MEASUREMENT OF RESPIRATORY WORK AND RESISTANCE BY ARTIFICIAL RESPIRATION UNDER CONDITIONS OF IMMERSION AND RAISED AMBIENT PRESSURES

by

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ABSTRACT

It was hypothesized that upright immersion without breathing pressure compensation increases airway resistance and the work of breathing. Secondly that compensation of breathing gas pressure to lung centroid pressure (P_{LC}) will return respiratory mechanics towards normal. Finally, increased ambient pressure causes an increase in the work of breathing due to the turbulent nature of respiratory airflow.

Five subjects were each mechanically ventilated under six experimental conditions: at 1 ATA in dry conditions; immersed at 1 ATA without hydrostatic pressure compensation of breathing gas; immersed at 1 ATA with breathing gas supplied at P_{LC}; and immersed at 2, 4, and 6 ATA, with breathing gas supplied at P_{LC}. The subjects were ventilated by a hydraulically driven breathing simulator whose frequency was controlled by the investigator. Subjects relaxed to their expiratory reserve volume and were then ventilated passively for 20 seconds at a 2 litre tidal volume. Ventilations were controlled at 20, 30, 40, 50, and 60 $L.min^{-1}$ in separate trials. Pressure and volume data were collected in each condition. From these measures elastic, and flow-resistive respiratory work, respiratory resistance and dynamic compliance were calculated.

Elastic work remained constant with increasing minute ventilation whereas increasing minute ventilation produced increases in flow-resistive work ($p<0.05$) for all conditions. During uncompensated immersion at 1 ATA elastic and flow-resistive work were increased and dynamic compliance was reduced from comparative trials in dry conditions. In contrast, immersion with breathing pressure compensation to P_{LC}, showed no significant differences from dry conditions. When immersed with breathing pressure compensation flow-resistive work was increased at 4 and 6 ATA compared with 1 ATA ($p<0.05$). In addition, an interaction effect was found between gas density and minute ventilation which is indicative of a turbulent flow component. Expiratory flow-resistive work exceeded inspiratory flow-resistive work at all gas densities ($p<.05$).

Respiratory resistance was shown to be significantly greater at higher gas densities and lower lung volumes. There was also an interaction effect between airflow rate and gas density. From theoretical modelling of experimental data, expiratory resistance appeared to be more turbulent in nature than inspiratory resistance which contained a greater laminar flow component.

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CHAPTER 1

INTRODUCTION

Respiratory mechanics have been the subject of detailed study for well over a century. Mechanical components include inherent stresses and strains, fluid dynamics and respiratory power. These elements are of particular importance to respiration during immersion and hyperbaric exposure, and consequently to the design of breathing apparatus for underwater work.

There are four components in the process of respiration: pulmonary ventilation; the diffusion of oxygen and carbon dioxide between the alveoli and the blood; the transport of oxygen and carbon dioxide in the blood; and the subsequent diffusion in the tissues. Although only the first component is related to respiratory mechanics, the mechanics of respiration affect the equilibrium of all subsequent components, which in turn affect the first.

Despite advances in hyperbaric physiology, human performance is still restricted in the high-pressure underwater environment. Insufficient alveolar ventilation and increased respiratory work have been recognized as two physiological limitations (Lanphier and Camporesi, 1982). Morrison (1988) identified underwater breathing apparatus as a critical factor. Understanding the respiratory requirements of the diver and the associated physiological costs would assist the design of improved breathing equipment.

1.0.1 Respiratory Musculature

The major muscle of inspiration is the diaphragm contributing to 25% to 75% of the tidal volume (Agostoni, 1964; Grimby *et al.*, 1968; Wade, 1954; Martin *et al.*, 1980; Bye *et al.*, 1984; Reid *et al.*, 1985). The lungs are also expanded by the elevation of the rib cage which acts to increase the anteroposterior diameter of the chest cavity. This action is affected by the movement of the sternocleidmastoid, scaleni, anterior serrati, and external intercostal muscles. At rest these accessory muscles are not usually functional. Rather, the diaphragm is solely responsible for inspiration

(Agostoni, 1964). Expiration is principally a passive action caused by the resultant elastic recoil of the lung and chest wall tissues. At rest, no expiratory muscles are involved, but during exercise the internal intercostals and abdominal recti serve to depress the rib cage and elevate the diaphragm, thus decreasing the volume of the chest cavity (Agostoni, 1964).

1.1 Respiratory Mechanics

Mechanical work is done by the respiratory muscles in order to counteract elastic, flow-resistive, and inertial forces (Fenn, 1951; Otis, 1964; Otis *et al.*, 1950). It is a function of lung volume and its derivatives ($f(V, \dot{V}, \ddot{V})$), where volume is associated with elastic forces, flow is associated with resistive forces, and acceleration is associated with inertial forces. Figure 1.1 illustrates the work components. Mathematically, the work of breathing can be represented as the integral of pressure with respect to changes in lung volume:

$$W = \int P \delta V. \qquad (1.1)$$

Hence, the work of breathing can be represented graphically via a pressure-volume diagram.

