

# Monitoring the Lung:

## Mechanics and Volume

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ADVANCES IN ENGINEERING, electronic and computer technology during the past decade have improved the capability for intermittent and continuous monitoring of specific organ-system function. Monitoring of respiratory function, necessary because of the high incidence of complications, with significant mortality and morbidity, unfortunately has experienced slow advancement. There is general lack of sophistication in the area of respiration, in contrast to those methods routinely employed for the monitoring of other systems, especially cardiovascular function.

Pulmonary function can be assessed in several ways: 1) function as a gas-exchange organ, including measurement of alveolar and vascular function; 2) mechanical function of the lung, including lung volumes and mechanical properties of the diaphragm and chest wall; 3) metabolic or non-respiratory function. The object of this review is to focus on the development and availability of currently existing techniques used to measure the complex mechanical properties of the respiratory system. Special emphasis is placed on application and clinical utility of such techniques in management of acute pulmonary injury and respiratory insufficiency.

### Physiology and Measure of Pulmonary Mechanics

The study of respiratory mechanics is concerned with the physical process of gas transport in the respiratory system. This includes the action of the chest wall (rib cage and dia-

phragm) and the physical properties of gas flow within airways. Mechanical function of the lung is inherently related to other aspects of pulmonary physiology, including gas exchange and non-respiratory function. In acute and chronic respiratory disease, abnormalities in respiratory mechanics are important pathophysiologic features, both as primary mechanisms in creating disturbance in function and as being altered secondary to injury.

Measurement of mechanical function of the lung or the mechanics of breathing requires continuous recording of pressure, flow and volume during the respiratory cycle. Pressure-volume relationships are defined during airflow (dynamic conditions) and at times of zero flow (static conditions). Although a complete review of pulmonary mechanics is not warranted, certain definitions and principles must be considered. Specific pressures are defined as follows:

*Pleural pressure* ( $P_{PL}$ ) is the pressure on the pleural surface and cannot easily be measured directly in the patient. Instead, it is measured by means of intraesophageal pressure ( $P_{ES}$ ). Change in esophageal pressure has been found to correlate well with change in pleural pressure during the respiratory cycle.<sup>1,2</sup>

*Transpulmonary pressure* ( $P_L$ ) is the pressure difference across the lung, measured as the pressure difference between the airway opening ( $P_{A0}$ ) and the pleural surface ( $P_{PL}$ ). During inspiration and expiration this pressure gradient is produced by a number of factors, including resistance to airflow, pulmonary and chest-wall compliance, and tissue and gas inertia. At points of no flow, both resistance to airflow and inertial forces are absent; thus,  $P_L$  become a function of pulmonary compliance alone. Transpulmonary pressure is equal to the difference between mouth and esophageal pressures ( $P_L = P_{A0} - P_{ES}$ ) and by definition is a positive value.

*Transrespiratory pressure* ( $P_{RS}$ ) is the pres-

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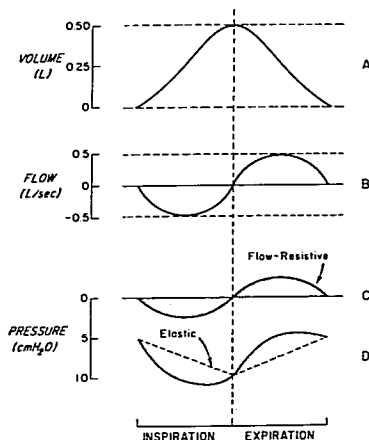


FIG. 1. Tidal volume, airway flow, and pleural pressure change during a single respiratory cycle. (Reproduced with permission from Laver MB, Austen WG: Lung function: Physiologic considerations applicable to surgery, Davis-Christopher Textbook of Surgery, Tenth edition, Edited by Sabiston DC Jr. Philadelphia, W.B. Saunders, 1972.)

sure difference across the respiratory system (the lung and chest wall) measured as pressure differential between airway and atmospheric pressures ( $P_{RS} = P_{AO} - P_{atm}$ ). It is important specifically in describing the pressure changes occurring during mechanical or positive-pressure ventilation. This pressure is not important during spontaneous respiration.

