevating the model equation beyond the status of an experimental correlation.

REFERENCES


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