1.1.1 Elastic Work

A component of the work of inspiration is stored in the elastic structures of the system and thus is an available energy source during expiration. Work done against elastic forces will vary depending on the tidal volume and the compliance of the respiratory structures.

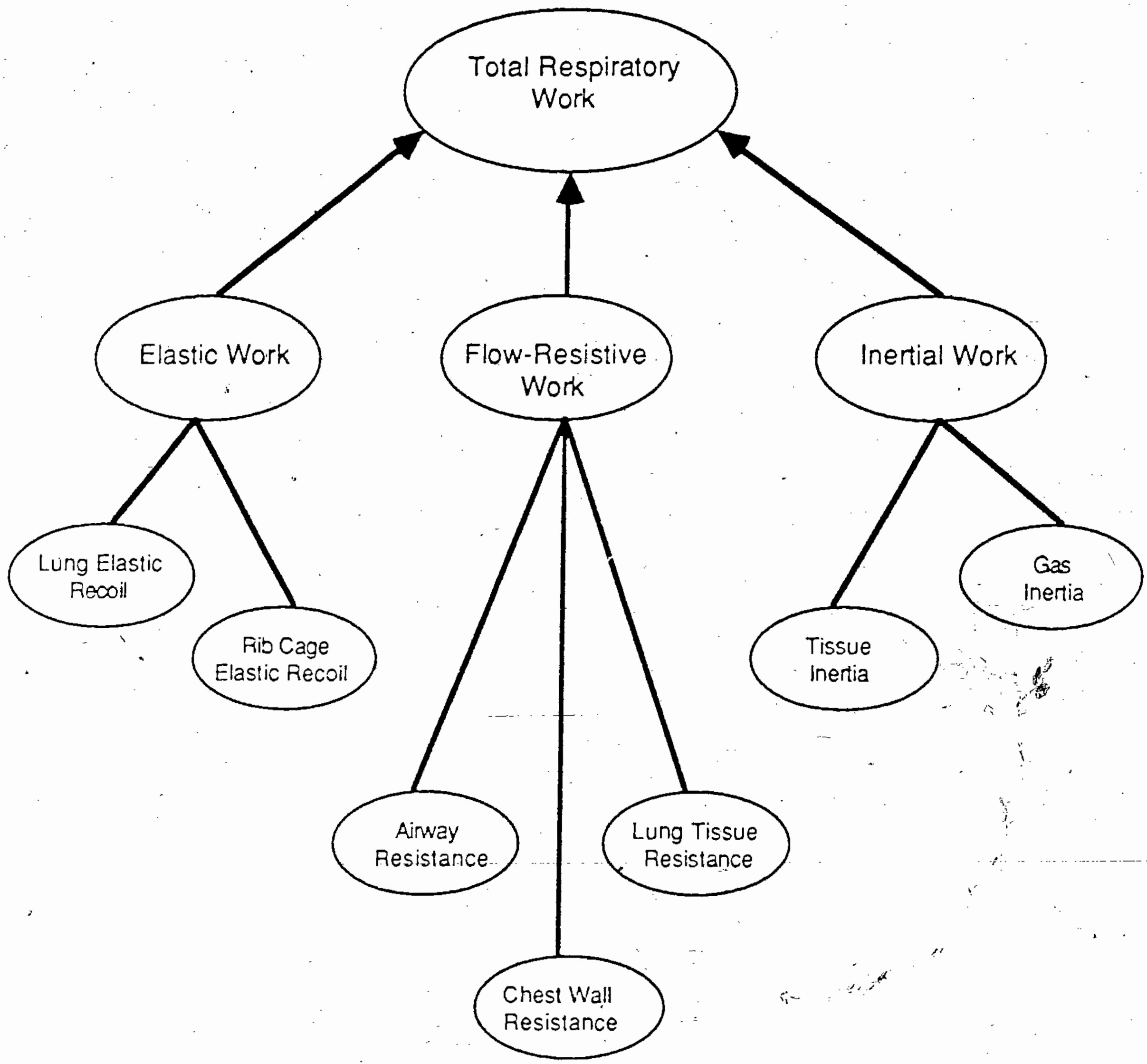


Figure 1.1: Mechanical work of respiration and its contributing forces.

1.1.2 Flow Resistive Work

Airway resistance and tissue resistance, as well as the rate of change of lung volume dictates the magnitude of flow resistive work (Fenn, 1951; Otis, 1964; Otis *et al.*, 1950). Airway resistance refers to the opposition to air movement created by friction which results in the loss of mechanical energy as heat (Taylor, 1987). Pulmonary resistance includes both airway resistance and frictional resistance to movement of the lung tissue. Total respiratory resistance is the sum of all resistive forces experienced during breathing, and is composed of pulmonary resistance and chest wall resistance.