Changes in volume, airway flow, and transpulmonary pressure during a single respiratory cycle are shown in figure 1. Inspiratory and expiratory periods are separated at points of zero flow. These analog waveforms provide information from which a number of important values may be derived. Compliance (volume change per unit pressure change) may be calculated by determining the change in volume and corresponding change in pressure on the appropriate curves in this figure. A change in pressure is necessary to overcome two forces: elastic recoil and resistance to airflow. Pressure change necessary to overcome elastic recoil can be separated from pressure secondary

to flow resistance by selecting a point during the respiratory cycle when resistance to flow, or flow-resistive pressure, is zero. Airway flow is zero at end-inspiration, immediately prior to expiration, and at the conclusion of expiration. At the point of zero flow the relationship between lung volume and pressure change yields compliance. This is indicated by the straight line drawn between points of zero flow (dashed line in curve D), which indicates that the volume change per unit pressure change remains constant throughout the inspiratory and expiratory cycles. The pressure difference between the total pressure generated (solid line, curve D) and the pressure utilized to overcome elastic recoil (dashed line, curve D) represents the pressure needed to overcome flow resistance. The pressure (flow resistance pressure) has been isolated for the entire respiratory cycle and is shown as an individual curve (curve C). Evaluation of this curve reveals that resistance to flow is not constant during the respiratory cycle, but generally has the same shape as the flow curve generated at the mouth (curve B). There is need for greater pressure to overcome resistance to flow as the flow rate (0.5 l/sec in this case) reaches a maximum.

### Mechanical Work Done by Respiratory Muscles

In breathing, respiratory muscles work against three principal types of forces. *Elastic forces* are developed in tissues of the lung and chest wall when a volume change occurs. *Flow-resistive forces* occur due to resistance to flow of gas within airways and the resistance secondary to nonelastic deformation of tissue. *Inertial forces*, which depend on the mass of tissue and gases, are generally considered of minor importance and neglected in the calculations. During quiet breathing the respiratory muscles do active work only during the inspiratory phase; expiration is passive. Inspiratory work is divided into those factors that work against elastic forces and those that work against flow-resistive forces. Work during inspiration may be expressed as the product of pressure and volume:  $W\ddagger = P\ddagger dV$ .

‡ Units of work are gram-centimeter (g-cm) or kilogram-meter (kg-m) when pressure is cm H<sub>2</sub>O/cm<sup>2</sup> or g/cm<sup>2</sup> and volume is cm<sup>3</sup>.

WORK OF BREATHING ( $W_{EL} + W_{RES}$ )

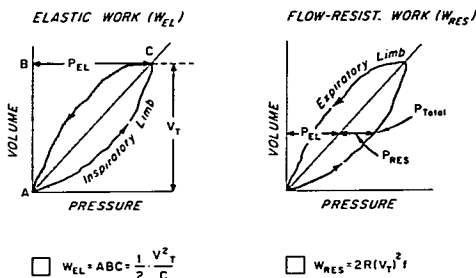


FIG. 2. Components of respiratory work; for explanation see text. (Reproduced with permission from Laver MB, Austen WC: Lung function: Physiologic considerations applicable to surgery. Davis-Christopher Textbook of Surgery, tenth edition, Edited by Sabiston DC Jr. Philadelphia, W.B. Saunders 1972.)

Several methods are available for expressing the relationship between work and respiration. Perhaps the easiest to comprehend is the pressure-volume plot, as shown in figure 2. Work of the respiratory muscles may be expressed as a relationship between pressure and volume changes during a respiratory (inspiratory and expiratory) cycle. The plot of pressure (transpulmonary) and volume (tidal) during inspiration and expiration produces a closed loop, with points A and C representing zero flow at end-expiration and end-inspiration, respectively. The straight line AC indicates the compliance (slope equals change of volume divided by change or pressure), or the elastic component of pressure change during inspiration ( $P_{el}$ ). The stippled area (ACBA) is the quantity of elastic work ( $W_{el}$ ) done during inspiration and is equal to half of the product of tidal volume and change in elastic pressure ( $P_{el}$ ). The elastic pressure change can be expressed in terms of compliance (C) and tidal volume ( $V_T$ ) as follows:

