AN ANALYSIS OF MAXIMAL EXPIRATORY FLOW-VOLUME CURVES BASED ON AN ANALYTICAL MODEL OF LUNG MECHANICS

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Maximal expiratory flow-volume (FV) curves were analyzed using a mathematical model of lung mechanics. In the lung model, the parenchyma contained an array of membranes, and geometrical properties were derived from morphological data. The elastic properties were deduced by requiring the bulk expansion of the array to match the average pressure volume curve for the lung. Nonuniform deformations were analyzed. Performance of the model was tested by altering viscous flow losses, airway mechanisms and the viscoelastic and plastoelastic effects of lung tissue. Plateaus on isovolume pressure flow and FV curves based on lung pressure-volume were predicted and normal maximal flow ranges were established. The authors note the success and achievements of the model and its drawbacks. The analytical model is capable of predicting thoracic cavity and lung interaction related to ventilation and pleural pressure gradients.

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ABSTRACT

The purpose of this study was to refine our analytical (mathematical) model of the mechanical properties of the lung primarily to better understand the flow-volume (FV) curve. Excised lungs were found by several experimental approaches to have a low shear modulus, a Poisson ratio between 0.40 and 0.45, and to exhibit relatively little interdependence. It has been possible with refinements in the model (1) to predict plateaus on isovolume pressure-flow curves and (2) to predict FV curves based on bronchial pressure-area and lung pressure-volume data that agree well with measured curves from excised human lungs. Normal ranges of maximal flow based on 200 subjects have been established but normal variability is large. The analytic model offers promise of predicting interactions between thoracic cavity and lung as they pertain to regional ventilation and pleural pressure gradients.

INTRODUCTION

1. Purpose and Scope of Present Work:

The purpose of this contract was to develop, refine, test, and validate an analytical (mathematical) model of the mechanical properties of the lungs in order to better understand the determinants of the maximal expiratory flow-volume curve.

The scope of work undertaken during the contract period may be divided into the following phases:

1. Testing our original analytical model's performance by altering the following parameters:

   A. Viscous flow losses: The first step was to carry out a more complete analysis of the velocity profile and viscous losses for expiratory flow in the bronchi, especially at low lung volumes.

   B. Airway mechanics: Determination was made of which nonlinear effects are most important causing airway compression and including these in the analytical model so as to bring the model's predicted maximum flow at high lung volumes into conformity with experimental data.

   C. Viscoelasticity: Incorporation of viscoelastic and plasto-elastic effects of lung tissue into the analytical model.

2. Maximal expiratory flow-volume curves and static pressure-volume curves were obtained on excised dog lungs, and excised normal human lungs at post-mortem, to test the ability of the current and modified analytical models to predict these curves.

3. Maximal expiratory flow-volume curves and other lung mechanics data were obtained in man, using breathing mixtures of differing density and viscosity and testing the analytical model's prediction of these data.

4. Data were obtained on normal humans of varying age to establish empiric normal values as well as provide material for the development of normal predictive values from the analytical model.

5. Preliminary steps were made in applying the modified analytical model to certain disease states.

2. General Background:

The flow versus volume presentation of a maximal expiratory forced vital capacity maneuver was first described by Hyatt, Schilder, and Fry in 1958 (1). These investigators demonstrated that maximal expiratory flow was largely effort dependent only near full inflation and that over the major portion of the vital capacity maximal expiratory flow was uniquely determined by the mechanical properties of the lungs. Evidence was also presented that the effort-independent portion of the flow-volume curve reflected the behavior of the intrathoracic portions of the respiratory system, and that lung elastic recoil was an important determinant of maximal expiratory flow (2).

Description of the flow-volume curve stimulated a search for the site of expiratory flow limitation. Fry (3) identified the major determinants of the flow-volume curve. Macklem and associates (4) defined the site of flow-limitation over the upper two-thirds of the vital capacity as residing in the trachea and main stem bronchi, and noted that it was associated with a rather distinctive shape of the flow-volume curve. Examination of the flow-volume curve at low lung volumes led Leith and Mead (5) to formulate an hypothesis regarding the factors that determine residual volume. The flow-volume curve has been particularly useful in the detection of lesions of the major airway (6) and in the detection of bronchoconstriction induced by inhalation of allergens or pharmacologic agents (7). The flow-volume curve has been shown to be a more sensitive test for detecting the presence of flow-limitation produced by disease of the intrapulmonary airways than the usual tests of ventilatory function via forced vital capacity (FVC), peak flow, one second forced expiratory volume (FEV_1) (8).

