LUNG AND CHEST WALL DISTORTABILITY

IN DOGS

LUNG AND CHEST WALL DISTORTABILITY IN DOGS

by

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ABSTRACT

In order to assess the relative distortability of the lungs and chest wall, we cannulated one lung in intact anesthetized dogs and starting at different lung volumes (V_L), inflated one lobe with known volume increments ($+\Delta Vx$) using positive pressure, leaving the other lobes free to change volume ($\pm \Delta Vy$), i.e. inflate or deflate. If the lung were very distortable (liquid-like) ΔVy would equal $-\Delta Vx$. If the chest wall were very distortable, ΔVy would equal zero. If neither were distortable, ΔVy would be very much greater than zero. At low V_L, ΔVy max was -12.5% of ΔVx max. At high V_L, ΔVy max was zero. Measured pleural surface pressure differences between lobes were large. Results indicate that: 1) lung distortability decreases as volume increases; 2) the pressure cost of chest wall deformations is substantial and results in marked regional differences in pleural surface pressures.

RESUME

Dans le but d'évaluer la distorsion relative entre le poumon et la cage thoracique, chez des chiens anesthésiés, après canulation d'un poumon, commençant à des niveaux pulmonaires (V_L) variables un lobe est gonflée à différents volumes ($+\Delta Vx$) alors que les autres peuvent varier de façon passive ($\pm \Delta Vy$), i.e., se gonfler ou se dégonfler. Si les poumons sont très déformables (comme un liquide) ΔVy sera égal à $-\Delta Vx$. D'autre part, si la cage thoracique est très déformable, ΔVy sera égal à zero. Enfin si aucun n'est déformable, ΔVy sera plus grand que zero. A bas volume, ΔVy max est -12.5% de ΔVx max alors qu'à haut volume, ΔVy max est égal à zero. Les différences de pression pleurales de surface entre les lobes étaient grandes. Les résultats indiquent que: la déformation pulmonaire est inversement proportionnelle au volume. Le coût en pression de la déformation de la cage thoracique est substantiel et se traduit pour la pression pleurale en des différences régionales marquées.

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Brian G. Murphy

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Chapter I

INTRODUCTION

Statement of the Problem

The mechanical events through which forces are applied to the lung are basic to respiratory physiology because they produce ventilation and determine in part its distribution within the lung and the work of breathing. In considering ventilation distribution, a parameter of utmost interst is the local transpulmonary pressure (P_L). Since pressure at the airway opening (P_{ao}) is easily measured, local pleural pressure (P_{pl}) over the whole surface of the lung becomes the unknown parameter. A number of techniques both direct and indirect have explored the topography of pleural pressure but none have yet been universally accepted. A brief review of the various techniques used will be presented, followed by some theories of pleural space mechanics. Since the study of mechanics ultimately leads to some form of stress analysis, experiments are described which attempt to qualitatively examine the distortability of the lung relative to the chest wall. The distortability of these two structures will have an important influence on local pleural pressure and ventilation distribution.

Early Measurements of Pleural Pressure

Carson (8) in 1820 was the first to measure the elastic recoil of the lung. This was done on a cadaver by connecting the trachea to a water manometer and observing the positive pressure developed when the thorax was opened to the atmosphere. Von Neergaard and Wirz (39) obtained measurements of pleural pressure on man in vivo by puncturing an intercostal space and using a pleural cannula to measure the pressure in a relatively large pneumothorax.

The size of the pneumothorax was increased from 100 to 1100 ml to yield data at different lung volumes. Christie and McIntosh (9) reviewed the work of these and other pioneering investigators, discussed their relationship to pulmonary elastic recoil and gave evidence that, with the technique of measuring pressure in a large pneumothorax variation of several centimeters of water were to be expected. They used a pleural cannula, 100 cm of tubing and several water manometers in series (which reduces the fluctuation, thereby improving the frequency response) to measure pressures in a small pneumothorax (40 ml) and standardized the whole technique as to position of injection, posture, recording of tidal air and lung volume. Since it was realized that even pneumothoraces of 40 ml caused a change of the pleural pressure, Farhi et al (18) sampled pleural pressure with a needle surrounded by a seal so that only 0.2 ml of air was introduced. They found that with the dog in the supine position, the pleural pressure was fairly uniform at all points over the lung surface except at the extreme bases on the dorsal surface and at the apex. However they found abrupt changes even after death which they could not explain and the technique has been severely criticized by Agostoni (1) in that it is probably subject to surface tension artifacts.

Recent Measurements of Pleural Pressure

A review of the techniques being used today show that many difficulties are yet to be overcome. First it is important to differentiate between pleural liquid pressure and pleural surface pressure. The latter is defined (1) as the net force per unit area perpendicular to the pleural surface and is the pressure that determines the degree of expansion of the lung under normal conditions. Inside the pleural space there is a continual net transfer of liquid from the "space" to the vascular bed which reduces the volume of pleural liquid

until contact between the lung and chest wall prevent a further reduction (44). If contact were not to occur, the pleural liquid pressure (which is simply the hydrostatic pressure of the pleural liquid relative to atmospheric pressure, P_B) would equal pleural surface pressure. Once contact begins to occur, net removal of more liquid lowers pleural liquid pressure below surface pressure until an equilibrium is reached between the forces removing liquid from the pleural space and the stresses between the surfaces in contact. The situation is very analogous to a large jar representing the chest wall which contains an inflated balloon representing the lung. The balloon is surrounded by liquid. If a syringe is used to suck out all the liquid through a side port in the jar, the balloon will expand until it comes into contact with the jar. Further suction lowers the liquid pressure between the balloon and the jar but since the balloon is unable to expand further, its transmural pressure must remain constant. Since the inside of the balloon is open to atmospheric pressure, the surface pressure on the outside of the balloon also remains constant. Hence, liquid pressure can be greatly different from surface pressure (40).

The techniques used to measure pleural surface pressure may be listed under the following headings:

INDIRECT

1) esophageal pressure
2) regional lung volume by ^{133}Xe
3) alveolar size by morphometry
4) weight-area hypothesis
5) lung density

DIRECT

6) intrapleural pressure-sensing flat capsule } the same principle

7) intrapleural pressure-sensing balloon

8) measurements in a small pneumothorax

9) counterpressure technique

Esophageal pressure. The measurement of esophageal pressure with adequate balloons has provided useful data about changes of pleural surface pressure and to some extent about its absolute value. It has also allowed the determination in living man of the static and dynamic volume-pressure relationship of the lung and of the chest wall. Hence this method has greatly contributed to the progress of the mechanics of breathing. However its contribution to the knowledge of the topography of pleural pressure is small because only a limited region of the pleural surface may be scanned from the esophagus and even data obtained with adequate techniques must be taken with reservation (1).

Regional lung volume by ^{133}Xe. Milic-Emili and his associates (7, 27, 36, 37) measured regional alveolar expansion by dilution of ^{133}Xe during inspiration and from measurements of esophageal pressure they calculated the transpulmonary pressure as a function of lung height in seated man over all the vital capacity range. A necessary assumption is that mean alveolar expansion of any horizontal slice of lung seen by a counter was representative of alveolar expansion at the pleural surface at the same level. That is, it assumes that the parenchyma at any one level is homogeneous. This may not be correct. Homogeneous may be defined as having uniform elastic properties and distending pressure throughout. Hence, neglecting differences in upper and lower lobe pressure-volume characteristics, an excised lung inflated with

positive pressure is homogeneous (28), whereas a lung in situ is not because there is a gradient of alveolar expansion (21, 36). That is, the lung in situ is deformed, but because the lung consists of lobes which may slide over each other, the degree of alveolar deformation is not as severe as it might otherwise be. In other words, the lung has some ability to suffer deformation and remain homogeneous. The sliding of lobes presumably helps to keep the parenchyma as homogeneous as possible during the lung and chest wall deformation that occurs with changes in posture (26). However, the lung undergoes smooth indentations at each interspace along the costal wall (J.C. Hogg, personal communication) meaning that it is deformed along its surface. Any subsequent inhomogeneity may not be very deep (12, 42) and therefore alveolar expansion close to the pleural surface may be different from mean alveolar expansion of a horizontal slice.

To determine whether or not alveoli at any one level are deformed because of shape differences between lung and chest wall, D'Angelo (10) carried out morphometric studies at the pleural surface on rabbit lungs in situ and on frozen sections taken from a core sample at the same site. He found that in most cases, the volume to surface ratio (V/S) of alveoli at the pleural surface was the same as alveoli deeper in the lung and that the geometry of alveolar expansion in situ was similar to that in the isolated lung. Their data provided a direct demonstration that under most conditions there is a unique relationship between alveolar size and transpulmonary pressure, despite the marked deformation that occurs in fitting the lung into the thorax. The lung for the most part therefore appears to adapt its shape to that of the container without undergoing local alveolar deformation at the pleural surface, "as if the alveoli could slide a little on each other". However, an important

exception to this was found. In the upper lung (2nd intercostal space) of head-up rabbits at FRC and in the upper lung of eviscerated supine rabbits with tungsten beads in the airways to increase total lung weight, alveolar deformation at the pleural surface was present. Although the precise mechanism is unknown, it must be caused by shape adaptation in the thorax since deformation was never found in isolated lobes. It is very significant that the deformation was found under conditions most akin to man (head-up and higher total lung weight). In rabbits it was limited to the upper 20% of the lung height. This evidence (yet to be confirmed in man) supports the previous criticism of the ^{133}Xe technique to determine pleural pressure. For the other 80% of the lung height in head-up rabbits and for supine rabbits, the data for sub-pleural and inner alveoli fit the relationship between (V/S) and transpulmonary pressure in the isolated lung. This is evidence in favor of the reliability of in situ measurements of local pleural pressure or its determination by regional lung expansion using ^{133}Xe or morphometry. However it would be wrong to use evidence from experiments on small animals to support the technique used on man in an effort to resolve issues of species differences. Such issues will be dealt with in the section "The Pleural Pressure Gradient". In summary, the results of D'Angelo's experiment are somewhat two-sided in that they reveal sub-pleural homogeneity under most conditions but inhomogeneity at the apex in the head-up posture.

Robertson et al (42) and D'Angelo & Michelini (12) both used morphometry on excised lungs to evaluate the deformation and alveolar inhomogeneity secondary to a large local stress on a small area of the pleural surface. The finding that the depth of inhomogeneity was no greater than 1 - 2 mm suggests that the force contribution of pleural pressure from one relatively small

area of the lung may be too small, in comparison to the pressure summated over the remaining area, to affect regional volume. Furthermore, in a lung placed on a table Katsura et al (28) found a very small depth of inhomogeneity in the region of contact. More morphometric data for human lungs in situ may provide a more solid basis for indirect techniques of pleural pressure measurement using ^{133}Xe or morphometry.

Alveolar size by morphometry. The topography of pleural pressure can be calculated from measurements of alveolar size in situ using the relationship between alveolar size and transpulmonary pressure of isolated lobes. Glazier et al (21) fixed dog lungs in situ by freezing and measured alveolar size by histologic morphometric techniques. The technique is subject to the same considerations (supportive as well as critical) as the Xenon technique previously discussed. The tests used to detect inhomogeneity were done on samples taken 5 cm and 24 cm below the apex but not near the visceral pleura where deformation takes place. Future experiments that closely examine the alveolar expansion at the visceral pleura could help resolve this issue.

Lung density and the weight-area hypothesis. To avoid using the tedious method of morphometry to measure regional lung volume and the implicit assumption concerning alveolar shape, Hogg and Nepszy (24) measured the distribution of lung density in intact dogs which should be inversely proportional to the degree of expansion expressed as per cent of TLC. For calibration, the procedure of sawing out blocks of lung tissue was repeated on excised lungs frozen at various known distending pressures. The volume of air per unit lung weight was determined and a pressure volume curve constructed which agreed well and so was combined with data from Frank (20) and Faridy et al (19).

Then the volume of air per unit lung weight at various heights in situ was determined so the corresponding pleural pressures could be calculated. Again this method overlooks possible alveolar deformation at the apex in situ. However, the results stimulated Hogg and Nepszy to compare their calculated pleural pressure gradient to that obtained by the weight-area hypothesis. This hypothesis was used by Glazier et al (21) to predict the pleural pressure at any level by dividing the cross-sectional area, A, into the weight, W, of the lung below it. Pressures obtained by the two methods widely disagreed. The discrepancy upheld the hypothesis that the lung partly supports the abdominal contents. This in effect would render inaccurate any oversimplified hypothesis used to predict pleural pressure.