Pulmonary air flow is predominantly laminar (Fenn, 1951) since turbulence is usually only developed in the lung when the air changes direction rapidly. Such a change could result from heavy exercise, or when the diameter of the airway is abruptly altered. The driving pressure, ΔP, relating to laminar flow can be calculated according to Poiseuille's Law which states:

$$\Delta P = (8 \cdot l \cdot \nu \cdot \eta) \cdot (981 \cdot r^4)^{-1} \tag{1.2}$$

where ΔP is the change in pressure (cmH_2O), l is the length of the tube (cm), ν is the velocity of air flow ($cm.s^{-1}$), η is the viscosity of the medium, and r is the radius of the tube (mm). When fluid velocity exceeds a critical value within the system, defined by Reynold's number, flow becomes turbulent. Reynold's number is given as the product of linear velocity, ν, gas density, γ, and airway diameter, d, all divided by the gas viscosity, μ:

$$\text{Reynold's Number, } Re = \gamma \cdot d \cdot \nu \cdot \mu^{-1} \tag{1.3}$$

At rest, the likelihood of exceeding the critical velocity in the airway is small due to low velocities and small diameters (Fenn, 1951). As Reynold's number will rise in proportion with gas density, respiratory airflow becomes more turbulent as barometric pressure rises. The result is an increase in airway resistance (Dahlback, 1975).

During turbulent flow the pressure difference is given by the equation

$$\Delta P = [(f \cdot l) \cdot (4\,\pi^2 \cdot r^5)^{-1}] \cdot \dot{V}^2 \qquad (1.4)$$

where f is a friction factor that depends on Reynold's number and the roughness of the airway walls, l is the length of the airway, r the radius of the airway, and $\dot{V}$ the volume rate of air flow (DuBois, 1964).

1.1.3 Inertial Forces

Inertial forces and the associated work have been neglected in experimental measures of respiratory work largely due to their negligible contributions to the cost of ventilation at normal breathing frequencies (Rohrer, 1925; DuBois, 1953; Mead, 1956; and Sharp *et al.*, 1964). Rohrer (1925) initiated the study of inertance, followed almost twenty-five years later by DuBois (1953). Inertance was studied in conditions of increased ambient pressures by Mead (1956). The results of his experiments on this topic indicated that inertance increased approximately in proportion to ambient pressure, up to a level of 4 ATA absolute pressure. The author concluded that the inertance measured was predominantly due to the gas stream as opposed to the lungs and the thorax.

The mechanical efficiency of breathing was calculated at 19% to 25% by Milic-Emili and Petit (1960). Margaria *et al.* (1960) further added that the mechanical efficiency of the respiratory muscles is the same as that of the muscles involved in performing useful external work. In general, the work of breathing is relatively small in magnitude with a cost of no more than 3% of the total metabolic rate (Otis and Cain, 1949; Otis *et al.*, 1950; Margaria *et al.*, 1960).

1.2 Respiratory Resistance

Resistance is the pressure drop across a system per unit rate of change of volume. When pressure is modelled as a combination of laminar and turbulent flow:

$$\Delta P = k_1 \cdot \dot{V} + k_2 \cdot \dot{V}^2, \qquad (1.5)$$

resistance can be modelled as

$$R = \Delta P \cdot \dot{V}^{-1} = k_1 + k_2 \cdot \dot{V}, \qquad (1.6)$$

where the intercept k_1 represents the viscous resistance in regions of laminar flow and the slope k_2 is dependent on system geometry in regions of turbulent flow.

1.3 Abbreviations and Definitions of Terms

1.3.1 Lung Volumes

The lungs can be divided into four unique pulmonary volumes, which, when added together, equal the maximum volume to which the lungs can be expanded.

1. *Tidal Volume (V_T).* The volume of air inspired or expired with each normal breath.
2. *Inspiratory Reserve Volume (IRV).* The amount of air that can be inspired above normal tidal inspiration.
3. *Expiratory Reserve Volume (ERV).* The amount of air that can be forcefully expired below normal tidal expiration.
4. *Residual Volume (RV).* The volume of air remaining in the lungs following a maximal expiratory effort.

1.3.2 Lung Capacities

Lung capacities are composed of two or more primary lung volumes.