$$C = \frac{V_T}{P_{el}} \quad (1)$$

and

$$P_{el} = 1/C \times V_T \quad (2)$$

and since

$$W_{el} = \frac{1}{2} P_{el} \times V_T \quad (3)$$

then

$$W_{el} = \frac{1}{2} \times 1/C \times (V_T)^2 \quad (4)$$

Hence, elastic work per breath is inversely proportional to compliance and directly proportional to the square of the tidal volume.

The right-hand portion of the figure depicts total pressure applied during a tidal breath in inspiration, and the hatched area (shown between the elastic pressure line and total pressure) represents flow-resistive work ( $W_{res}$ ) performed during inspiration.  $W_{res}$  may be measured by planimetry of the stippled area shown to be right of the diagonal. It may also be calculated by using certain approximations. First, the rate of volume change must be known. The value of the latter may be approximated if  $V_T$  and the duration of inspiration are known. Let us assume a respiratory frequency ( $f$ ) and an equal distribution of time between inspiration and expiration, then:

$$\frac{60}{2f} = \text{duration of inspiration (sec)} \quad (5)$$

From Equation 5:

$$\text{Average rate of volume change} = \frac{2fV_T}{60} \text{ / sec}$$

$$\text{or } 2fV_T/\text{min} \quad (6)$$

The resistance (R) then:

$$R = \frac{P_{res}}{2fV_T} \quad (7)$$

Flow-resistive work ( $W_{res}$ ) may be ex-

pressed as the product of tidal volume and the average flow-resistive pressure:

$$W_{res} = 2R(V_T)^2 \times f \quad (8)$$

Thus, flow-resistive work during a single inspiration is proportional to flow resistance, respiratory rate, and the square of the tidal volume. The total respiratory work per unit time is equal to the sum of the elastic and the flow-resistive work per breath  $\times$  the respiratory rate. This may be expressed as follows:

$$\text{Total work per minute} = (W_{el} + W_{res}) \times f \quad (9)$$

or

$$\frac{1}{2} \times 1/C \times (V_T)^2 \times f + 2R(V_T)^2 f^2 \quad (10)$$

Work performed against resistance to flow is dissipated as heat during inspiration; elastic work (inspiration) represents potential energy at end-inspiration to be utilized to provide work to overcome flow-resistive forces during expiration—remembering that expiration is usually (but not always) a passive phenomenon using elastic lung and chest-wall recoil, e.g., inspiratory  $W_{el}$ .

Values for work of breathing and its metabolic correlate, the oxygen cost of breathing, are shown in table 1. Increased levels of exercise and a corresponding increase in minute ventilation lead to a disproportionate increase in respiratory work (e.g., Kg-m/unit of ventilation) at the higher levels of ventilation.<sup>2,4</sup> In addition, there is a parallel increase in oxygen consumption by the muscles of respiration—a tenfold increase from quiet breathing to heavy exercise. The elastic component, 66 per cent of

the total during quiet breathing, decreases to 39 per cent with heavy exercise, while resistive work increases from 33 to 61 per cent at high levels of ventilation. Thus, in normal lungs, increase in respiratory muscle activity is associated with a disproportionate increase in cost per unit ventilation (mechanical work and energy requirement) and a redistribution of the subdivisions of total work.

#### On-line Computation

Measurement and computation of the work of breathing are possible by recording simultaneous pressure and volume changes. During spontaneous breathing, transpulmonary pressure ( $P_{Ao} - P_{rD}$ ) is measured with simultaneous volume change, generally recorded by integration of airflow measured with a pneumotachograph. This method will measure elastic work of the lungs and thorax and flow-resistive work done on moving gas and lung tissue; flow-resistive work performed on the thorax (i.e., rib cage, diaphragm, and abdominal contents) is not measured.