Flat capsule, balloon and small pneumothorax. Hoppin et al (26) made direct measurements of pleural surface pressure using a modification of the intrapleural balloon. An assembly of flat balloon capsules that minimized local deformation of the lung were located at various points in the craniocaudal direction between the lung and costal wall of nine dogs. At end expiration, transpulmonary pressure was several centimeters of water less beneath the ribs than beneath the interspaces. A residual local variation was found in every experiment. This data underlines the difficulty in obtaining repeatable representative measurements of pleural pressure regardless of the technique. Ideally, in the process of measuring the pleural surface pressure, the existing equilibrium of static forces should not be changed. For example the presence of a pneumothorax is a departure from this ideal. An evaluation of various methods using a model and a discussion of the potential principles have been made by McMahon and Bromberger-Barnea (32). In their model, pressures recorded from needles, cannulas and fluid-filled catheters could be

completely independent of surface pressure. Pressures recorded by balloons such as have been used by Turner (46) and Kreuger et al (30) varied, though not directly with surface pressure. In contrast the flat capsule device was found to measure surface pressure accurately. At the very edge of any object in the pleural space, no matter how thin, the pleura must separate. The resulting deformation of lung may have quite small radii of curvature, resulting in large local pressure variations. The design of the device removes the deformation at the periphery to a distance from the pressure-sensing portion, by surrounding the sensor with a wide skirt. Error would also result from inward displacement of the lung, the equivalent of a reduction of transpulmonary pressure by a reduction of lung volume. To minimize this error, the device was as thin as possible. Rather than using an assembly of pressure sensors all on one skirt, Zidulka (personal communication) used single capsules surrounded by a plastic dental-dam skirt about 5 cm in diameter. The devices could be moved about in situ by attached silk ligature. Hard plastic backing was optional.

Counterpressure technique. Agostoni and his co-workers have carried out the most systematic direct measurements of pleural pressure on rabbits and dogs (1). Although his counterpressure technique is technically difficult and has not been repeated by any other group to date, his data show remarkable repeatability between animals and between observers and has the advantage that it does not invade the pleural space. Stated briefly, the technique consists of clearing endothoracic fascia and making a tiny incision in the parietal pleura of an apneic animal without producing a pneumothorax. This is possible because surface tension at the edge of the incision prevents the lung and chest wall from moving apart. A small area of the lung becomes exposed to atmos-

pheric pressure and recoils inward. By sealing a capsule to the region surrounding the incision and lowering the pressure of air within it, the surface of the lung was brought back to the normal position. The pressure required for this should be equal to the pleural surface pressure of that region. The limits of this method are: 1) it can provide only static measurements; 2) it can be applied only to the intercostal region; and 3) it can be applied only to apneic animals at or above the resting volume of the respiratory system, because breathing movements or deflation break the air-liquid miniscus at the rim of the incision. The third limit was overcome by applying the capsule to a small region of exposed intact parietal membrane. The surgery is more difficult but the rest of the procedure is the same. The tension of the parietal membrane causes an artifact which is only appreciable if the radius of curvature is small as in very small animals. The second limit means that if pleural pressure under the ribs is different from that under the interspaces then the counterpressure technique gives a biased result of mean pleural pressure because all the measurements are made in the interspaces. Evidence that the pressures are different have been supplied by Hoppin et al (26) who found costal variations greater than 5 cm H_2O using their flat balloon capsules. In addition the results contained herein show large differences in pleural pressure swings between upper and lower lobes under conditions of lung deformation, thus demonstrating pressure discontinuity. Therefore pleural pressure is almost certainly different under ribs than under the interspaces.

The Pleural Pressure Gradient

Several studies have shown that in both man and dog there is a vertical gradient in pleural pressure with more negative values at the lung top (18, 30, 41). The nature of the pleural pressure gradient has been well

reviewed by Agostoni (1) from which particular highlights will be presented included with important work published in this area since 1970.

Rohrer (43), Duomarco et al (17), Mead (33) and Proctor et al (41) felt that the different shape of the lung and of the chest wall, besides lung weight, could cause local differences in pleural pressure. Despite other theories that have intervened, this concept is still important. Kreuger et al (30) measured pleural pressure in head-up dogs using a pleural balloon and suggested that the lung behaves as an homogeneous fluid of same mean density which develops a hydraulic gradient of pleural pressure. However, Turner et al (46) found that the vertical gradient of transpulmonary pressure did not decrease with decreasing lung density, achieved by increasing lung volume. His results therefore contradict the 'liquid-lung' hypothesis.

In man, invasive techniques cannot frequently be used. Instead, Milic-Emili and his associates (7, 27, 36, 37) measured regional alveolar expansion with ^{133}Xe and showed that the regional volume of the ventilating units as a percentage of their volume at TLC (V_R/TLC_R) was greater at the top than at the bottom. This was confirmed by Glazier et al (21) and Hogg and Nepszy (24) on dog lungs frozen in situ and by D'Angelo (10) on rabbit lungs in situ. The fact that this gradient of expansion exists in the species studied seems indisputable but the difficulty has been to correlate the gradient of alveolar expansion with the gradient of pleural pressure. According to the weight-area hypothesis described by Glazier et al (21) if the lung hangs from the chest wall and the support at the hilum is small, pleural surface pressure at a given height should be given by the weight of the lung below that height divided by the cross-sectional area at that height. The hypothesis is inconsistent (1) because the lung is assumed to be supported by a single zone

which is contradicted when the next zone is considered. Furthermore, in an isolated lung suspended from the apex, regional lung expansion is uniform and pleural surface pressure is nil proving that expansion of the lung is not significantly affected by its own weight (28).

West and Matthews (47) refined the weight-area hypothesis and without accounting for Katsura's observation (28) wrote, "it is probable that both the topographical differences in pleural pressure and expansion of the lung are part of the same phenomenon: deformation of the elastic lung by its own weight." We are unable to understand or evaluate the paper. It is impossible to repeat the finite element analysis they did for lack of information concerning the computer algorithms used. Nor was the model tested under conditions for which the answer is known (as every model should be so tested).

Duomarco and Rimini (17) suggested that the lung supports part of the weight of the abdominal contents. Hogg and Nepszy (24) were led to the same conclusion by qualitative analysis of their data and Agostoni (3), has shown it by eliminating the effect of the abdomen on the rib cage and diaphragm of head-up rabbits and dogs by evisceration. Lung FRC dropped 26% (rabbits) and 53% (dogs) of normal FRC. Greene et al (23) provided further evidence in man. They seated four human subjects in water up to the xiphoid to eliminate the weight of the abdomen on the chest wall. Lung FRC fell by 10.4% VC, and regional lung volumes using ^{133}Xe at FRC were less at all lung heights than at FRC in air. However, regional lung volumes were proportionately the same when compared to control lung volume in air. These results and those of Agostoni (3) show that the lung is inflated in part by the weight of the abdominal contents acting principally through the diaphragm.

According to Agostoni (2), the distribution of pleural surface pressure

is essentially related to the regional expansion that the lung undergoes in fitting the chest wall whose shape in turn is mainly determined by the action of gravity on its parts, particularly the abdomen-diaphragm. In extensive studies on various experimental animals (1, 2, 3, 4, 11, 13) Agostoni and his co-workers have repeatedly proved this point and concluded the following: 1) In all postures at FRC, pleural surface pressure becomes progressively more negative from the bottom to the top of the lung and is nearly the same at a given level. (Top and bottom are defined only with respect to the gravitational field.) 2) In a given posture and species, there is a unique relationship between the pleural surface pressure and per cent of lung height. The vertical pressure gradient decreases as the animal size increases both within and among species. 3) The pressure gradient is not mainly related to the lung weight as previously maintained (30, 7, 36, 21), but to the action of gravity on the chest wall. In eviscerated animals, the pleural pressure gradient was reduced 2 - 3 times. Further to this they observed that the pressure-gradient decreased and eventually disappeared when the respiratory system was expanded by increasing alveolar pressure. Also, a cranio-caudal gradient was obtained in the supine posture, by lowering the abdominal pressure with open airways.

In experiments designed to separate the additive effects of the abdomen (evisceration), the diaphragm (diaphragmectomy) and lung weight (pulmonary exsanguination) on the pleural pressure gradient they found that in prone and suspended rabbits, about 40% of the vertical gradient of transpulmonary pressure should be due to the abdomen, 35 - 40% to the effect of gravity on a hypothetical diaphragm-rib-cage interaction and 15 - 20% to the lung weight and the gravity dependent distribution of pulmonary blood. In head-up rabbits at FRC the contribution of the abdomen to the difference of transpulmonary

pressure between top and bottom should be about 60%, that of the diaphragm-rib-cage interaction about 25% and that of the lung weight less than 15%. In this posture a little could also be contributed by a gravity dependent shape effect on the rib-cage. The above partitioning of the factors contributing to the vertical pressure gradient could be different in man.

Since the gradient of pleural pressure under normal conditions seems to be mainly determined by the effect of gravity on the chest wall, it should be possible to produce marked differences of pleural pressure by applying uneven stresses to the chest wall. Agostoni and D'Angelo (2) decreased the pressure over the caudal part of the abdomen of supine rabbits and dogs at FRC and found that the pressure in the cranial region decreased more than in the caudal one at the same height, i.e., a cranio-caudal gradient was produced. These results showed that, 1) under special conditions pleural surface pressure in the costal region may vary markedly at a given height and given posture, 2) lung and chest wall deformation may change the distribution of pleural surface pressure.

In direct contrast to Agostoni's results on eviscerated animals (3), Greene et al (23) concluded that the pleural pressure gradient in man was unaltered when the effect of the weight of the abdomen was eliminated (the distribution of regional alveolar expansion remained the same). This is an important paradox. Either 1) there is a substantial difference in the nature of the pleural pressure gradient between small animals and man (22), or 2) the indirect ^{133}Xe technique or the counterpressure technique are not measuring mean pleural surface pressure at a given height. (See pages 4 and 9 for discussion of the techniques).

Part of this confusion in correlating pleural pressures with regional

alveolar expansion surely must be due to the simplifying assumption that pleural pressure at only one height of the lung is responsible for expansion of the lung at that height. A quick look at the geometry of the lung would show that this cannot be so; the expansion of any region must be affected by pleural surface pressures at many heights that are equidistant from that region (Fig. 1A). The assumption of homogeneity at one apical level may itself be invalid in man, as it is in small animals (10), which further pushes one to the notion that there may be a dissociation between local pleural pressure and regional lung inflation and ventilation. This dissociation would be due as well to intercostal and subcostal variations in pleural pressure (26) and to the probability that local deformations do not extend very deeply into the parenchyma (12, 42). More knowledge is needed on how all the pleural pressures around the lung integrate by transmission through the lung to expand it. This requires that the topography of pleural pressure be studied in still more detail and that it be used with interdependence theory to predict the expansion of any lung region.

Forces Expanding Airspaces Deep to the Pleural Surface

Whereas it is obvious that alveoli at the pleural surface of a lung in situ are exposed to pleural pressure, what is the pressure expanding airspaces deep within the lung? Respiratory physiologists have generally assumed it to be the pleural pressure as well, but the mechanism is perhaps not so obvious. The problem has now been illuminated and to a large extent solved by Mead, Takashima and Leith (34) who made a good well-defined stress analysis of a mathematical model of the lungs. They first consider a lung which has 1) homogeneous parenchyma, 2) uniform pleural pressure, P_{pl} and 3) uniform alveolar pressure, P_{alv}. Since statically, gas pressure in all interconnected