1. *Functional Residual Capacity (FRC).* The volume of gas remaining in the lungs at the end of a normal expiration.

$$FRC = ERV + RV. \quad (1.7)$$

2. *Inspiratory Capacity (IC).* The volume of air that can be inspired after a normal expiration.

$$IC = V_T + IRV. \quad (1.8)$$

3. *Total Lung Capacity (TLC).* The volume of air contained in the lungs at the end of a maximal inspiration.

$$TLC = RV + ERV + V_T + IRV. \quad (1.9)$$

4. *Vital Capacity (VC).* The greatest volume of air that can be expelled from the lungs after maximum inspiration.

$$VC = ERV + V_T + IRV. \quad (1.10)$$

1.3.3 Respiratory Pressures

All pressures and components of respiratory work and flow resistance are expressed relative to an anatomical location or dimension by use of the following subscript abbreviations: airway (aw); airway opening (ao); mouth (m); alveolar (alv); oesophagus (oes); pleural (pl); lung (l); lung tissue(lt); body surface (bs); total respiratory system (rs); chest wall (cw); transpulmonary (tp); transrespiratory (trs); and transthoracic (tth). Transpulmonary pressure, P_{tp}, refers to the pressure difference across the lungs. P_{bs} refers to the mean hydrostatic or atmospheric pressure acting on the surface of the thorax. P_{ao} refers to the pressure within the airway at the mouth, and is measured within the mouth piece of the breathing loop. P_m refers to the external pressure (ie. atmospheric pressure) at the mouth. Figure 1.2 is a schematic diagram representing these physiological pressures.

The following expressions define pressure inter-relationships. It should be noted that for measurement purposes, it is assumed that $P_{oes} = P_{pl}$.

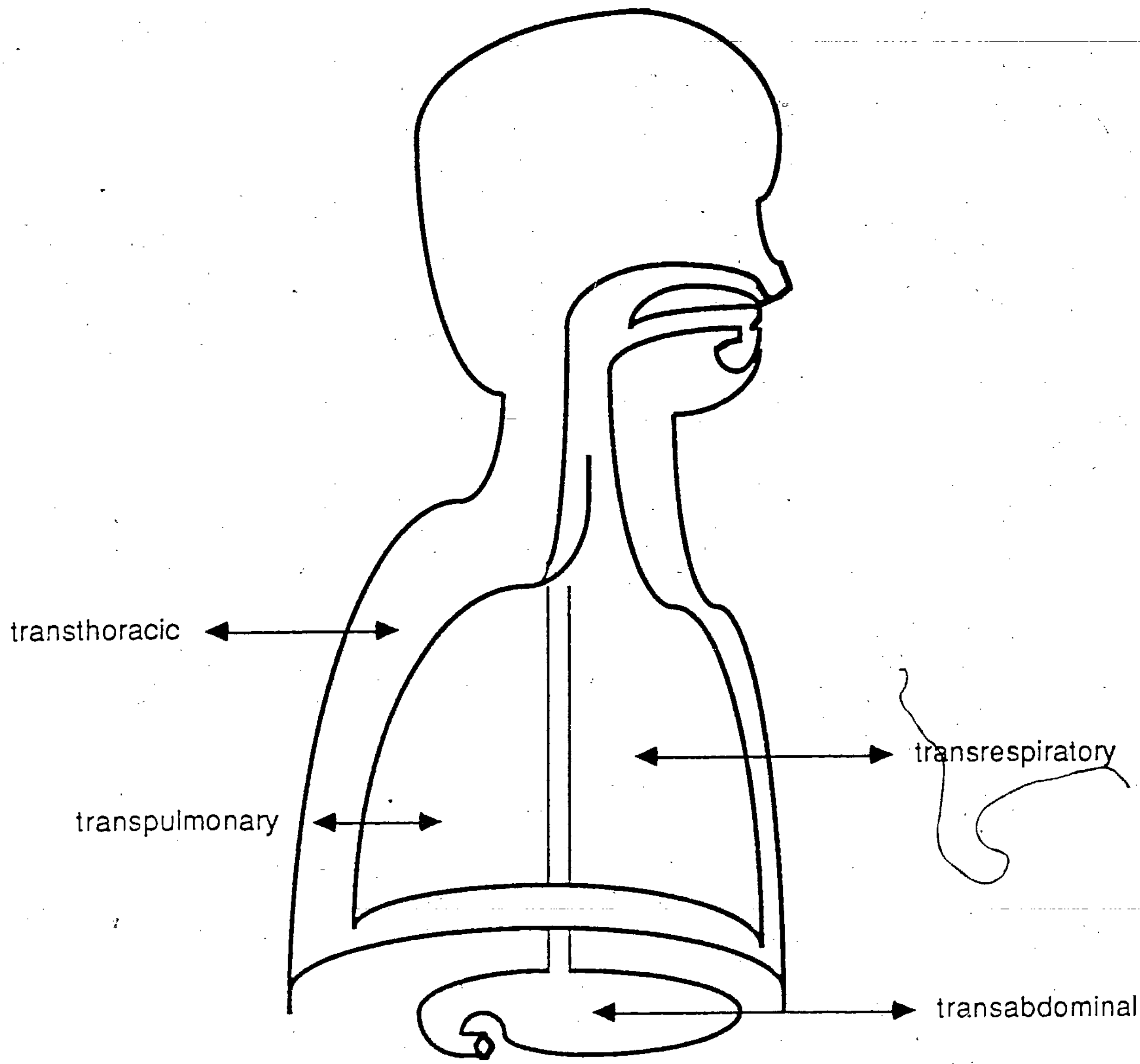


Figure 1.2: Schematic interrelationships between respiratory pressure measurements.