During positive pressure of mechanical ventilation, total respiratory work is generally measured utilizing transrespiratory pressure ( $P_{Ao} - P_{atm}$ ) and simultaneous volume change, provided that the subject does not have an active respiratory (including abdominal) muscular contraction during the measurement. Use of esophageal pressure, i.e., transpulmonary pressure, allows one to define the properties of the lung alone, excluding the chest wall.

Applications of analog and digital computers during the past decade have produced systems capable of providing breath-by-breath calcula-

TABLE 1. Work and Oxygen Cost of Breathing

	External Work (kg-m/min)	Tidal Volume (ml)	Respiratory rate (breaths/min)	Minute Volume (l/min)	O <sub>2</sub> Consumption (ml/min)	Respiratory Work			
						(kg-m/min)	(% Elastic)	(% Resistive)	(O <sub>2</sub> cost ml/min)
Quiet breathing	0	500	15	7.5	300	0.3	66	33	3
Moderate exercise	620	1,600	23	37	1,500	5.2	57	43	52
Heavy exercise	1,660	2,400	48	115	3,500	35.2	39	61	352
Maximal voluntary ventilation	0	1,500	120	180	—	65	20	80+	—

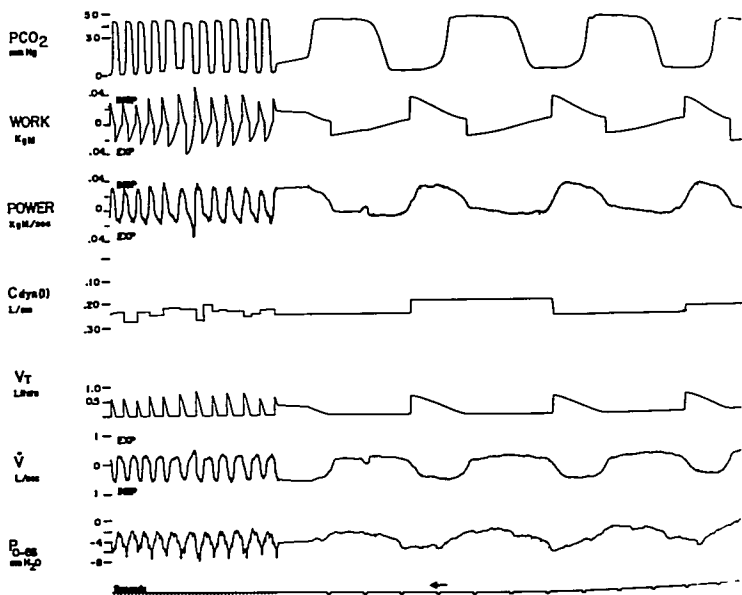


FIG. 3. On-line computation of pulmonary mechanics during spontaneous breathing. Tracing is continuous from right to left showing simultaneous recording of esophageal pressure, flow, volume, lung compliance, power, and work of breathing and airway  $P_{CO_2}$ . (Reproduced with permission from Fletcher G, Bellville JW: On-line computation of pulmonary compliance and work of breathing. *J Appl Physiol* 21:1321-1327, 1966.)

tion of mechanics of breathing without manual operations.<sup>3-7</sup> A typical configuration of the recorded output of such a system, as described by Fletcher and Bellville,<sup>8</sup> is shown in figure 3. Measured and derived values are indicated as follows:

**Volume ( $V_T$ ):** Tidal volume is obtained by integration of flow =  $(\int \dot{V} dt)$ .

**Compliance ( $C_{dyn}$ ):** Dynamic compliance is derived from volume change divided by change in pressure ( $P + p$ ) from points of zero flow at beginning and end-inspiration =  $(V/\Delta P_{in})$ .

**Power:** Expressed in kg-m/sec is the power developed in overcoming elastic forces or moving gas and lung tissue. It is obtained from the product of pressure and flow =  $(P_L \times \dot{V})$ .