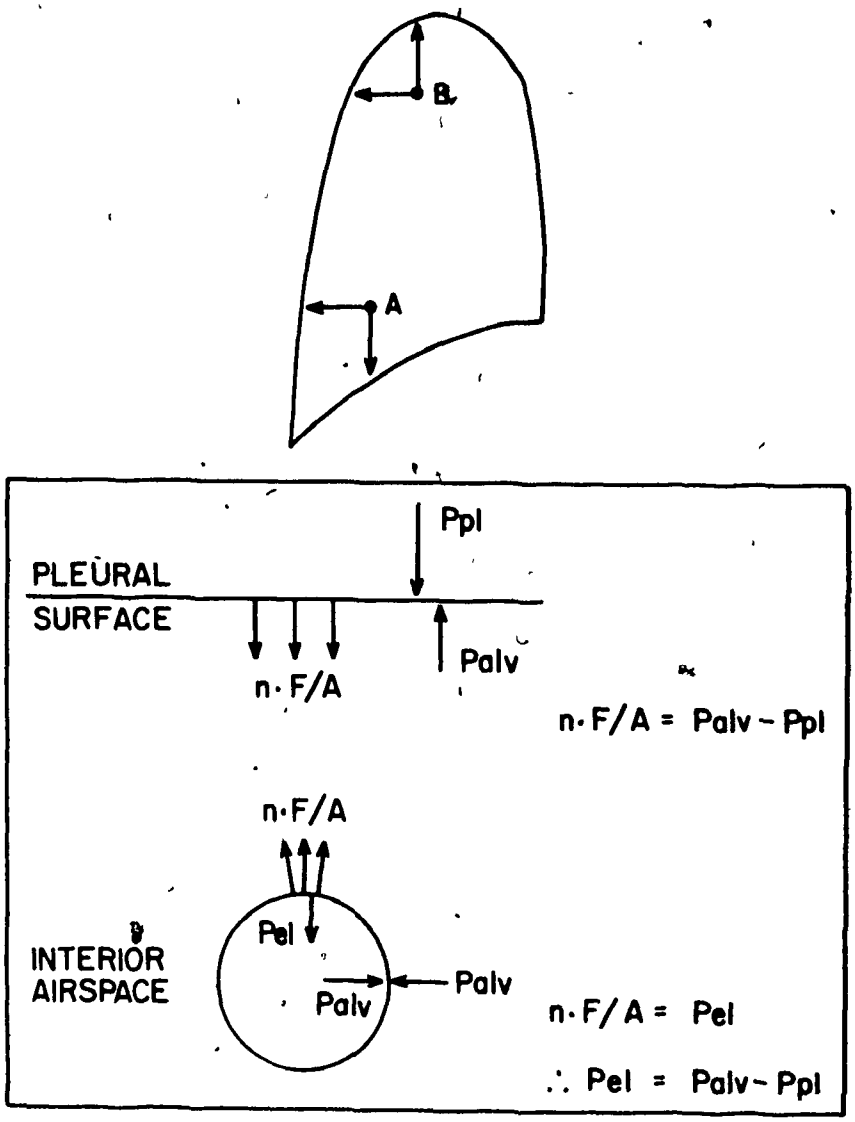

Fig. 1 a) top: Hypothetical lung showing arbitrary points A and B that are both equidistant from two different levels of pleural surface. See text. b) bottom: Diagram of forces at the pleural surface and at an interior airspace showing how the airspace becomes subjected to transpulmonary pressure. n is the number of tissue attachments per unit area and F is the average force per attachment. nF/A is constant throughout the homogeneous lung.

airways and airspaces is the same, there are no pressure differences across alveolar walls and all forces distending airspaces must arise from tissue attachments. If n/A is the number of tissue attachments per unit area at any transecting surface and F be the average normal force per attachment, then the stress at any surface is nF/A and is constant throughout the parenchyma. At the pleural surface (considered essentially flat and having no recoil of its own) the stress nF/A is balanced by and equal to transpulmonary pressure, $P_L = P_{alv} - P_{pl}$ (Fig. 1B). The outward distending stress of an interior airspace having an elastic recoil, P_{el} is also nF/A. Since P_{alv} acts on both sides of the airspace thus cancelling, $P_{el} = nF/A = P_{alv} - P_{pl}$. Thus the interior alveoli are subjected to the same distending pressure, acting via tissue attachments, as those at the pleural surface.

The authors next considered one interior region of the lung to be at a new volume, V different from its volume, V_o, in the homogeneous lung and made two more assumptions, 4) that the dimensional changes associated with the volume change were uniform, so that the surface area, A, of the region varied as $(V)^{2/3}$ and, 5) that the strain in immediate tissue attachments was negligible. It was admitted by the writers that the latter assumption would underestimate the expanding stress applied to the region by the recoil of the surrounding tissues. In any case, the stress became $P_{pl}\ (V_o/V)^{2/3}$. This is different from the distending pressure, P_{pl}, of an alveolus in a homogeneous lung because of "mechanical interdependence" via the tissue attachments. The principal function of the mechanical interdependence appears to be to support uniform expansion of airspaces which is important to the static and dynamic stability of the lung as well as to other aspects of pulmonary function.

Can this analysis be extended to account for pleural pressures that

vary topographically? Such variation immediately invalidates the foregoing assumptions, 1) and 2) and perhaps 4) and 5) if there is sizeable deformation present. Although the principles outlined by Mead et al are operative in the lung and require more detailed study we are unable to carry the analysis forward. Holland (25) looked at interdependence using a mathematical lung model that assumed that all stress-strain relations in the parenchyma obeyed Hooke's law. It possessed the desired properties of interdependence and could be used to simulate small changes of P_L. However to correctly simulate the distribution of alveolar expansion under the influence of a pleural pressure gradient, a model is needed that treats the lung as composed of finite polyhedral elements with each element possessing a true pressure volume characteristic expressed mathematically as an exponential. The pressure-volume curve defines the stress-strain characteristic of the element (25). An equation of static forces could then be written for each element. All the elements at the pleura would be assigned different pleural pressures. Permitting a little speculation, perhaps the resulting enormous array of equations defining static equilibrium could be solved by matrix elimination on a computer (5). Known differences between upper and lower lobe elastic properties could be included in the input data. Such an analysis would enable usage of pleural pressure data obtained by direct measurement, for the prediction of regional alveolar expansion for the inhomogeneous lung and following that, ventilation and its distribution, at least at very low flow rates. Addition of a multi-compartment visco-elastic model to the analysis would be a further refinement enabling calculations at higher flow rates.

In any case, alveoli deep inside the lung must receive different contributions of distending pressure from the entire lung surface through the

network of tissue attachments. The contribution from any one area of the lung surface will be a complex function of the geometry of the lung. To date, no studies have looked at this, but it seems probable that expansion of a lung region at any arbitrary surface 'A' of the lung can be achieved by a negative change of pleural pressure at some other arbitrary surface 'B'. In the process, the pleural pressure at 'A' may also change negatively, but it will be the result of and not the cause of the regional expansion at 'A'.

Chapter II

THEORY AND METHODS

THEORY

Katsura et al (28) and Robertson et al (42) have shown that when excised dog lungs are subjected to a localized distorting force, they do not behave as a liquid as Kreuger (30) and later Milic-Emili and his associates (7, 27, 36, 37) proposed to explain the pleural pressure gradient. Hoppin et al (26) suggested that the lung may be liquid-like at low lung volumes but stiffer at higher volumes. Agostoni and D'Angelo have provided powerful evidence against the hypothesis of liquid-like behaviour of dog and rabbit lungs in vivo and the entire debate has been well reviewed by Agostoni (1). (See also Chapter I).

Evidence for the actual distortability of the lung relative to the chest wall is still lacking. In mechanics, deformation is defined as a change in the shape of a body accompanying a stress condition (38). Distortability or deformability of the lung or chest wall may then be defined as their ability to be deformed into shapes other than their shape under homogeneous conditions. The lung is homogeneous when excised and inflated with positive pressure (28) and the chest wall may be considered to be in its natural shape at a given volume when the pressure inside it is everywhere uniform. In contrast to the lung and chest wall, materials such as rubber, plastics and metals have nearly the same stress-strain relationship in all directions and are called isotropic. Certain crystals have a relationship that is different on each of two perpendicular axes and are called anisotropic. Strictly speaking, isotropicity can only be attributed to continuous elastic media and not to bodies

or structures. It is sometimes better to consider the latter as having one or more degrees of freedom (29) where the elastic properties in each degree of freedom may be different. The chest wall is such a body with two degrees of freedom, i.e., there may be movement of the rib cage and/or the diaphragm (29). The lung in one sense is almost isotropic to homogeneous stresses because the stress-strain characteristic of lung tissue, although curvilinear (33, 45), is the same at all points for any one lung volume. Hence it expands proportionately by approximately the same amount in all directions (10, 16). However, if the lung is unevenly deformed, different regions will be at a different point on their curvilinear stress-strain characteristic. Areas that are compressed may reach their maximum strain (λ_{max}) thus resisting further deformation while others remain relatively unchanged. Therefore the lung in general is anisotropic. As a result it may be much harder to deform the lung than would be expected from knowing its homogeneous compliance alone. The difficulty arises because the lung in fact is a body composed of many parts (tissue attachments) and not a continuous elastic medium. Unlike most solids, its elastic properties do not obey Hooke's law. Strain arises from the bending as well as from the stretching of lung fibers as in a "nylon stocking" (33). The distinction between distortability and compliance is important. Consider a steel cylinder with a moveable piston. This is a body with one degree of freedom. It may be very compliant to forces acting in a direction to move the piston but relatively undistortable to forces acting perpendicular to the wall. Thus a body may be very compliant but relatively undistortable. On the other hand, rubber being isotropic, is compliant in every direction so the more compliant it is, the more distortable it is likely to be.

The lung and chest wall are compliant when acted upon by forces caus-

ing homogeneous expansion but may resist forces tending to deform them because these act in a different direction. It follows that to compare distortabilities in two different bodies, the stress applied to both must be the same as to direction and surface area to which it is applied. If there is variation of size and geometry of two otherwise identical bodies, their response to a deforming stress will be different. In destructive testing of concrete, all these conditions may be easily controlled whereas in comparing distortabilities of lung and chest wall they cannot. However if the above factors are neglected for the moment, some insight into the distortability of lungs and chest wall can be obtained.

In this thesis we describe experiments which test the hypothesis that the lung is very distortable (liquid-like) and which help to describe the nature of the mechanical coupling between lung and chest wall. The experiments involved isolating one lung in situ and inflating one or two lobes of that lung with positive pressure and measuring the change in volume of the remaining lobes, which had been left open to atmosphere to inflate or deflate as the case may be. The change in volume of these remaining lobes, (ΔVy) expressed as a per cent of the total volume pushed in (ΔVx max) was used to indicate the distortability of the lung. It is useful to consider this experiment in the light of some theoretical limits of behaviour of the lung and chest wall. Refer to Fig. 2.

1) If the lung behaved purely as a liquid, it would be infinitely distortable. If positive pressure inflation of one region were performed slowly enough, the same volume of air would come out of the other region even for a very compliant (not infinitely compliant) chest wall. Pleural pressures would remain unchanged, indeed very little positive pressure would be required

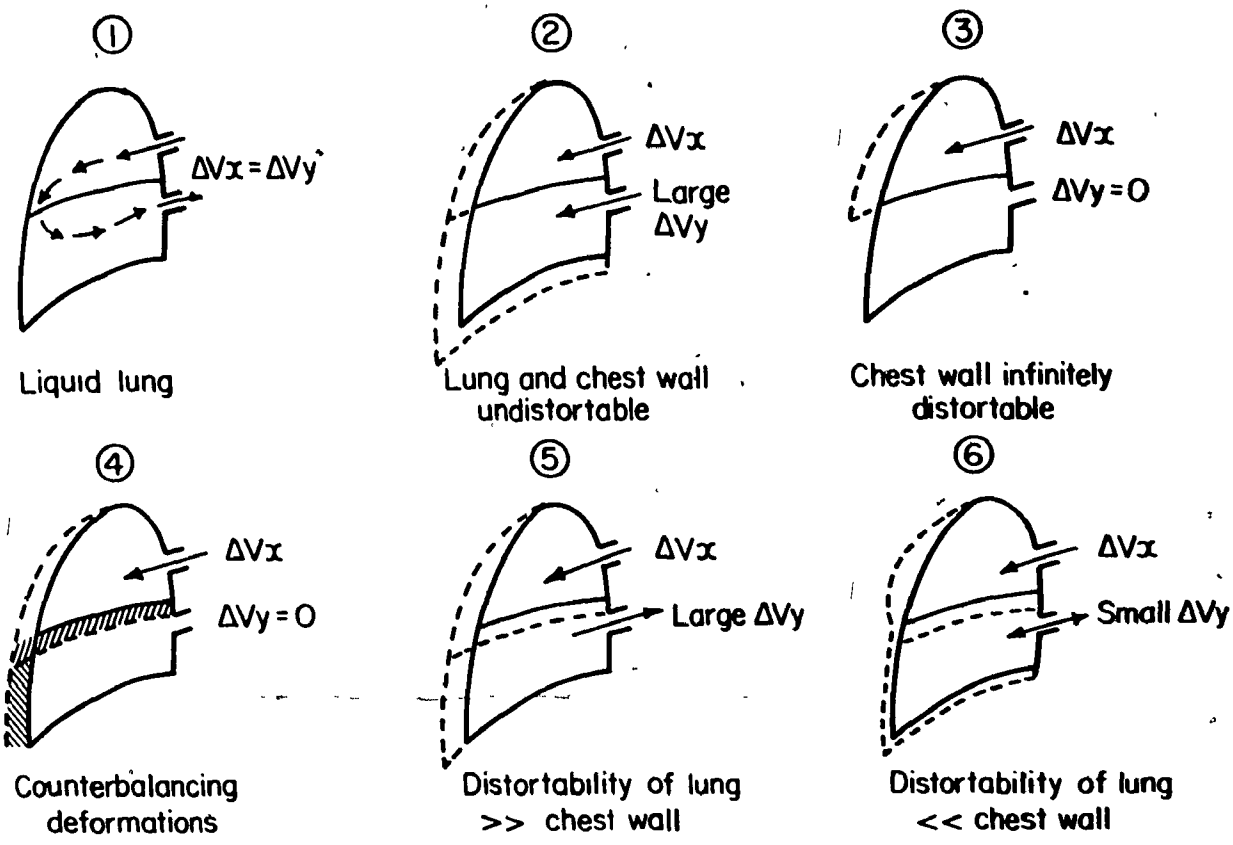

Fig. 2. Schematic representation of the six categories of lung and chest wall distortability described in the text. The solid outline is the lung-chest wall boundary before inflation of the upper region. The fissure line is also shown. Dashed lines indicate the new positions that the chest wall and fissure might assume for each category.

to execute the maneuver. Obviously the lung has elastic properties and is not a liquid but it may show substantial liquid-like behaviour in that inflation of one region would make pleural pressures everywhere more positive (Pascal's law) resulting in a large deflation of the other region.