$$P_{trs} = P_{tp} + P_{tth} \quad (1.11)$$

$$P_{trs} = (P_{alv} - P_{pl}) + (P_{pl} - P_{bs})$$

$$P_{trs} = P_{alv} - P_{bs}$$

$$P_{tp} = P_{alv} - P_{pl} \quad (1.12)$$

$$P_{tth} = P_{pl} - P_{bs} \quad (1.13)$$

1.3.4 Resistance

1. *Airway resistance (R_{aw}).* The opposition to air movement created by friction, and resulting in loss of mechanical energy as heat. Airway resistance will vary with lung volume when airflow is laminar. When airflow is turbulent, airway resistance will also vary with the flow rate.

2. *Pulmonary Resistance (R_{pul}).* Frictional resistance is imposed by movement of lung tissue moving across itself (R_{lt}). Pulmonary resistance is the summation of airway and lung tissue resistances.

$$R_{pul} = R_{aw} + R_{lt} \quad (1.14)$$

3. *Respiratory Resistance (R_{rs}).* Frictional resistance is imposed by the movement of the chest wall, R_w. Respiratory resistance is the sum of all resistive forces experienced during breathing.

$$R_{rs} = R_{aw} + R_{lt} + R_w \quad (1.15)$$

1.3.5 Work

1. *Elastic Work.* The amount of energy required to expand the lungs against resistive forces.
2. *Tissue Resistance Work.* The amount of energy required to overcome the viscosity of the lungs and chest wall.
3. *Airway Resistance Work.* The amount of energy required to overcome the resistance to airflow through the respiratory passageways.
4. *Total Respiratory Work.* The amount of work required to overcome all respiratory forces, including elastic, resistive, and inertial forces.

1.3.6 Power

The time rate at which energy is transferred into respiratory work.

1.3.7 Miscellaneous Terms

1. *Lung Centroid (P_{LC}).* Derived from the phrase *'center of pressure of thorax'*, and used initially by Paton and Sand (1947). Lung centroid pressure is defined as the pressure required to return the immersed lung relaxation volume to the level that exists in air. The centroid is a spatial location within the thorax which represents the position of the mean hydrostatic pressure acting on the outside of the thorax during immersion. In air, pressure at this point would be atmospheric pressure. During immersion, with an occluded airway, it would equal the hydrostatic pressure found at some point between the apex and the base of the lungs.
2. *Relaxation Pressure ($P_{R(oc)}$).* Intrathoracic pressure obtained during complete respiratory relaxation against an occluded airway, with glottis open.
3. *Relaxation Volume (V_R).* Lung volume obtained during complete respiratory relaxation with glottis and airway open. The airway

may or may not be open to the atmosphere. In the case of immersion, when breathing from self-contained underwater breathing apparatus, the airway is generally not equilibriated to atmospheric pressure, but to the delivery pressure of the apparatus. When the airway is open to the atmosphere, while seated in air, the relaxation volume is usually equivalent to the functional residual capacity.

CHAPTER 2

REVIEW OF LITERATURE

Respiratory work has been the topic of study for several years. As early as 1915, Rohrer developed an explanatory equation relating respiratory pressure change to rate of volume change:

$$\Delta P = k_1 \cdot \dot{V} + k_2 \cdot \dot{V}^2. \qquad (3.1)$$

Although this empirical equation is an over-simplification of the problem, it has served well for rough prediction purposes (Lanphier and Camporesi, 1982). The two components acting in series are modelled to represent laminar flow and turbulent flow respectively. The values of k_1 and k_2 were determined by Rohrer (1915) both to be 0.8, with ΔP measured in cm.H_2O, and $\dot{V}$ in L.sec^{-1}.

At one time, it was thought that the first coefficient depended on viscosity alone and the second coefficient on density alone. This assumption has been shown to be incorrect. Maio and Farhi (1967) found that changes in gas density influenced ΔP even at low levels of flow. Wood and Bryan (1969) noted the importance of pressure change in proportion to both density and $\dot{V}^2$.