However, a serious lack of knowledge of the precise mechanical factors that determine maximal expiratory flow and the unique characteristics of the flow-volume curve has limited the usefulness of this technique. The assumptions underlying Fry's model of the mechanical behavior of the lung needed to be tested. The validity of the present model must be checked by direct measurements of flow-volume curves in normal human subjects and in subjects whose mechanical properties of the lungs have been altered by disease. Successful completion of the proposed model and its validation should enable investigators using the flow-volume curve to quantify in detail the mechanical properties of the lungs and detect early abnormalities in these functions caused by obstructive airway disease.

This could have broad application to the field of occupational disease because the flow-volume (FV) curve offers a means to accomplish large-scale field studies for the detection of changes in resistance. In addition, it has been used as a means of detecting the effects of acute exposure in workers who have been exposed to a variety of agents (for example, cotton dust, coal dust, isocyanates, etc.) that affect flow resistance in small airways.

It is appropriate at this point to review rather briefly the preliminary work that we had done in the area of lung modeling, the results obtained, and the areas most in need of further refinement.

It seemed unlikely that it would be possible to measure directly all of the pressures, flows and deformations of the lung which would be required to give a complete description of the mechanisms that determine maximum expiratory flow. Furthermore, if the FV curve is to be used as a diagnostic test, some means of making inferences from limited data is required. A successful mathematical model would describe the interaction between flow and lung that determines expiratory flow, uncover general relations between the mechanical variables, interpret the properties of the flow in terms of mechanical and geometrical properties of the lung, and suggest tests which would be sensitive to variations in particular parameters of the system.

The model which we proposed to test and to develop into a reliable basis for interpretation of the expiratory flow characteristics consisted of two parts: First, a model for the mechanical properties of the parenchyma and airways which can be used to analyze deformations of the bronchi and surrounding parenchyma due to a decrease in pressure within the airway; and second, a model for calculating the pressure drop along the airways during expiratory flow. The two parts were, of course, coupled in the calculation of expiratory flow. In the following paragraphs we give a brief review of the model, a critique of the results obtained to date and a list of some of the weaknesses of the initial model. Under Research Plan we discuss the improvements and tests which we expected were needed to increase the accuracy of the model and establish its usefulness.

In the model for the mechanical properties of the lung, the parenchyma was pictured as an array of interconnected plane elastic membranes. The geometrical properties of the membranes were taken from the morphological data reported by Weibel (9) and the elastic properties were deduced by requiring that the bulk expansion of the array match the average pressure-volume curve for the lung. The elastic properties of these membranes was initially taken to be very similar to the elastic properties of cat mesentery observed by Hildebrandt et al. (10). The forces in the membranes were then averaged over all membrane orientations and the relation between the stress in the array and the deformation of the array obtained. These have the same form as the constitutive equations for an elastic continuum. The elastic constants in these relations were obtained in terms of the elastic constant of the membranes which are, in turn, known in terms of the pressure-volume curve for the whole lung. At this point, the techniques of continuum mechanics were brought to bear to analyze the nonuniform deformations which are of interest. A model for the elastic properties of

the bronchi was also formulated by assuming that the bronchi have the elastic properties which would result in the bronchi expanding similarly to the lung during a uniform expansion of the lung.

The model had been applied to two problems. First, the change in volume due to a change in pressure in a small region within the lung was calculated and compared with measurements by Woolcock and Macklem (11) of the specific compliance of small regions in the lung. The results from the model gave excellent agreement with the measurements taken along the deflationary limb of the pressure volume curve. The model did not satisfactorily predict the measurements taken during inflation of the whole lung. Second, the model was applied to the problem which is pertinent to flow limitation, the decrease in area of a bronchus with a decrease in pressure within the bronchus. The predictions of the model showed a considerable influence of the surrounding parenchyma in opposing the deformation of the bronchus, especially at high lung volume.