2) If the lungs and chest wall possessed normal compliance in their respective degrees of freedom but were undistortable, inflation of one region would move the chest wall out and result in an inflation of the other region so that both regions would be inflated uniformly to the same transpulmonary pressures. However, because one region was inflated with positive pressure and the other with negative pressure, there would be large differences in pleural surface pressure.

3) If the chest wall were infinitely distortable which can be visualized by thinking of it as being absent, or the lungs being immersed in liquid, the region being inflated would not in any way stress the other region. Transpulmonary pressure differences between regions would be large but pleural pressures would not change.

These limits are admittedly abstract; the next step considers the question, "What is the relative distortability of the lung with respect to the chest wall?". Relative distortability could only be analyzed in quantitative detail if body size, shape and direction of stresses in relation to direction of degrees of freedom of lung and chest wall were all taken into account. That is beyond the scope of this thesis but such geometrical factors can be considered qualitatively by the way they modify the underlying phenomena of interaction between the lung and chest wall.

4) The distortabilities of the lung and chest wall, as modified by geometrical factors, may be such so that the volume change resulting from a

region being deformed inward at the fissure would be exactly counterbalanced by the volume change resulting from outward deformation along the chest wall. The spirometer would record $\Delta Vy = 0$. One would expect differences in pleural and transpulmonary pressure, the magnitude of which would depend on the absolute lung or chest wall distortability at which the counterbalancing occurred.

Cancellation of inflating and deflating deformations may also occur even if the lung is very much more distortable than the chest wall because the chest wall has two degrees of freedom to produce volume changes without deformation. In practice, though not exactly compensatory, this mechanism may operate in conjunction with that of case 5 following.

5) The lung may be much more distortable than the chest wall so that even geometrical factors would not be sufficient to reverse the large negative ΔVy. Under these circumstances, as one region was inflated with a known amount of air, a substantial but lesser volume of air would leave the other region. Pleural pressures would become more positive and there would be differences in the transpulmonary pressures across the two regions. There would be a small deformation of the chest wall and a large deformation of the lung. The final limit would be category 1 already discussed (liquid-lung).

6) The lung may be much less distortable than the chest wall so that again geometrical factors may be neglected. As one region was inflated, the other region might be expected to deflate by a relatively small amount (the limit being category 3 above) or inflate because of a stiff chest wall. The limit then would be that described in category 2. In any case, one would expect differences in pleural surface pressures, the magnitude of which would depend on chest wall distortability. Differences in transpulmonary pressure across the two regions would be less than would be the case if the lung were

more distortable than the chest wall and would approach zero as described in category 2. The end result of this condition might also be very similar to the end result of category 4.

To summarize: To the extent that the lung is more distortable than the chest wall, transpulmonary pressure differences over the two regions would be large but differences in pleural surface pressure over the regions would be small. To the extent that the lung is less distortable than the chest wall, transpulmonary pressure differences over the two regions would be smaller at the cost of larger differences in pleural surface pressure. Finally the foregoing analysis clearly demonstrates that lung deformation in situ cannot be analyzed without taking into account the effect of the chest wall.

Methods

Experiments were done on anesthetized paralyzed supine mongrel dogs weighing between 45 and 55 lbs. (20 - 25 kg). The experimental apparatus (Fig. 3) was designed to permit inflation of the right upper lobe with a 1500 ml syringe while the lower, middle and cardiac lobes were isolated and connected to a spirometer to measure their volume change as they were subjected to deforming stresses at the fissure and the chest wall. The left lung was intubated and obstructed at FRC with a 30 cc. Foley catheter located fluroscopically. This catheter had the rubber portion containing the main lumen sliced off to reduce its thickness and thereby minimize tracheal obstruction. Another 5 cc. Foley catheter was located in the right intermediate bronchus by the following technique: It was first inserted an appropriate distance (determined by experience) intended to be ideal for isolation of upper and lower lobes. Then minor adjustments were made according to the results of a homogeneous inflation of the entire right lung with the syringe. When the inspira-

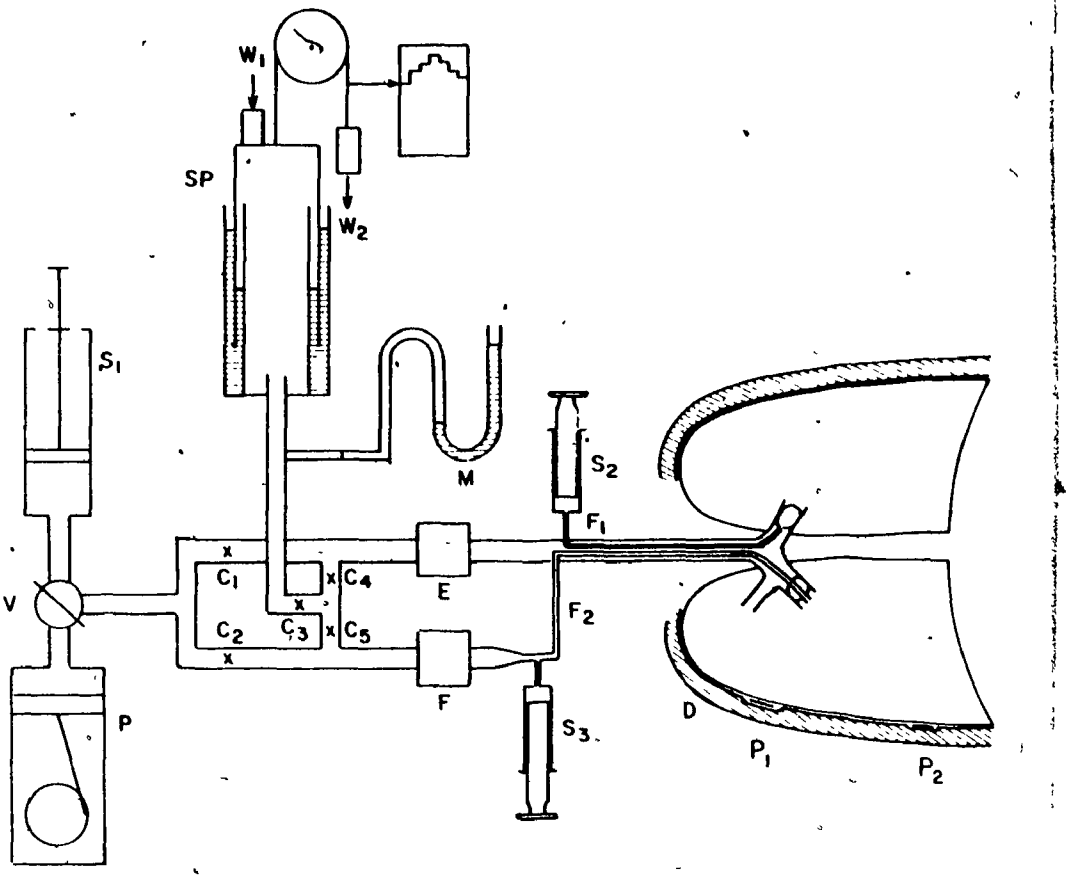

Fig. 3. Schematic diagram of the experimental apparatus for separate inflation of upper and lower lobes.

C_1, C_2, C_3, C_4, C_5 - hose clamp sites
D - dog, showing lung and chest wall
F_1 - 30 cc. Foley catheter with the portion containing the main lumen removed
F_2 - 5 cc. Foley catheter
E - Fleisch number 1 pneumotachograph
F - Fleisch number 0 pneumotachograph
P - ventilating pump (Harvard number 607)
P_1, P_2 - pleural pressure-sensing balloon capsules
M - water manometer
S_1 - 1500 millilitre syringe
S_2, S_3 - 10 millilitre syringe
SP - spirometer (4.56 millilitres/millimeter displacement)
V - three-way valve
W_1 - weight for positive airway pressure
W_2 - weight for negative airway pressure

tory capacities of the right upper and lower lobes were measured and found to be in the order of 300 ml and 800 ml respectively, and the transpulmonary pressure held without gradual leaking, we assumed the position to be correct. If not and if the upper inspiratory capacity was too large (e.g. 600 ml, indicating that the upper region was in fact including the middle lobe) we moved the catheter outward. When the catheter was out too far, partial obstruction of the right main bronchus would occur and the upper lobe would not inflate and deflate properly. Data were successfully collected in eight dogs while several others could not be studied because the upper, middle and lower lobe bronchial take-offs were too close together. The constancy of the inspiratory capacity measurements ($\pm$ 100 ml) indicated that the Foley catheter position had not changed. For these inflations of the right lung, total lung capacity was assumed to correspond to a transpulmonary pressure of 30 cm H_2O as indicated by P_{ao} relative to esophageal pressure (P_{es}). Site C3 (Fig. 3) being clamped, regional volume changes were separately measured by electrically integrating the signal from two Fleisch pneumotachographs (No. 1 and No. 0) connected to the inflow of each region. Regional inspiratory capacity was recorded for subsequent reference during inflation of an individual region. (Under these circumstances, esophageal pressure was not considered truly representative of regional pleural pressure.)

The steps of the actual experimental maneuver began with an inflation to TLC by blocking the expiratory line of the pump for a few breaths and then stopping the pump at FRC. Site C2 and C4 were clamped while C1, C3 and C5 were left unclamped. The Foley catheters were inflated thus obstructing the left lung and isolating the right upper and lower lobes. The upper lobe was inflated by the syringe in steps of 100 ml to full inspiratory capacity while

the lower lung was left free to inflate or deflate via the Foley catheter into the spirometer. (A spirometer was used because it was found that the pneumotachograph-transducer combination was not stable enough to measure the small volume changes.) From peak inflation, the upper lobe was deflated to a P_L of about -10.0 cm H_2O and inflated back to FRC all in steps of 100 ml. The entire maneuver took about 30 seconds and was done twice. It was then repeated twice with the upper and lower regions reversed, i.e. C1 and C5 were clamped allowing the lower lung to be inflated while the upper lobe was connected to the spirometer (C2, C3 and C4 unclamped).

In order to determine if lung volume influenced the relative distortabilities of lung and chest wall, the entire foregoing procedure was repeated starting at right lung alveolar pressures of 10, 20 and -5 cm H_2O, i.e., the alveolar pressure for both regions started at the same value. The alveolar pressure in the region connected to the spirometer was achieved by adding weights to the spirometer bell until the water manometer indicated the desired pressure. In this way, as the spirometer bell moved during the maneuver, alveolar pressure of its associated region was kept constant as the other region was inflated and deflated. To create negative alveolar pressures the weights were simply hung on the opposite side of the spirometer pulley. In general, inflation of the lower region tended to push the inflated Foley balloon back up the bronchus which hindered the collection of data for that type of maneuver.

The entire protocol was combined with that of simultaneous pleural pressure measurements in dog 1. This proved to be very lengthy and so to look at pleural pressure under conditions of lung deformation, additional measurements were carried out independently on eight dogs during spontaneous and

intermittent positive pressure ventilation (IPPV). The surgical procedure and insertion of the flat pressure measuring capsules were based on procedures first used by Hoppin et al (26). Basically it consisted of making an 8 cm incision in the fifth intercostal space. A pressure-sensing capsule with surrounding skirt and 0.2 mm thick X-ray film backing was located over the right upper and right lower lobes, in the fissure and over the ventral and dorsal portions of the diaphragm. Each device was held taut by attached silk ligature brought externally through perforations in the costal wall. These could be used to alter the position of the capsules under radiological control. Pneumothorax was removed by repeated nitrous oxide infiltration followed by pneumatic suction via a catheter inserted in the pleural space. The capsules were deflated in situ to a pressure of -20 cm H_2O and inflated with a predetermined ideal volume to minimize artifacts due to balloon elasticity.

All pressures were measured with Hewlett-Packard model 267B transducers and the recording channels were balanced and calibrated with a water manometer before every experiment. An electrical calibrating signal equivalent to 40 cm H_2O was established using the zero suppression circuit on the carrier amplifiers; calibration remained within $\pm$ 0.5 cm H_2O throughout each experiment.