It is possible that a transition from laminar flow to turbulent flow exists, affecting the relative contribution of the two modelled components. DuBois (1964) suggested that since Reynold's number is increased with raised gas density and/or flow rate, flow becomes turbulent in some parts of the airways that were previously laminar. The result would be a decreasing contribution of the laminar component and an increasing contribution of the turbulent component.

2.1 Measurement of the Work of Breathing

Measurement of the work of breathing, partitioned into components, has been investigated extensively under normal environmental conditions. The determination of the elastic work of breathing was initially accredited to Romanoff (1910–1911). Further investigations were conducted by Rohrer (1916) and Rahn *et al.* (1946). Rahn and his associates were the first recognized for the production of respiratory pressure–volume diagrams. The elastic recoil of the respiratory system under static conditions was measured from these diagrams.

The elasticity of the respiratory system may be divided into distinct sections (Agostoni and Rahn, 1960; Mead and Milic–Emili, 1964):

1. Parallel elasticity of the rib cage and abdomen–diaphragm within the chest wall.
2. Series elasticity of the chest wall and lung tissue.

Lung tissue elasticity is represented by transpulmonary pressure change while chest wall elasticity is obtained via transthoracic pressure change. (Refer to Figure 1.2 for transpulmonary and transthoracic pressures.) Figure 2.1 represents schematically the components of respiratory elasticity. The sigmoidal shape of the pressure–volume curve of the total respiratory apparatus results from the summation of chest wall and lung tissue curves. Figure 2.2 displays the classical sigmoidal shape.

Otis *et al.* (1950) reported the contribution of elastic work, at rest, to the total work of respiration to be 63% with the remaining 37% accredited to flow resistive work. Attinger and Segal (1959) reported an almost identical value of flow resistive work at 38% of the total. McIlroy *et al.* (1954) reported elastic work to contribute 70% to the entire work of breathing performed at rest. Since the contribution of inertial forces to respiratory work are considered negligible, the remaining 30% to 40% of total respiratory work may be attributed to work against flow resistive forces. McIlroy *et al.* (1955) established lung tissue resistance to contribute 35% to total resistance. When considering only the work done against flow resistive forces, the airway resistance component has been shown to be the major contributor to the

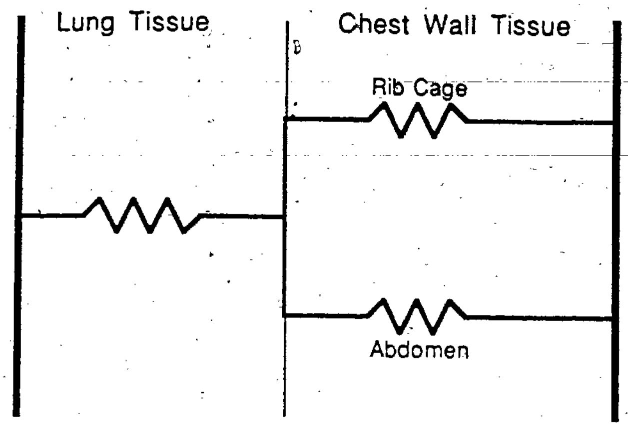


Figure 2.1: Schematic representation of the components of thoracic elasticity.