**Work:** Expressed in kg-m, is obtained by integration of power =  $(\int P dt)$  over a specified period of time.

On-line computation creates a practical solution by providing for unlimited duration of study with direct recording of computed results in real time. As previously mentioned, the determination of both power and work of breathing by conventional methods requires additional use of planimetry to measure areas on the photographs of pressure-volume loops. This is not only tedious but introduces a number of sources of error and random variations into results. Examples of error include recorded artifacts, especially those produced with use of esophageal pressure monitoring. Swallowing and other changes in esophageal

muscular tone occur with slow frequency and extend over several respiratory cycles. In continuous on-line recordings such slow-frequency artifacts are easily detected and subsequently eliminated. Cardiogenic oscillations vary in amplitude and phase relationships with the respiratory oscillation. Such oscillations are easily filtered from the respiratory signal and the effect is greatly reduced by on-line computation. Other artifacts such as movement of the esophageal balloon during breathing, effect of weight of the mediastinal contents on esophageal pressure, and other extraneous signal noises are greatly minimized with on-line computation. Such systems provide analyses of multiple breaths (e.g., five to ten consecutive breaths) rather than basing information on a single-breath plot of the pressure-volume relationship. Indeed, studies have shown that analog computation reduces considerably the coefficient of variation in computed values of respiratory mechanics, e.g., lung compliance, compared with manual methods.<sup>5,9</sup> While no direct comparison of on-line work with conventional methods is available, it is evident that differences in variability of results with the two methods may be even greater than those in the computation of compliances. Studies of mechanics of breathing and respiratory physiology are facilitated by the use of on-line computation as a result of reduction in data analysis time, continuous rather than intermittent computation, and increased accuracy.

During the past decade there has been widespread interest in application and use of the digital computer in areas of data processing and storage. Several groups have described methods for monitoring physiologic variables with use of time-shared or dedicated digital computer systems. Efforts have been directed at their use in the critically ill, with emphasis on cardiovascular monitoring,<sup>10-12</sup> and more specific attempts have been made to focus on the respiratory system.<sup>14-16</sup> The applications include data-processing display and storage. Analog data are obtained as previously described, i.e., flow and pressure generated via a sensor-amplifier system. Analog signals, initially displayed on an oscilloscope for visualization, are transmitted via cables, telephone lines, etc., to analog-to-digital converters. Digitalized signals are then directed

to the computer, which provides both processing and data storage. Various methods of display of digital and graphic material, generally using a bedside cathode-ray tube and computer-generated read-out for permanent records, are then made available.

### Clinical Application

On-line measurement of pulmonary gas exchange and mechanics has been advocated to provide the clinical data needed for the diagnosis and treatment of pulmonary, cardiovascular, and metabolic abnormalities in the critically ill patient. These measurements, if they are to be useful, must be provided simply and automatically with maximum accuracy. The relevant question as to ultimate clinical usefulness is extremely difficult to resolve, since experience to date has focused on system design, technical development, and early applications. Further efforts are needed to demonstrate whether such systems are justified on a cost-effectiveness basis.

Thus far, the greatest interest has centered around the measurement of the work of breathing and its specific components with on-line methodology, while values such as respiratory rate, tidal volume, and forced expired volumes, FEV<sub>1</sub> and vital capacity, are easily obtained with simple inexpensive bedside techniques. Experience has shown that these variables are slow to change with time, thus making elaborate continuous monitoring techniques unnecessary. The work of breathing, however, has been of interest to many investigators, since it is directly related to oxygen cost and hence, efficiency of the lung, under normal conditions and in the presence of pulmonary and cardiovascular disease.<sup>7,17-19</sup>

Several attempts have been made to measure the work of breathing following trauma, especially non-thoracic trauma, and immediately after major cardiovascular and thoracic operations. In the late 1960's, Peters and colleagues<sup>20</sup> used an analog technique to measure changes in both work of breathing and chest-wall compliance following thoracotomy. The study was stimulated by a previous observation that the breathing pattern after thoracotomy was markedly irregular, even in the absence of pulmonary disease, a variation in the breath-by-breath pattern that had not been taken into account during previous analyses.