While the dog was being ventilated, a consistent volume history was established by blocking the expiratory line until the lung reached TLC. Following return to FRC, the left main and the right intermediate bronchi were obstructed. Pleural pressures were recorded before and after obstruction. The procedure was similar for the spontaneous breathing dog except that prior inflation to TLC was omitted.

Chapter III

RESULTS

Combination of the data for upper and lower lung regions for each alveolar pressure was possible because the results were qualitatively the same in both cases. Although more air was usually pushed out of the caudal lobes, the difference was not significant. The effect of differences in the sizes of lobes, has been normalized by expressing the results as per cent.

Volume Data

The term ΔVx refers to the volume change of a lobe or lung region being inflated (positive ΔVx) or deflated (negative ΔVx) by the applied action of the syringe while ΔVy refers to the volume change of the region connected to the spirometer. Hence ΔVx is always the independent and ΔVy the dependent variable. They are both referenced to either FRC ($P_{alv} = P_B$) or to lung volumes which have alveolar pressures of 10, 20 and -5 cm H_2O.

Figure 4 shows mean results for upper and lower lung regions connected to the spirometer held at atmospheric pressure, P_B. While one region was inflated with a syringe, the other region deflated progressively. The total mean volume change reached -12.5% $\pm$ 2.5 (S.E.) of the total volume pushed in by the syringe (ΔVx max). Then ΔVy remained constant while ΔVx decreased by 20% revealing "hysteresis". Further decreases of ΔVx caused ΔVy to become progressively more positive (inflation). In one individual experiment ΔVy max was 39% of ΔVx max (at $P_{alv} = P_B$) while in all other cases it was between 4% and 14.5% of ΔVx max. To consider the possibility that the lung may be very liquid-like for smaller deformations, see Fig. 5. The points at V_{in} equal to 20, 40, 60 and 80 ml are nearly on the same line produced by larger deformations.

Hence no marked increase of distortability was demonstrated. In retrospect it would be highly unlikely that the response for any of the data could start off with a very steep rate of change of ΔVy for small volume changes. To do so, it would then have to flatten off in order to get back on the response for larger volume increments.

At volumes below FRC, (P_{alv} = -5 cm H_2O), there was a slight reduction in slope of the curve (Fig. 6) as ΔVx approached ΔVx max. This is evidence that airway closure began to occur in the region deflating into the spirometer.

At volumes above FRC (Figs. 7 and 8) there was reversal of the direction of hysteresis and ΔVy was less than when $P_{alv} = P_B$. Also, ΔVy became positive before ΔVx returned to the starting point at zero. Thus lung inflation occurred secondary to the applied deformation.

A comparison of lung distortability relative to the chest wall at different lung volumes can be obtained from Fig. 9. None of the responses came close to the isovolume line that represents pure liquid-like behaviour. Greatest distortability (i.e. slope of the ΔVy vs. ΔVx plot) occurred at FRC where $P_{alv} = P_B$. Surprisingly, the mean initial slope was less at a lower lung volume (P_{alv} = -5 cm H_2O), but this difference was not significant. Since one would expect the lung itself to be more distortable at low lung volumes on the basis of directly observable properties of excised lobes, this must mean that the chest wall also becomes more distortable. As one would expect, at higher lung volumes, the lung parenchyma becomes quite stiff and at P_{alv} = 20 cm H_2O, ΔVy was zero for all ΔVx.

Pressure Data

Although simultaneous pleural pressure measurements were made in only

one experiment, the results are all quite consistent (Figs. 10 - 13). In all cases, pleural pressures at the costal wall rose progressively over the region being inflated by the syringe as one would expect. The transpulmonary pressure-volume curves all have the characteristic well-known shape. However, in all cases, very little change in pleural pressure occurs over the region connected to the spirometer (at the point measured). Thus very large differences in pleural pressure along the costal wall occurred as in Fig. 10A where the difference reached 20 cm H_2O. Also the direction of the pleural pressure gradient became completely reversed in going from high lung volume (upper right quadrant) to low lung volume (lower left quadrant). Some of the difference may be due to a costal artifact caused by one pressure capsule being in an interspace while the other was under a rib but Fig. 10B shows a difference of 11 cm H_2O in the opposite direction when the lower lobes were inflated. The devices were not manipulated between these two maneuvers and it is unlikely that they could have changed position significantly due to the motion of ventilation. If the difference of 20 cm H_2O is too high because of costal artifact, then the difference of 11 cm H_2O must be too low for the same reason; therefore the real difference between mean pleural pressures at each site must lie somewhere between 11 and 20 cm H_2O which is still very large.

Fig. 11 shows the pressures taken from Fig. 10A plotted against the volume change of the corresponding lobe having $P_{alv} = P_B$. At first sight it seems paradoxical: pleural pressure became more negative as volume decreased. One can only conclude that pleural pressure was increasing at some other point (most likely in the fissure), thus re-emphasizing that differences in direction of change of pleural pressure can occur.

Results of the pleural pressure measurements made in eight independent

dogs are summarized in Table I (page 46). Only the changes in pressure are shown because these are directly relevant to the theory developed herein. Absolute pressures recorded by the capsules are true at the point measured but may not represent the mean pressure at that lung height because of costal variations, despite the plastic backing. Therefore, without more measurements at different lung heights, interpretation of absolute pressures at only two sites might be misleading.

The data clearly demonstrate that 1) pleural pressure changes during normal ventilation are radically different over different surfaces of the lung. Furthermore, regional differences are not systematic among animals. 2) The differences increase under conditions of increased inhomogeneity of ventilation such as occurs during lobar occlusion. 3) The reduction of pleural pressure change ΔP_{pl} on the costal side acts in a direction to increase trans-lobar pressure ($P_{ao} - P_{pl}$) at end inspiration (e.i.) of the occluded lobe. The latter point is readily seen by noting the change of magnitude of ΔP_{pl} over the lower lobes from control to occlude conditions. During IPPV it was reduced without exception by amounts up to 4.7 cm H_2O. Therefore $P_{ao} - P_{pl}$ at e.i. must have increased. The increase of pleural pressure over the right upper lobe simply reflects the increased tidal volume received from the constant volume pump circuit following lower lobe occlusion. Being careful to account for algebraic sign, $P_{ao} - P_{pl}$ at e.i. increased on the lower costal side for both IPPV and spontaneous ventilation.

Observations 1) and 2) are of central interest to the theoretical discussion. Observation 3) is not as directly relevant and cannot be extended to apply to the entire pleural surface. Fissure and diaphragmatic pleural pressure changes occurred in both directions of magnitude. For example, in

dogs 1 and 8 during IPPV, the pressure in the fissure increased, rather than decreased, by 0.9 and 0.8 cm H_2O respectively. The occurrence of these increases simultaneously with decreases at the costal surface reflects the fact that the lung was being subjected to deforming stresses.

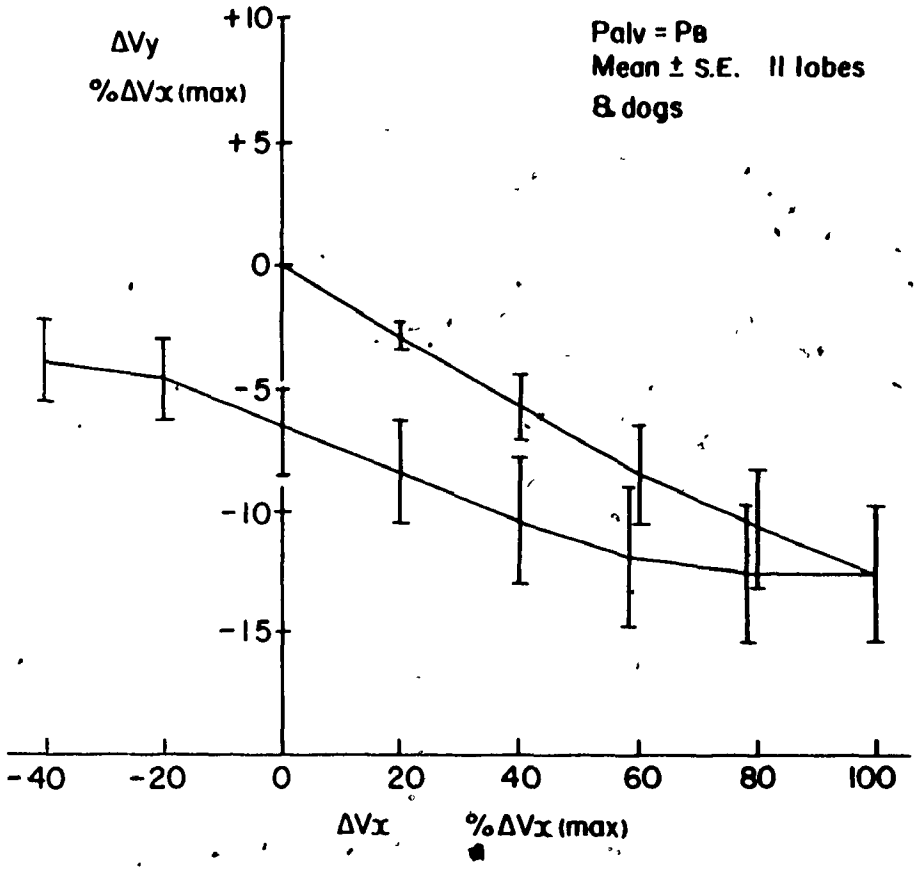

Fig. 4. Results of ΔVx% plotted against ΔVy% (mean ± S.E.) for 11 lobes with $P_{alv} = P_B$. ΔVy progressively reached -12.5% (deflation) of the total volume of air pushed in with the syringe (ΔVx max). Upon withdrawal of the syringe, ΔVy increased revealing "hysteresis".

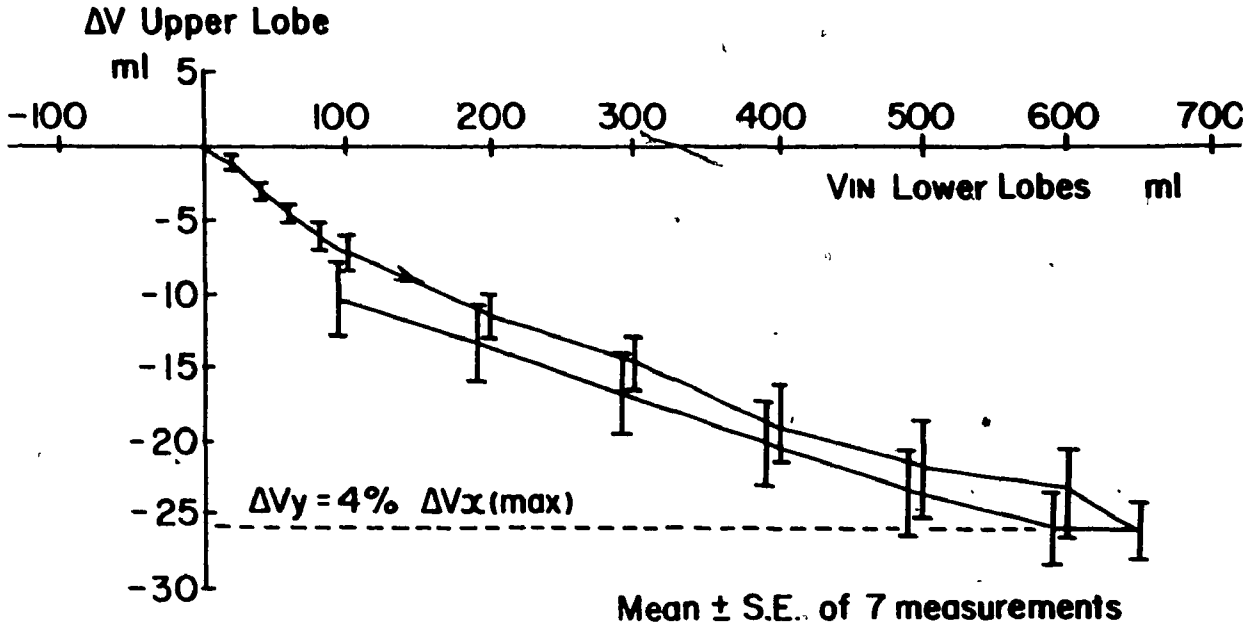

Fig. 5. Result of ΔVx in ml plotted against ΔVy in ml (mean $\pm$ S.E.) for one dog. Considering the slope as an indicator of lung distortability, no marked increase (i.e. by several orders) was found for very small changes of ΔVx.