In the model for the flow through the airways, the pressure drop which balances the viscous forces acting at the airway wall was calculated using the Poiseuille formula. This gives a lower limit for the viscous pressure loss for a given flow. The pressure drop associated with the convective acceleration which occurs as the cross sectional area decreases in the flow direction was calculated using the average flow speed in the Bernoulli formula.

Isovolume pressure-flow ($IVP\dot{V}$) curves were computed by choosing a lung volume: this sets the values of the mechanical parameters of the model; choosing a volume flow ($\dot{V}$) and calculating the pressure drop along the bronchial tree. For each generation of the bronchial tree, the change in airway diameter due to the pressure drop to that generation was taken into account. Then a different value of $\dot{V}$ was chosen and the computation repeated. Below a critical value of $\dot{V}$, two solutions existed to the simultaneous equations for the change in airway radius with pressure and change in gas pressure with airway radius. Above that critical value of $\dot{V}$, which was identified as maximum expiratory flow, no solution existed. The calculated curves were only carried to the point where a radius change of 50% in the third generation had been computed. The computed flows fell off at higher driving pressure but the accuracy of the model was questionable at large deformations. Most of the pressure drop was computed to occur at the entrance to the third generation and the largest deformations occurred in these and the larger airways. The calculated viscous pressure drop was small compared to the Bernoulli drop and a simplified algebraic calculation was made to obtain the following formula for the predicted maximum expiratory flow.

$$\dot{V}_{max} = \frac{32}{25} \left(\frac{2\ pi^2\ d_3^4}{\rho\ k}\right)^{\frac{1}{2}}$$

Here, d_3 is the diameter of the undeformed airways of the third generation at the given lung volume, ρ is the gas density, p is pressure, and k is an elastic parameter which can be evaluated at each lung volume from the pressure-volume curve.

The model was successful in the following ways.

1. It was quantitative.

2. It was based on an idealization of the lung structure and the correspondence between the features of the model and the real system was clear.

3. All of the parameters were evaluated from independent information. There were no parameters left to be adjusted to fit the flow curves.

4. The predictions of expiratory flow were successful in that:

   a. A maximum flow was predicted.

   b. The predicted values of maximum flow and the pressure at which maximum flow is achieved were reasonable.

   c. The predicted dependence on gas density was accurate.

The following features of the model could be identified as possible sources of inaccuracy.

1. The visco-elastic or plasto-elastic properties of the structural elements had been ignored. These effects, as evidenced by the hysteretic properties of the lung, are not negligible.

2. A number of linear approximations had been made in averaging the forces in the membranes. This means that the model gave a linearized description of nonuniform deformations and could be inaccurate if the size of the deformation was large. Large deformations occur at high driving pressures.

3. The Weibel model for the geometry of the bronchial tree was convenient but may not be an accurate enough description of the real system. The effects of asymmetry or nonuniformity in the branching pattern and in the other lung properties needed to be checked.

4. The model used for the mechanical properties of the bronchi was also motivated partly by a desire to keep the model simple. It needed to be checked against direct measurements. The present model was not consistent with Hyatt's (12) measurements on excised dog bronchi.

5. The assumption used in calculating the viscous pressure drop, namely, that the flow was fully developed, gave the minumum pressure drop that could occur for a given flow. This may be a poor estimate.

The general logic of the approach we proposed was as follows: Relate the present model to our current normal data in man after making theoretical changes in the model by incorporating the viscous pressure losses in the flow through the compressed segment. After determining the importance of this change, incorporate a more precise description of the tube mechanics based on currently available data (12). This step should identify the need for additional experimental data on bronchial mechanics. Next the model was to be expanded to include a realistic description of the visoelastic behavior of the lung. Certain excised lung studies would permit a validation of the theoretical treatment of this parameter. Then pertinent nonlinear characteristics would be incorporated to evaluate the importance of large deformations during dynamic airway compression. Finally it might become important to seek more precise descriptions of the airway geometry and the branching behavior than are currently available.