Pulmonary compliance (dynamic) and respiratory work were measured in the post-operative period on the first, third and seventh days and approximately two weeks after operation, and results were compared with control values obtained in 15 normal subjects. Results of this study were divided into groups depending on the operative approach and operative procedure. Group I consisted of patients treated by lateral and anterolateral thoracotomy without pulmonary resection; Group II, lateral thoracotomy with excision of lung tissue; and Group III, midline sternotomy without resection. Postoperative values for compliance and work showed considerable variation within each group and from group to group. Post-operative decreases in total compliance (lung and chest-wall) were found in all groups and were similar to values reported by other investigators. The numbers of patients and determinations were too small to define any pattern of compliance change in response to location of surgical incision. The work of breathing was generally increased in the immediate postoperative period (day after operation) and remained increased during the first postoperative week; again, significant intra- and intergroup variations were apparent. Respiratory work was expressed as work per minute and work per liter ventilation; the latter, a measure of efficiency of the lung and an expression of energy requirement, was more markedly elevated above control levels than work per minute.

Proctor *et al.*<sup>21</sup> measured mechanical function of the lung following direct traumatic pulmonary injury in Vietnam combat casualties. Total and resistive work of breathing were increased from the time of hospital admission until measurements were terminated on the tenth post-injury day.

An analysis of mechanical function following thoracoabdominal operations by on-line digital computer has been reported by Lewis.<sup>22</sup> Variability in technique, patient population, and the limited number of determinations limit the usefulness of the specific values.

There has been great interest in application of on-line techniques in patients with acute respiratory insufficiency and related multi-system disorders. Specific applications of such systems toward improved clinical management have been described, and justification

for such elaborate systems has included the improved feedback relating to patient-machine interaction during mechanical ventilation, and the early detection of mechanical problems such as system leaks, "out-of-phase," and airway and tube obstruction. In addition, such a system provides a continuum with additional information about respiratory function in real time, including advance warning of developing pathologic processes that otherwise would not be detected due to the lack of repetitive, serial data collection and analysis. The third and perhaps the most useful bit of information is the availability of dynamic measurements of respiratory function, especially indices of respiratory work. The relationship between pulmonary mechanics and circulatory function and its alteration by changes in respiratory pattern is shown in figure 4. With this information it is possible to improve indications for ventilator support and increase efficiency for prediction of weaning ability.

In a recent review, Saklad *et al.*<sup>16</sup> listed the following advantages of clinical on-line monitoring: 1) anticipate oncoming ventilatory failure; 2) recognize ventilatory inadequacy and the responsible mechanisms; 3) establish the need for mechanical ventilatory support; 4) improve management of ventilatory performance; 5) review the past history of ventilatory function during therapy and project needs and trends; 6) establish alarm systems to inform attendants when certain variables have exceeded established limits.

These authors have made use of a cathode-ray tube located at the bedside, and the analog display includes flow, tidal volume, and airway pressure. When the interrelationships between the simultaneous traces are observed on the cathode-ray tube, changes in both pulmonary and mechanical function may be diagnosed with relative ease. The examples cited include early detection of changes in ventilator characteristics, e.g., volume delivery, with associated changes in pulmonary compliance and resistance, especially in the case of pressure-pre-set ventilators. Other aspects of ventilatory management and the ventilator setting that become apparent with visual display of the analog signals include prolongation of inspiratory times secondary to a low inspiratory flow, the relationship of inspiratory to