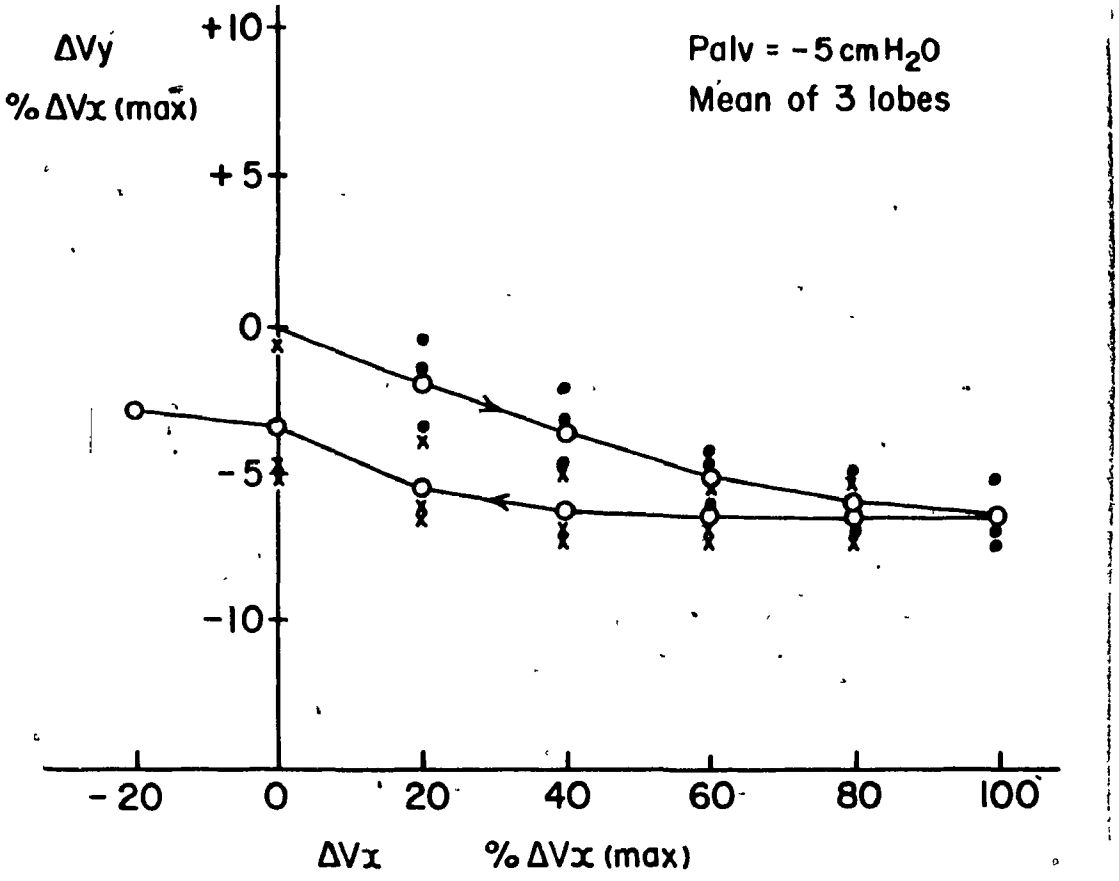

Fig. 6. Results of ΔVx% plotted against ΔVy% for 3 lobes at very low lung volume. Closed circles are data points for ΔVx% increasing and the x's are data points for ΔVx% decreasing. Open circles represent the mean.

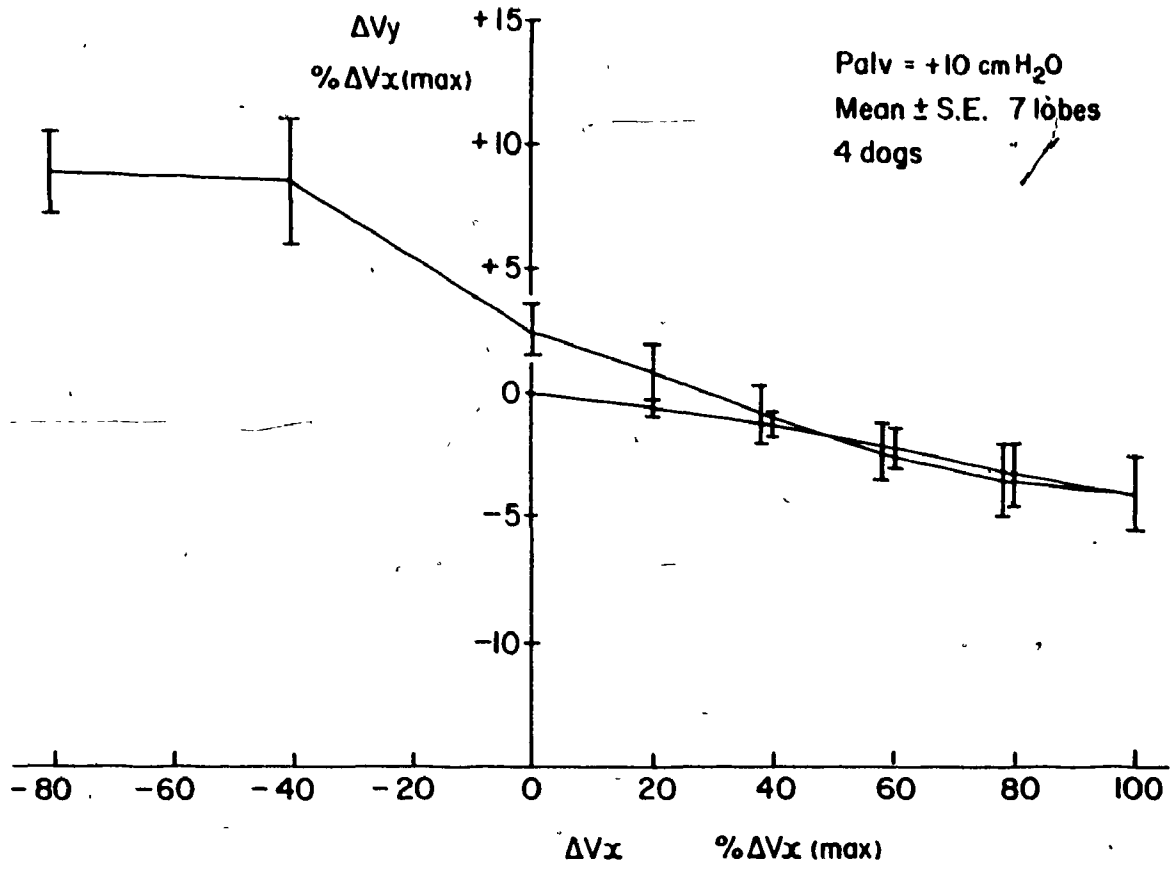

Fig. 7. Results of $\Delta Vx\%$ plotted against $\Delta Vy\%$ (mean ± S.E.) for 7 lobes at P_{alv} = 10 cm H_2O.

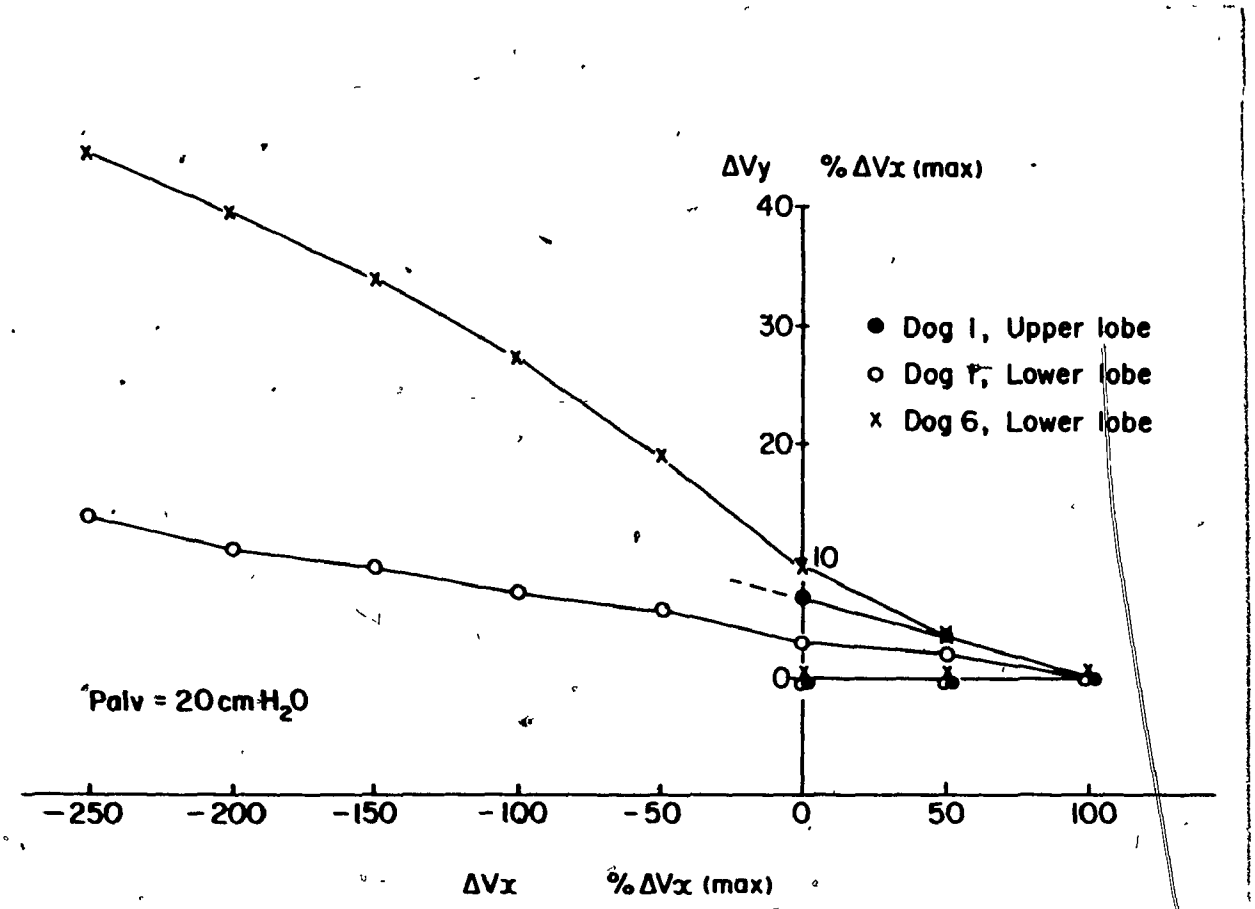

Fig. 8. Results of ΔVx% plotted against ΔVy% for 3 lobes at a high lung volume (P_{alv} = 20 cm H_2O).

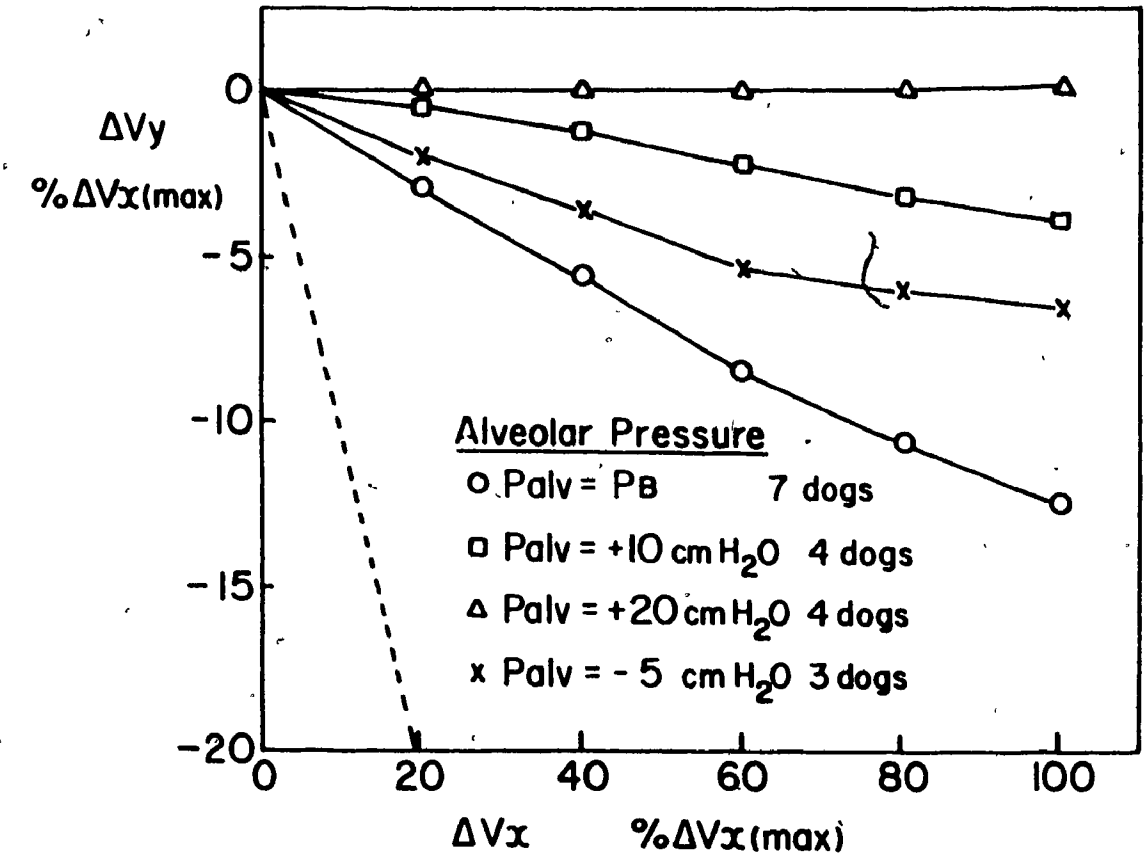

Fig. 9. Summary of the results at different lung volumes for ΔVx% increasing, taken from Figs. 4, 6, 7 and 8. The slope is steepest for lobes at their FRC. The line of identity (dashed) represents pure liquid-like behaviour (See text, category 1).