At each step in the above sequence the modified model was to be compared to normal human and animal data. Other ways of validating the model are to change the respired gases (altering density and viscosity) and determine how well the model predicts the resulting FV curve. These studies should then put us in a position to apply the model to patients with early, and advanced, obstructive and restrictive lung disease.

3. Description of Various Phases of Project: See "Research Plan" and "Results and Interpretation."

RESEARCH PLAN

1. Tests of Analytic Lung Model:

We believed that the viscous pressure drop has been badly underestimated in the above analysis. The first step in our program was to carry out a more complete analysis of the velocity profile for the flow in the bronchi during expiration. The Bernoulli pressure drop in the flow direction tends to produce a flatter velocity profile in the center of the tube and a steeper velocity gradient at the wall. This analysis was carried out, with some effort, using familiar boundary layer analysis techniques. We hoped this would bring the predictions at low lung volumes into agreement with experimental data. If so, the results would be checked by testing the changes predicted to those observed with gases of different viscosity. The existing literature in this area was not sufficient to test the model.

Second, the approximations made in averaging the forces in the membranes were reviewed in order to determine which nonlinear effects were most important and to include them in the analysis of airway compression. We believed these nonlinear corrections might result in an increased stiffening effect of the parenchyma at large deformations and result in a leveling off of the predicted IVPV curves and a slight increase in predicted maximum flow at high lung volume.

Third, a program for modifying the description of the properties of the membranes to include visco-elastic and plasto-elastic effects was organized. This required consultation with individuals such as Fung and Hildebrandt and the specialized knowledge in this field which Professor Warner provided. Additional experimental information on the behavior of biological membranes, sections of lung tissue, and excised animal lungs under various loadings was needed to test the model before it could be used with confidence.

The accuracy of the description of the mechanical and geometrical properties of the airways was checked and the sensitivity of the predictions of the model to these parameters investigated.

2. FV Curves in Excised Lungs:

We obtained FV curves by applying negative pressure to the airways of excised dog lungs suspended in a chamber that had an attached spirometer. Flow was measured at the airway and corrected for gas decompression. Airway minus chanber pressure provided transpulmonary pressure. Static transpulmonary pressure during inflation and deflation was measured with this preparation. Absolute lung volume was measured by water displacement, weighing of the lung, and a knowledge of lung specific gravity. Similar studies were obtained on normal human lungs that we succeeded in obtaining at postmortem. As needs dictated, these excised lungs were ventilated with a wide variety of gas mixtures.

3. FV Curves with Various Gases:

A useful way to check the model was to alter in a known fashion the physical parameters of the system, predict from the model the FV curve and compare predicted to experimentally measured curves. The most easily altered parameters were the physical properties of the respired gas. These studies were performed in a body box. Flow was measured by a device at the mouth that was not sensitive to change in gas density or viscosity (Wedge Spirometer).

4. Normal FV Data in Man:

We obtained approximately 200 FV curves from normal subjects using a volume-displacement body plethysmograph. These spanned the ages of 12-65 years, including males and females, smokers and non-smokers. We submitted these curves to the following emperic analyses:

a. Established normal prediction values for $\dot{V}$max at 75-50-25% VC and TLC with flow normalized in VC's and TLC's per second.

b. Examined the slope of the FV curve as a function of age, sex, and smoking habit. These data provided emperic normative data, currently not available, for the FV curve.

5. Application of Model to Disease:

Once we convinced ourselves that the model was applicable to normal man and animal, testing was begun in patients with various stages of obstructive lung disease.

RESULTS AND INTERPRETATION

1. Tests of Analytic Lung Model:

A. Viscous flow losses -- An analysis of this problem using the Weibel geometry showed that for flows greater than 4 l/s the non-parabolic velocity profiles in the intermediate bronchi and turbulence in the large bronchi resulted in a viscous pressure drop of more than twice that predicted from the Poiseieulle formula. However, this additional pressure drop was still not sufficient to decrease predicted flows at low lung volumes significantly.

Other flow effects were considered such as the mixing of flow coming from regions with different alveolar pressures. Preliminary analysis indicated that the pressure due to this effect would also be small.