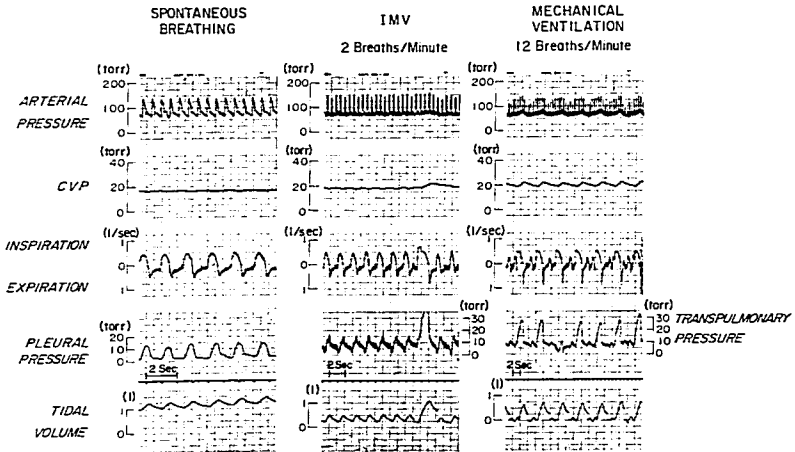


FIG. 4. Arterial pressure, central venous pressure, flow, pleural or transpulmonary pressure and tidal volume shown during spontaneous breathing, intermittent mandatory ventilation, and mechanical ventilation, measured in one patient over a 30-minute steady-state period. A, during spontaneous breathing ( $V_T = 250$  ml); pleural pressure  $\Delta = 10$ –12 torr; note stability of BP and CVP. B, IMV, 2 breaths/min ( $V_T$  spont = 250 ml,  $V_T$  IMV = 750 ml); transpulmonary pressure is shown to change by 15 torr during a spontaneous breath and by 30 torr with a ventilator-assisted breath. CVP is lowest at peak inspiration (spontaneous) and highest at peak inspiratory pressure of ventilator-assisted breathing. Arterial pressure decreases 25 torr following a ventilator-assisted breath. C, mechanical ventilation at 12 breaths/min; note intermittent spontaneous breathing. There is marked variation in CVP and arterial pressure compared with previous patterns.

expiratory time, and the effects of an end-inspiratory pause as well as an end-expiratory pressure. The appearance of spontaneous respiratory activity is easier to detect during the course of controlled ventilation when analog signals are available.

With utilization of the digital computer (as in fig. 5), it is possible to convert flow and pressure data into a number of derived variables. The information may be projected as an alpha numeric display on a remote or a bedside cathode-ray tube and may be provided in print-out form with a teletypewriter. When available, such data allow for bedside "titration" of ventilator performance against the mechanical characteristics of the lung-chest-wall system. In addition, data may be stored for future use to enable analysis of serial, e.g., daily or hourly, values in order

to appreciate more fully changes in patient's respiratory status. Osborn and colleagues recently evaluated a computer-based monitoring system in use in the Cardio-Pulmonary Unit of Pacific Medical Center for the past eight years.<sup>23</sup> Although the system is designed for both cardiac and respiratory monitoring, the latter, which includes measurement of flow, pressure and gas composition, provides a major component of the output. From these data, oxygen uptake, carbon dioxide output, respiratory quotient and breath-to-breath measurements of respiratory mechanics are determined. Derived variables of lung mechanics include compliance, nonelastic resistance, ventilation volumes, and total work of breathing during positive-pressure ventilation. According to the authors, advantages of the system include: detection of respirator and



mechanical malfunction, improved facility to adjust the ventilator settings appropriately, and early detection of alterations in pulmonary function such as hemothorax, pneumothorax, and need for pulmonary toilet.

Turney and colleagues<sup>21-26</sup> have devised an on-line respiratory waveform and pulmonary gas-exchange system utilizing both a computer and a respiratory-gas mass spectrometer. The utility of the system has been limited to cases where continuous monitoring has been desirable to provide information about mechanical problems, *e.g.*, airway occlusion, which, if detected early, allow for intervention before the physiologic consequences have become serious.<sup>24-26</sup>

Although these systems have incorporated on-line gas analysis (inspiratory and expiratory oxygen and carbon dioxide), their derived benefits are difficult to substantiate. Complexity and cost of the system are increased substantially once the need for sampling lines, gas analyzers, and calibrating techniques has been established. When gas analysis is combined with volume measurement, derived values include oxygen uptake, end-expiratory carbon dioxide, and carbon dioxide production. Several authors have expressed hope that

with such non-invasive methods of gas analysis it would be possible to reduce or eliminate the need for standard serial arterial blood-gas analysis. Such on-line gas analysis does provide a basis for alarm systems (*e.g.*, inspired and expired oxygen fraction) and has in theory and practice detected malfunction of airway and mechanical apparatus.<sup>16,21</sup> According to Hilberman *et al.*,<sup>23</sup> availability of airway gas analysis did not reduce either the number of days arterial cannulas were in place or the number of arterial blood-gas determinations.