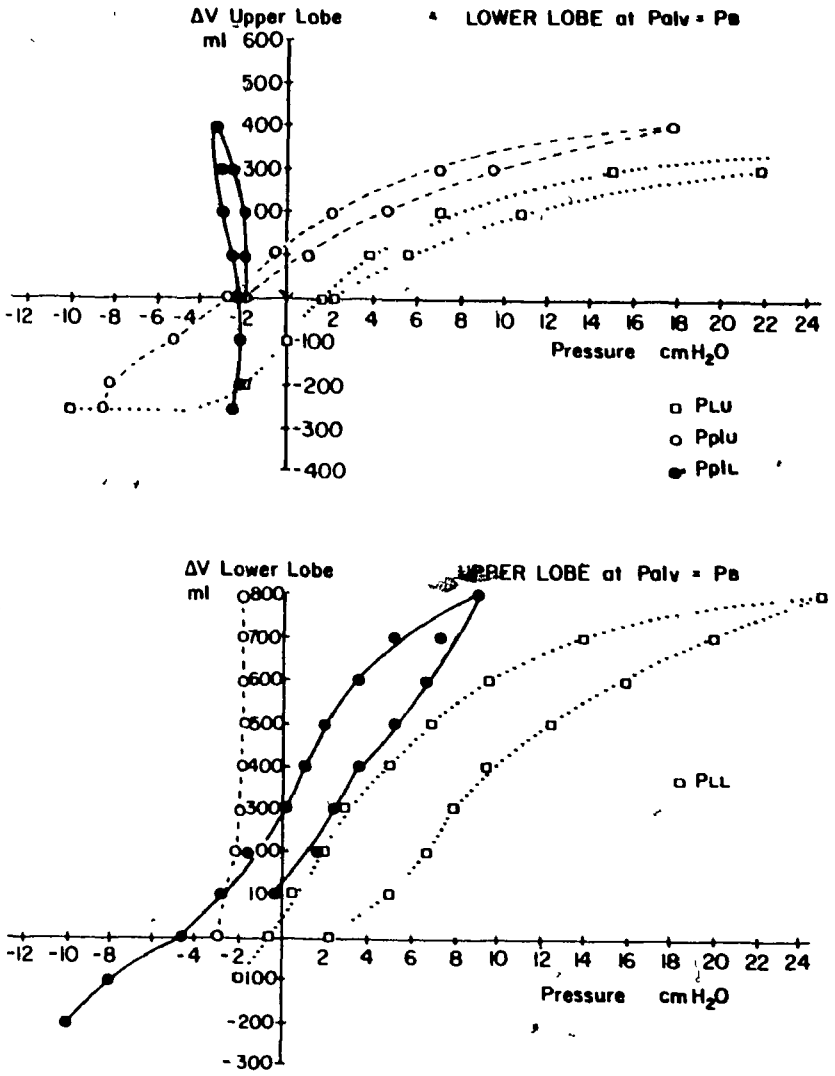

Fig. 10 a) top: Upper and lower regional pressures plotted against the applied change of volume ΔV of the upper lobe. b) bottom: Upper and lower regional pressures plotted against the applied change of volume ΔV of the lower lobe. The starting point ($\Delta V = 0$) for a) and b) was with both upper and lower regions at $P_{alv} = P_B$.

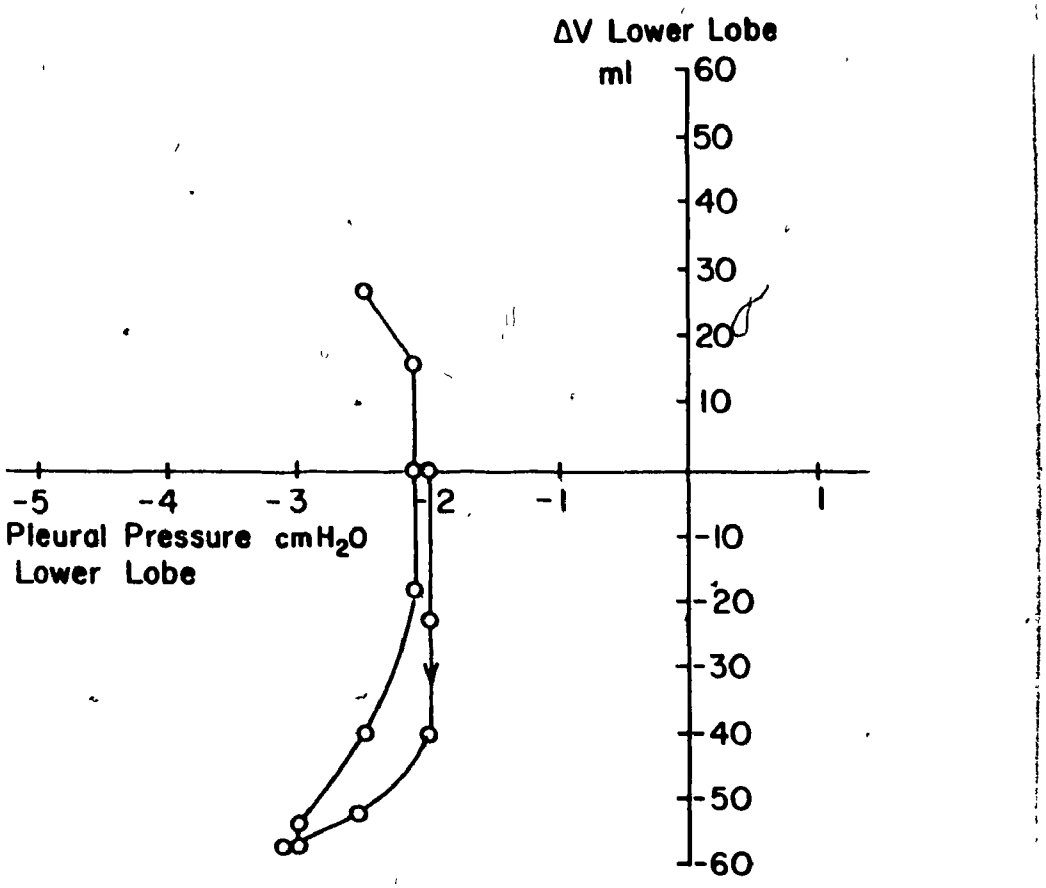

Fig. 11. Pleural pressure over the lower lobe (from Fig. 9a) plotted against the volume change ΔV of the lower lobe suffering impingement. The direction of change of pressure and volume indicate that stresses of opposite sign were being applied to the lobe.

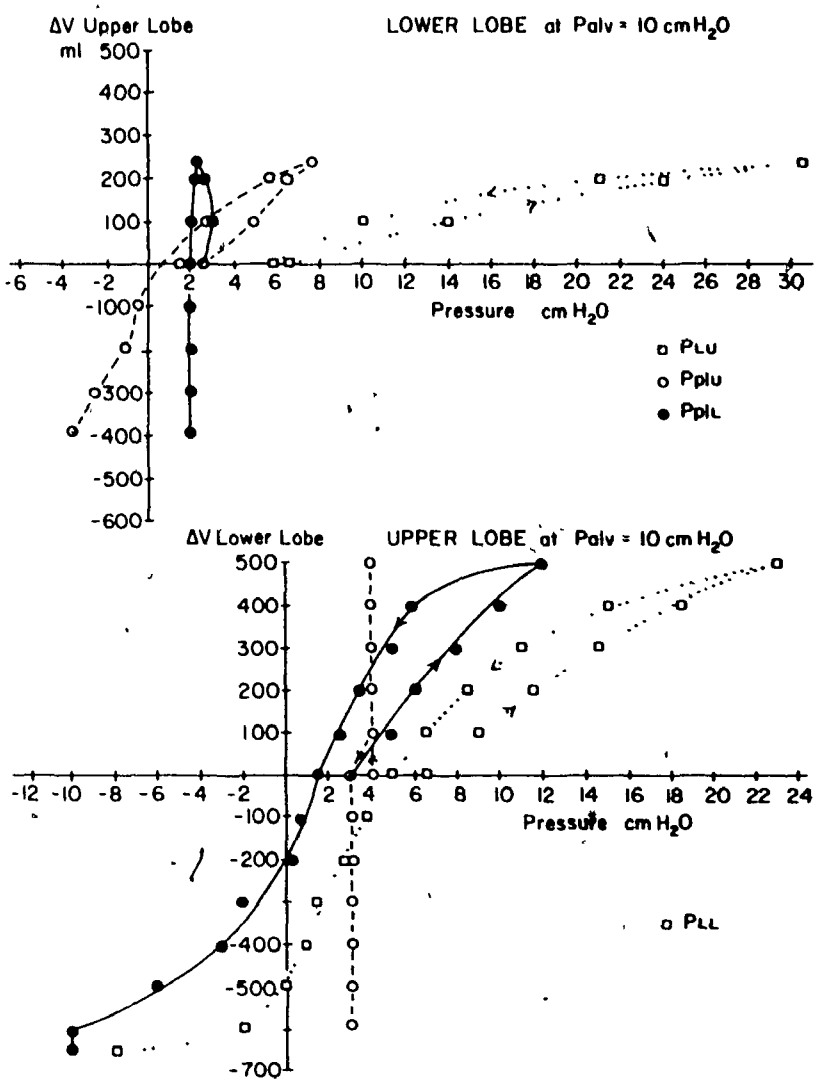

Fig. 12 a) top: Upper and lower regional pressures plotted against the applied change of volume ΔV of the upper lobe. b) bottom: Upper and lower regional pressures plotted against the applied change of volume ΔV of the lower lobe. The starting point ($\Delta V = 0$) for a) and b) was with both upper and lower regions at $P_{alv} = 10$ cm H_2O.

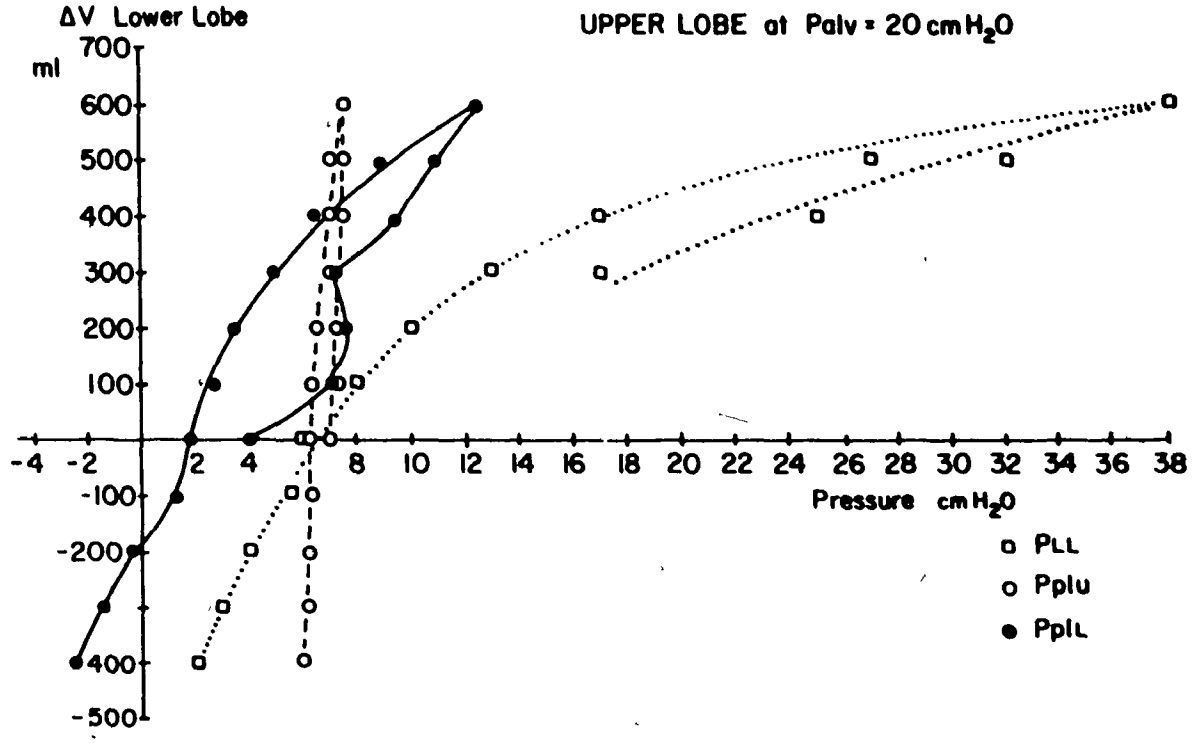

Fig. 13. Upper and lower regional pressures plotted against the applied change of volume ΔV of the lower lobe. The starting point ($\Delta V = 0$) was with both upper and lower regions at P_{alv} = 20 cm H_2O.