B. Airway mechanics -- Work in this area has occupied a considerable amount of time. Pressure-diameter (PD) behavior of human bronchi was studied in 21 specimens. PD behavior was comparable to that found in the dog, namely, large airways were less compliant than small airways. Furthermore, measurements of the total cross-sectional area of all parallel pathways of the larger airways in isolated human lungs showed that the point of minimum area lies in the trachea or main stem bronchi for positive airway distending pressure -- not in the segmental bronchi as reported by Weibel. For negative transmural pressure, the point of minimum area moves peripherally because of the greater compliance of smaller airways.

Based on the measured PD behavior of airways the model failed to show plateaus on isovolume pressure-flow (IVPV) curves. This meant that to obtain plateaus the lung would need to stiffen bronchi undergoing compression -- that is, that there was considerable "interdependence" between bronchi and parenchyma. To estimate this effect it became necessary to obtain precise estimates of the elastic constants of the lung, namely the bulk modulus and Young's modulus. Several approaches were used on excised dog lobes and a few human lungs. Specimens were subjected to uniaxial loading, indentation tests and the production of artificial holes in the parenchyma. An additional study using excised portions of "trapped" lobes provided a validation of these experimental approaches. More details of certain of these procedures are described in the attached manuscript. However, in brief, all tests showed Young's modulus to be small compared

to the bulk modulus. This indicated that the lung is more easily deformed in shear than in dilitation with a Poisson ratio between 0.40 and 0.45. Our constants differ from those of the Boston group for reasons not yet apparent. Nevertheless, when applying these constants to the model it was found that interdependence was too small to alter the geometry of compressed airways sufficiently to produce plateaus.

This led to a consideration of other models of flow-limitation stimulated in large part by Dawson's "choke point" concept (13). In brief, this concept says that there are critical points at which total airway area does not change with axial position while airway compliance decreases in the direction of flow. The result is that $\dot{V}$ can remain constant as driving pressure increases, thus producing a plateau on an IVP$\dot{V}$ curve. Modification of this idea using bronchial areas measured from human lungs looks very promising in predicting FV curves. An extension of these ideas in terms of axial tension developed at "choke" points is presented in an attached manuscript.

Other studies were also conducted to evaluate interdependence. We were not successful in perfecting an isovolume, perfused, excised lung preparation. On the other hand, we have used trapped lobes to estimate interdependence (see attached abstract) of bronchi and again find results compatible with the above findings of a low shear modulus.

We now feel we have accurate estimates of the elastic constants of the lung and a more promising approach to predicting maximal expiratory flow based on bronchial cross-sectional area and lung recoil pressure.

C. Other Accomplishments in this Area -- The role of the pleura in determining the lung PV behavior is under study. It appears that at low inflating pressures a considerable portion of the recoil pressure may be due to the pleura. An analysis of lung deformation due to gravity based on engineering analysis has shown good agreement with experimentally obtained data on excised dog lobes. We also developed an on-line computer program to measure static PV curves in human subjects. And finally a strain energy function was developed to describe large deformations (manuscript attached).

2. Flow-Volume Curves of Excised Lungs:

We first started obtaining FV curves on excised dog lungs but found these too variable to be of value. However, we became interested in the effect of rendering the lung "stiff" (increased lung recoil) on bronchial PD behavior. The results are described in the attached reprint (J. Appl. Physiol., 39:429, 1975).

Most of our effort has been directed toward predicting maximal flow ($\dot{V}max$) in excised human lungs. We have studied twenty but only in the last several did we obtain adequate data on bronchial area as a function of bronchial generation; We also measured static PV behavior and obtained FV curves on air, 80% He - 20% O_2, and 100% SF_6. Flows were predicted by the above model utilizing the interaction between flow (Bernoulli pressure loss) and bronchial pressure area behavior. The model was successful in predicting the plateau in the IVP$\dot{V}$ curve, in predicting a shift of the flow-limiting site toward the periphery at low lung volumes, and in predicting the value of maximum flow at higher lung volumes. It was found that the predicted flows were too high at lower lung volume and that a viscous (rate-dependent) loss of pressure was required to achieve acceptable predictions at lower lung volumes. This suggested a viscous pressure loss due to either viscous flow losses in peripheral airways or a visco-elastic or stress-relaxation behavior of the lung parenchyma. Theoretical estimates of viscous flow losses predicted that these would be negligible. This implied that a rate-dependent tissue behavior was significant and that dynamic recoil pressure was less than that measured statically. One study addressing this point, in which airway pressure was measured after stopping flow at various points on the maximum flow curve, showed as much as 3 cm H_2O difference between static and dynamic recoil. For this one case for which we obtained all the required data; namely, PD behavior of airways of the first few generations and a dynamic P-V curve for the lung, the predicted flows showed good agreement with increased flow over most of the range of lung volume. Reduced recoil during dynamic maneuvers might account for supermaximal flows which are observed in partial FV curves and for much of the decrease in the dependence of flow on gas density at lower lung volumes, which is of current interest in using air versus He-O_2 FV curves in man as an index of early airway disease.