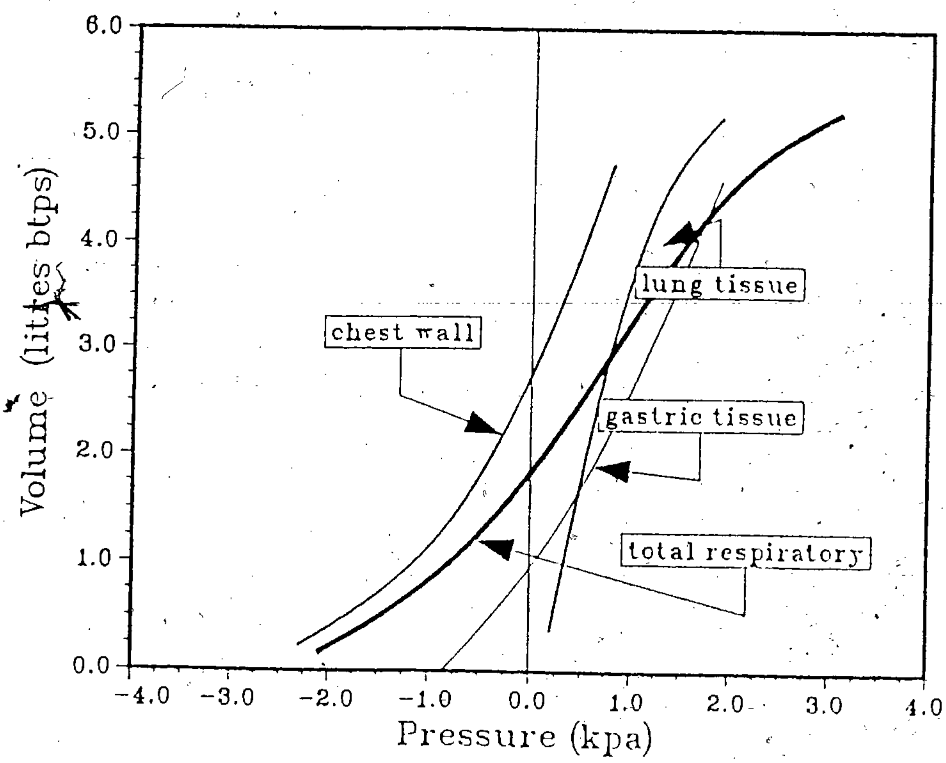


Figure 2.2: The pressure–volume diagram of the chest wall and lungs in healthy man: the relaxation pressure curve.

overall total, in the neighbourhood of 60% to 80% (Otis *et al.*, 1950; Marshall and Dubois, 1956; Ferris *et al.*, 1964; Gauthier *et al.*, 1982). Flow mechanics and its inherent variables would largely dictate the dynamic work of respiration, where the flow resistive work increases non-linearly with minute ventilation (Otis *et al.*, 1950; Fritts *et al.*, 1959; Holmgren *et al.*, 1973). Table 2.1 summarizes the findings of the presented studies regarding the relative contribution of the components to the work of the work of breathing. Table 2.2 summarizes the literature regarding the relative contribution of resistive force compononents to the total.

Work of breathing was also calculated at exercise. McIlroy and his co-workers observed that the increase in the work of breathing during exercise is dependent on two factors. First, there is a relative change in the rate and depth of respiration as minute volume is increased. Second, there is an exercise effect on the magnitude of the resistances to lung movement. A mean value for work of breathing was not calculated as steady-state was not achieved by the subjects.

2.2 The Immersed Environment

The effect of immersion on respiratory and cardiorespiratory functions has been studied extensively. Most experimental procedures involve measurements made with the subject immersed to the neck in thermally neutral water (Agostoni *et al.*,1966; Craig and Ware, 1967; Hong et al., 1969; Arborelius et al., 1972; Begin *et al.*, 1976). Control measurements were often taken with water at the level of the xiphoid process (Hong *et al.*, 1969; Farhi and Linnarsson, 1977). The effects of immersion on respiration are numerous. Several common findings were established through the various experiments. In general, immersion tends to:

1. Decrease rate of ventilation at a given rate of oxygen exchange (Arborelius *et al.*, 1972; Thalmann *et al.*, 1979; Grismer and Goodwin, 1983).
2. Increase alveolar ventilation at a given rate of ventilation (Begin *et al.*, 1976; Thalmann *et al.*, 1979).

Table 2.1: **Contribution of work components to respiratory work at rest: literature.**

SOURCE	ELASTIC WORK	RESISTIVE WORK
Otis *et al.*,1950	63%	37%
McIlroy *et al.*, 1954	70%	—
Attinger & Segal, 1959		38%

Table 2.2: **Contribution of airway, lung tissue and chest wall resistive forces to total flow-resistive forces: literature.**

SOURCE	AIRWAY RESISTANCE	LUNG TISSUE RESISTANCE	CHEST WALL RESISTANCE
Otis *et al.*, 1950	77.6%	22.3[1]	
McIlroy *et al.*, 1955		35%	
Marshall and Dubois, 1956	86%	13.7%	
Ferris *et al.*, 1964	60%	1%	39%
Gauthier *et al.*, 1982	82%	18%	

[1]Combined chest wall and lung tissue resistance.

3. Decrease vital capacity (Agostoni *et al.*, 1966; Craig and Ware, 1967; Hong *et al.*, 1969; Robertson *et al.*, 1978).

4. Increase blood flow to the apical region of the lungs and decrease blood flow to the basal region of the lungs (Arborelius *et al.*, 1972).

5. Decrease expiratory reserve volume (Agostoni *et al.*, 1966; Hong *et al.*, 1969; Flynn *et al.*, 1975; Dahlback *et al.*, 1978; Robertson *et al.*, 1978; Taylor and Morrison, 1989).

6. Increase airway resistance (Agostoni *et al.*, 1966) and pulmonary resistance (Morrison *et al.*, 1987).

7. Increase the total work of breathing, resulting mostly from an increase in elastic work (Hong *et al.*, 1969; Sterk, 1973; Dahlback *et al.*, 1978; Taylor and Morrison, 1989).

8. Increase pulmonary capillary blood flow (Begin *et al.*, 1976; Farhi and Linnarsson, 1977).

9. Decrease functional residual capacity (Begin *et al.*, 1976; Farhi and Linnarsson, 1977; Robertson *et al.*, 1978; Grismer and Goodwin, 1983; Taylor, 1987).