Table I

Pleural Pressures During Lobar Occlusion

Intermittent Positive Pressure Ventilation (V_T = 200 - 300 ml)

Dog No.	Condition control (C) occlude (O)	ΔP_{pl} Upper Lobe	ΔP_{pl} Lower Lobe	ΔP_{pl} Fissure	ΔP_{pl} Ventral Diaphragm	ΔP_{pl} Dorsal Diaphragm
1	C	3.8	2.3	4.5	4.1	1.7
	O	4.1	1.7	5.4	1.4	1.3
2	C	9.6	5.4	8.0	3.8	3.3
	O	15.8	2.6	5.2	2.5	0.5
3	C	6.9	5.7	9.3	0.4	5.4
	O	16.0	1.0	8.0	0.4	0.4
4	C	2.4	3.7	2.8	-	-
	O	2.4	1.7	1.5	-	-
5	C	5.0	3.0	4.0	3.0	-
	O	7.0	2.4	3.4	2.2	-
6	C	5.4	4.1	3.2	0.9	-
	O	10.0	1.5	0.9	1.6	-
7	C	4.0	2.5	1.4	-	-
	O	5.0	0.8	0.2	-	-
8	C	4.4	3.4	3.2	-	-
	O	8.6	2.3	4.0	-	-

Spontaneous Ventilation

1	C	-2.0	-1.8	-1.3	-1.3	-2.0
	O	-2.2	-3.0	-2.1	-1.6	-3.3
2	C	-1.4	-1.8	-1.7	-0.7	-3.8
	O	-0.7	-2.9	-3.5	-2.5	-5.1
3	C	-3.7	-1.8	-4.3	-9.0	-6.9
	O	-3.8	-6.3	-10.3	-13.7	-14.9
5	C	-1.7	-1.3	-1.8	-1.1	-0.4
	O	-1.9	-4.0	-3.1	-2.5	-2.9
8	C	-2.2	-1.0	-2.4	-	-
	O	-2.2	-5.2	-3.3	-	-

Chapter IV

DISCUSSION

Before discussing the implications of our data regarding the mechanics of the pleural space, it is important to consider how much of ΔVy was attributable to gas exchange. To theoretically determine the effects of gas exchange on ΔVy, it would be necessary to know the percentage change in lung volume ($\Delta V/V_L$) during a breath hold of 30 seconds in 20 kg anesthetized paralyzed dogs. We were unable to find such data but ($\Delta V/V_L$) may also be determined from the following formula:

$$\frac{\Delta V}{V_L} = \left(\frac{P_{N2}}{P'_{N2}} - 1\right) \times 100$$

where P_{N2} = partial pressure of alveolar nitrogen at the start of breath hold.

P'_{N2} = partial pressure of alveolar nitrogen at the end of breath hold.

Lanphier and Rahn studied breath hold in humans at rest and during mild work (31). From their partial pressures of O_2 and CO_2, P_{N2} and P'_{N2} can be calculated according to $P_{N2} = (P_B - 47 - P_{O2} - P_{CO2})$. Mean initial lung volume was 5.35 litres.

at rest: P_{N2} was 567 mmHg at t = 0

P'_{N2} was 568 mmHg at t = 30 sec.

therefore, $\Delta V/V_L$ was ≈ 0

during mild work: P_{N2} was 567 mmHg at t = 0

P'_{N2} was 588 mmHg at t = 30 sec.

therefore, $\Delta V/V_L$ was -3.5%

It seems therefore, that the mean respiratory quotient, R, during breath hold was much higher at rest than during exercise. Further conclusions can only be qualitative. The effect of gas exchange on ΔVy in our experiments was probably very small and acted in a direction to make ΔVy more positive through two mechanisms: 1) by a direct effect on the lobe connected to the spirometer and, 2) indirectly by causing volume changes in the left lung. This might shift the mediastinum and produce an inflating force over the lobe connected to the spirometer leading to an additional increase in ΔVy. If the effect of gas exchange were large, then one would expect the return limb of Fig. 4 to have a much steeper slope than the forward limb because oxygen uptake would be acting to enhance the movement of air into the spirometer. The hysteresis loop would be in the opposite direction. Thus we feel that the effect of this uncontrolled variable was small.

Our results show that induced deformations result in large differences in pleural and transpulmonary pressures consistent with the data of D'Angelo et al who also measured pleural pressures during lobar occlusion using the counterpressure technique (14). However, when they passively inflated the right upper lobe, mean pleural pressures over the lower occluded lobe went from -2.1 to -1.4 cm H_2O. With passive inflation of the right lower lobe, the change over the upper lobe was -2.5 to -1.9 cm H_2O. That is, in both cases, pleural pressure went slightly more positive. The small difference between their results and those shown in Figs. 10, 12 and 13 (pleural pressure remained almost constant) is explained by the fact that alveolar pressure of their occluded lobes could change with isovolume lung deformation whereas in our experiment alveolar pressure was held constant.

Robertson et al (42) examined lung distortability when the excised

lung was subjected to a deformation at a single point. Although this type of deformation was quite different from that of the present study, qualitatively they both reveal increased lung resistance to deformation at higher alveolar pressures. Interdependence within a lobe increases with lung volume (35) probably for the same reason that it gets stiffer: the tension of the elastic fibres is increased. Results in Figs. 4 to 9 show that there is also mechanical interdependence between lobes even though there are no parenchymal attachments. That is, the volume of one lobe impinged on that of the others. Mead et al (34) claimed that "there can be little mechanical interdependence at the lobar level" but this statement overlooks this form of mechanical interdependence which arises from the sharing of a common container. Fig. 8 also reveals lobar interdependence acting via a moveable chest wall. The plot is at first flat indicating that the lung was very undistortable and/or that any deflation was counterbalanced by inflation of another region due to an expanding chest wall. Then when ΔVx was reduced, ΔVy went positive. All the points reflect a volume higher than the starting volume, which can only occur if the return of the chest wall lagged behind reductions of ΔVx. This lag yielded net inflating forces on the lobe connected to the spirometer. The lag was presumably due to chest wall elastic hysteresis, however the presence of hysteresis is not necessary to the mechanism of interdependence via the chest wall; it just served to enable its observation.

Considering again the categories of relative distortability in Fig. 2, we conclude that the data demonstrate that the dog lung in situ is not liquid-like even at low lung volumes. However, both the lung and chest wall are distortable but the chest wall is stiff enough to cause large differences in pleural pressure. This eliminates categories 1, 2 and 3.

Fig. 11 shows evidence that counterbalancing stresses were applied to the lung (category 4). Although transpulmonary pressure, $P_{alv} - P_{pl}$ became more positive, ΔVy went more negative. Obviously stress on some other lung surface decreased. Since stress changes of opposite sign occurred, the resulting deformations would cause volume changes that would tend to cancel. Hence ΔVy is not representative of the total deformation of that region of lung. The flatness of the response in Fig. 8 may also be explained at least in part by the same phenomenon as well as by decreased lung distortability. Relatively minor shifts to either side of the counterbalanced situation could account for the overall data.

The fifth category does not describe the data well because ΔVy was not relatively large. However the sixth category might be as important as the fourth. That is, there are two models, both of which explain the small ΔVy and large differences in pleural pressure. The first of these (category 4) acts by producing deformations of opposite sign using a rigid chest wall and the second (category 6) acts by producing relatively small lung deformations but large deformations of the chest wall. The issue could be resolved by measuring chest wall deformations with magnetometers during the same type of experiment.

So far, geometrical factors have not been included in the discussion. What is the consequence? Very little, because quantitative comparisons of distortability using some appropriate index have not been made. Rather, models that simply possess various mechanisms of lung and chest wall coupling have been discussed.

The fact that it was possible to reverse the pleural pressure gradient by deforming the lung (Figs. 10, 12 and 13) supports the hypothesis that dif-

ferences in pleural pressure are caused by the deformation that the lung undergoes in fitting the chest wall (43, 17, 33, 41). Future experiments that examine the pressure cost of deformation must take into account the ability of the lungs to adapt to different types of deformation. For example, the finding by Grassino et al (22) that the gradient of alveolar expansion of the human lung in vivo did not seem to be altered by large but physiological thoracoabdominal shape changes, may simply reflect the lungs built-in capacity to adapt to those changes via inter-septal slippages, lobation and perhaps other unknown mechanisms. On the other hand, the lung does not adapt well to the deforming stresses applied by Robertson et al (42), Zidulka et al (48, 49) or by those applied in this study. The inhomogeneity of lung parenchyma was greatly increased coupled with large differences in pleural pressure. From a teleological point of view, adaptation is probably better when the deforming stresses resemble those to be coped with in real life. To the extent that adaptation is not achieved, deformation and concomitant inhomogeneity will become more severe.

Years ago, Dubois et al found that movement of the lower chest wall lagged the upper chest wall under dynamic conditions (15). A similar conclusion can be made from the results of Bake et al in normal subjects (6). From measurements of ^{133}Xe distribution at different physiological flow rates they found that the lower lung (and hence, the chest wall) lagged behind the upper lung (and chest wall). Diseases which cause airway obstruction may also result in one lung region lagging another in volume. The experiments described herein are static equivalents to the dynamic phase lag situation; i.e., two lung regions are subjected to different transpulmonary pressures at the same time. It is therefore probable that the results of the static experiments are

applicable to the dynamic situation. To this extent, deformations of the lung occurring under disease conditions, such as might occur with grossly uneven time constants, result in deformations of the chest wall overlying the lung. The pressure cost of chest wall deformations is substantial and results in an amplification of the pressure applied to the lung which acts to minimize the lung deformation.

Chapter V

SUMMARY AND CONCLUSION

The relative distortability of the lung and chest wall has been examined theoretically and it has been shown that the behaviour of one when stressed is intricately linked to the physical properties of the other. Experimental evidence clearly demonstrates that the lung in situ does not behave as a liquid even for relatively small deformations. In addition the data reveal decreased lung distortability as lung volume increases, explicable by the accompanying increased lung tissue fiber tension. To the extent that the lung cannot adapt to regional stresses via lobation, interseptal slippage and perhaps other mechanisms that help to prevent further parenchymal inhomogeneity, regional differences in pleural surface pressure are established which presumably deform the chest wall. These results are compatible with the hypothesis that the shape discrepancies between the sponge-like lung and the chest wall are responsible for the gradient in pleural pressure known to exist under normal physiological conditions.

The interaction of the lung and chest wall that leads to large differences in pleural pressure under stress conditions, results in a homeostatic mechanism which tries to preserve the same alveolar expansion and ventilation distribution. Even in normal lungs, detectable phase lags are present between upper and lower lung zones at physiological flow rates so that this mechanism may be operating continuously.

Pressure transmission through the lung cannot be simplified by application of Pascal's law ("Pressure in a liquid is transmitted equally and undiminished throughout the liquid"). The transmission must be a more complex

function of parenchymal geometry and stress-strain characteristics. Further to this, the hypothesis that there is a dissociation between local pleural surface pressure and regional alveolar expansion may account for some discrepancies in the literature. Future studies might pursue this by examining pleural pressure and regional lung volumes during various types of quantitated stresses on the chest wall.

POINTS OF ORIGINALITY

1) A new procedure has been described to examine lung and chest wall distortability and the influence that applied pulmonary stresses have on the pleural pressure in dogs.
2) It has been shown that an inhomogeneous stress on the dog lung in situ results in substantial differences in pleural surface pressure.
3) Evidence has been presented that the dog lung in situ becomes less distortable at higher lung volumes.
4) Direct measurements of costal, fissure and diaphragmatic dynamic pleural surface pressures reveal important topographical differences in control supine dogs and even greater differences in dogs undergoing lobar occlusion.

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