Finally, we have noted that maximal flows are higher in lungs that are hanging compared to $\dot{V}max$ when the lung is supported. This would be consistent with the concept mentioned above that axial tension in the airways can influence $\dot{V}max$. It also indicates that how the lung is supported in the chest can effect $\dot{V}max$.

3. FV Curves with Various Gases:

Work in this area has lagged as efforts were directed toward developing a more accurate lung model. However, we did perform a preliminary study on 20 young male smokers matched by age with 20 non-smokers comparing air and He-O_2 FV curves. No difference between the groups could be detected. This study was preceeded by an evaluation of the within-day and between-day variability of the He-O_2 versus air FV curves. Variability was acceptably small.

We have also studied 20 subjects with obstructive (non-asthmatic) disease (maximal midexpiratory flow less than 60% of predicted). About 50% of these subjects showed normal density dependence. Seventeen of these subjects were studied before and after bronchodilator therapy. Those

subjects who showed little density dependence before therapy increased their density dependence after therapy. The converse was true for those showing normal density dependence before therapy. These findings suggest that the site of flow-limitation may vary with the gas breathed and may move considerably with bronchodilator therapy.

4. Normal FV Data in Man:

The attached reprint (Black et al., Amer. Rev. Resp. Dis., 110:282, 1974) describes our efforts in this direction. In brief, our findings are that there is a considerable variability among normals for $\dot{V}max$ even after correction for age, sex, and height. It seems likely that some estimate of airway size would improve prediction. However, a preliminary study of treacheal diameters (measured from standard chest x-rays) did not improve our prediction accuracy. Nevertheless, further exploration along these lines seems warranted since refinement in prediction of FV curves is required before the full potential of this test in comparing subjects, such as in epidemiologic studies, is realized.

5. Applications of Model to Disease:

This task has lagged since we are just arriving at the point where we have sufficient confidence in the ability of the model to predict $\dot{V}max$ in normal lungs. Some baseline data and experience have been obtained in the $He-O_2$ FV curve studies mentioned above. However, further advances in this area must await confirmation of our analytic model. We feel confident that this will eventually be achieved but fully recognize the problems in studying the diseased lung with its gross nonuniformities of mechanical properties.

CONCLUSIONS AND RECOMMENDATIONS

1. Excised lungs of dog and man show a low shear modulus and a Poisson ratio greater than 0.4. This indicates the isolated lung shows relatively low resistance to shear and hence a relatively low magnitude of "inter-dependence."

2. All our studies to date have indicated that the principles of continuum mechanics can be applied to the lung.

3. Plateaus on isovolume pressure-flow curves can be modelled using the "choke" point concept or a modification of this approach based on measured bronchial pressure-area curves for the first few bronchial generations.

4. The predominant flow regimen in normal lungs is convective acceleration. The magnitude of the viscous pressure loss varies among normal lungs. The cause of the viscous loss may relate in part to a time-dependent effect of lung elastic recoil (tissue viscosity). This would play a role at intermediate and low lung volumes. Viscous flow losses would become important at low lung volumes.

5. The study has identified the need to improve the ability to predict normal maximal expiratory flows. This is not a major problem for studying a given individual but is of major importance in studying populations. Attention should be directed toward measuring additional parameters, such as airway size and dynamic recoil, that would narrow the currently available range of normality.