Perhaps the most direct clinical utilization of such a monitoring system has been that described by Peters *et al.* in an attempt to delineate objective indicators for respirator therapy in both post-trauma and postoperative patients.<sup>27</sup> Using a simple bedside cart linked to a remote digital computer, respiratory measurements were determined rapidly during mechanical ventilation in order to assess the effectiveness of therapy and the feasibility of discontinuing respirator support. Retrospective analysis of the data indicated that compliance and airway resistance did not discriminate between patients who could and those who could not be weaned from respirator

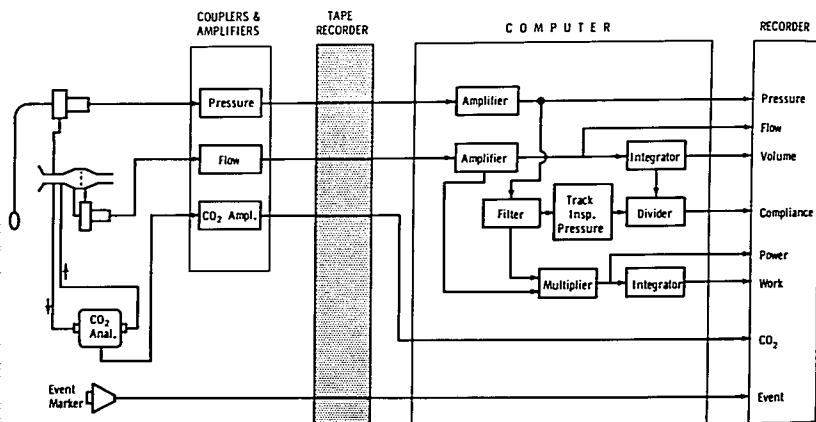


FIG. 5. Block diagram of digital computer system for data collection, storage, processing and output. (Reproduced with permission from Fletcher G, Bellville JW: On-line computation of pulmonary compliance and work of breathing. *J Appl Physiol* 21:1321-1327, 1966.)

support. However, calculation of respiratory work, as work per liter or work per minute, appeared to be a reliable method for discriminating between those in need of continued ventilator support and those capable of being weaned. For example, total work per liter was 0.05 kg-m for patients who did not need the respirator; the corresponding value for those who needed the respirator was 0.3 kg-m. Total work per minute was also a good discriminator, with the division between the two groups at a value of 1.0 kg-m per minute, although the overlap was greater than that observed with the work per liter determination. Proctor and Woolson<sup>28</sup> attempted to predict respiratory muscle fatigue by measuring work of breathing (lung) in postoperative patients and found means of 0.83 kg-m/min in patients who did not require ventilatory support and 3.5 kg-m in those who did. Furthermore, at a calculated work of breathing of 1.34 kg-m, the fewest errors were made (13.8 per cent false-negative; 13.8 per cent false-positive) in decisions whether to use respirator therapy. In conjunction with standard bedside pulmonary function analysis (e.g., vital capacity,  $V_T/V_T$ ,  $\dot{Q}_T/\dot{Q}_T$ , etc.), Peters *et al.*<sup>27</sup> were able to define a total work per minute greater than 1.8 kg-m, or a work per liter greater than 0.18 kg-m, as the critical level beyond which ventilator support was needed. Whether such methods are either clinically beneficial on a general basis, i.e., outside of specific research-oriented intensive care units, is unsettled pending further evaluation.

Specific aspects requiring further clarification include the utility and ease of operation in clinical units without elaborate engineering and computer support services, especially if application under these circumstances will reduce accuracy and reproducibility of data. Overall cost effectiveness, a problem of growing importance in intensive care medicine, bears further analysis.

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