6. It is important to continue refining our current analytic model of lung mechanical behavior. Not only is this important in terms of forced expiration and the FV curve but also in terms of understanding such problems as intrapulmonary stress distribution, the effects of non-homogenieties on lung function, and the matching of lung to thoracic cavity -- this latter problem is particularly pertinent to understanding the origin of the pleural pressure gradient and to understanding the factors that can alter regional lung volumes and regional ventilation.

References

1. Hyatt, R. E., Schilder, D. P., and Fry, D. L.: Relationship Between Maximum Expiratory Flow and Degree of Lung Inflation. J. Appl. Physiol., 13:331, 1958.

2. Dayman, H.: Mechanics of Airflow in Health and Emphysema. J. Clin. Invest., 30:1175, 1951.

3. Fry, D. L.: A Preliminary Lung Model for Simulating the Aerodynamics of the Bronchial Tree. Comp. Biomed. Res., 2:111, 1968.

4. Macklem, P. T. and Mead, J.: Factors Determining Maximum Expiratory Flow in Dogs. J. Appl. Physiol., 25:159, 1968.

5. Leith, D. and Mead, J.: Mechanisms Determining Residual Volume of the Lungs in Normal Subjects. J. Appl. Physiol., 23:221, 1967.

6. Miller, R. D. and Hyatt, R. E.: Obstructing Lesions of the Larynx and Trachea: Clinical and Physiologic Consideration. Mayo Clin. Proc., 44:145, 1969.

7. Olive, J. T. and Hyatt, R. E.: Maximal Expiratory Flow and Total Respiratory Resistance During Induced Bronchoconstriction in Asthmatic Subjects. Amer. Rev. Resp. Dis., 106:366, 1972.

8. Lapp, N. L., Hankinson, J. L., Burgess, D. B., and O'Brien, R.: Ventilatory Capacity After a Shift in Coal Miners. Arch. Environ. Health., 24:204, 1972.

9. Weibel, E. R.: Morphometry of the Human Lung. Academic Press. New York, 1963.

10. Hildebrandt, J., Fukaya, H., and Martin, C. J.: Stress-Strain Relations of Tissue Sheets Undergoing Uniform Two-Dimensional Stretch. J. Appl. Physiol., 27:758, 1969.

11. Wookcock, A. J. and Macklem, P. T.: Mechanical Factors Influencing Collateral Ventilation in Human, Dog, and Pig Lungs. J. Appl. Physiol., 30:99, 1971.

12. Hyatt, R. E.: Bronchial Mechanics. In: Current Research in Chronic Airway Obstruction: Ninth Aspen Emphysema Conference. Public Health Service Publication No. 1717, Washington, D. C., 1968.

13. Dawson, S. V. and Elliot, E. A.: Expiratory Flow Limitation at Wave Speed. Fed. Proc., 33:324, 1974.

APPENDIX ITEMS

1. Black et al.: Variability in the Maximal Expiratory Flow Volume Curve in Asymptomatic Smokers and in Nonsmokers. Am. Rev. Resp. Dis., 110: 282, 1974. (reprint)

2. Hyatt, R. E. and Rodarte, J. R.: "Closing Volume," One Man's Noise -- Other Men's Experiments. Mayo Clin. Proc., 50:17, 1975 (reprint).

3. Hyatt et al.: Effect of Increased Static Lung Recoil on Bronchial Dimensions of Excised Lungs. J. Appl. Physiol., 39:429, 1975. (reprint).

4. Lai-Fook et al.: Quantification of Parenchymal-Bronchial Interdependence in Isolated Lobes. Amer. Rev. Resp. Dis., 113:215, 1976 (abstract).

5. Lai-Fook et al.: The Behavior of Artificially-Made Spherical and Cylindrical Holes in Lung Parenchyma. Fed. Proc., 35:231, 1976 (abstract).

6. Wilson, T. A.: Axial Bronchial Tension as a Determinant of Expiratory Flow (manuscript).

7. Lai-Fook et al.: The Elastic Constants of Inflated Lobes of Dog Lungs. J. Appl. Physiol. (accepted) (manuscript).

8. Lai-Fook, S. J.: Lung Parenchyma Described as a Prestressed Compressible Material (manuscript).