10. Increase pulmonary air trapping (Dahlback and Lundgren, 1972).

11. Decreased maximum voluntary ventilation (Flynn *et al.*, 1975).

12. Increase heart rate (Krasney *et al.*, 1984).

13. Increase cardiac output (Farhi and Linnarsson, 1977; Krasney *et al.*, 1984).

14. Shift the relaxation pressure-volume curve to the right (Jarrett, 1965; Hong *et al.*, 1969; McKenna *et al.*, 1973; Flynn *et al.*, 1975; Minh *et al.*, 1979; Taylor, 1987; Morrison and Taylor, 1988).

Full body immersion in water exposes the subject's lungs to a change in hydrostatic pressure due to the hydrostatic gradient acting on the body surface. This hydrostatic gradient imposes an imbalance between the thoracic surface pressure and the airway opening (mouth pressure). Morrison and Reimers (1982) indicated that the division of work between inspiration and expiration was altered with immersion due to the effect of a hydrostatic pressure imbalance between the pressure of the breathing gas supplied (usually at the level of the mouth) and the mean hydrostatic pressure acting on the surface of the thorax. The greater the imbalance, the more the subject will exhale below the normal end expiratory volume in an attempt to equilibriate the external pressure acting on the thorax by readjustment of the elastic tissues of the lung-chest wall system.

An investigation involving the work of breathing when in an immersed environment was conducted by Hong *et al.* (1969). Four subjects were immersed in the seated position at two distinct levels, the xiphoid process and the neck. The total work of inspiration rose 64.6% with immersion to the neck, largely due to an increase in elastic work. Other studies, using a variety of methods but all incorporating the use of oesophageal balloons to collect pressure data, were performed to determine the changes in flow resistance imposed by upright immersion. Agostoni *et al.* (1966) observed an increase by 57.7% in the airway resistance with immersion to the xiphoid process. Sterk (1970, 1973), using a dynamic transpulmonary pressure technique, deduced the pulmonary resistance to rise between 185% and 243%. Dahlback (1978) and Dahlback *et al.* (1979) reported very different increments in pulmonary flow resistance with immersion, at 31% and 42.5% respectively. Lollgen *et al.* (1980), who employed the oscillation method to derive total respiratory resistance, found resistance to increase by 57.4% with immersion. A recent investigation by Taylor (1987) observed that when immersed, inspiratory, expiratory, and total pulmonary resistances were elevated to at least double those in dry conditions. Breathing air supplied at lung centroid position rather than at mouth pressure restored normal lung subdivisions, returned elastic and flow resistive work towards normal, and allowed greater work load tolerance with less respiratory distress. Table 2.3 summarizes the literature findings.

Table 2.3: **Effects of immersion on components of respiratory work and resistance: literature.**

SOURCE	INSPIRATORY FLOW RESISTIVE WORK	AIRWAY RESISTANCE	PULMONARY RESISTANCE	TOTAL RESISTANCE
Agostoni *et al.*, 1966		+57.7% [1]		
Hong *et al.*, 1969	+64.6%[2]			
Sterk, 1970;1973			+185–243%	
Dahlback, 1978			+31%	
Dahlback *et al.*, 1979			+42.5%	
Lollgen *et al.*, 1980				+57.4%
Taylor, 1989	+301%[3]		+220%[3]	

[1]Immersion to the xiphoid process.
[2]Increase in work from immersion at the xiphoid process to immersion at the neck.
[3]Total body immersion.

2.3 The Hyperbaric Environment

Wood and Bryan (1969) measured maximum expiratory flow at six lung volumes between 1 and 10 ATA. At larger lung volumes, the maximum expiratory flow varied with density; at lower lung volumes (less than 25% vital capacity), flow tended to be less density dependent. The authors concluded that the maximum breathing capacity and peak expiratory flow varied inversely and exponentially with the density of air.

Salzano *et al.* (1970) had subjects perform graded exercise tasks on a bicycle ergometer at varying ambient pressures, up to 31.3 ATA in a dry compression chamber. The gas mixture breathed was composed of 99.1% helium and 0.9% oxygen, with a resultant density of the mixture of over four times that of air at sea level. Work at depth was performed with increases in oxygen consumption, V_{O_2}, and tidal volume, as well as decreases in heart rate and respiratory rate, when compared to the same work rate at sea level. Increased gas density was thought to be the factor in most of the altered responses.

Pulmonary function in divers "living" at 49.5 ATA was monitored by Spaur *et al.* (1977). Ventilatory function was reduced, as expected, with a decrease in the maximum voluntary ventilation by 45%. An increase in functional residual capacity, *FRC*, and transpulmonary pressure was observed. During underwater work periods, the subjects wore closed circuit breathing apparatus while pedalling a cycle ergometer. The underwater work led to severe dyspnea. Physiological adjustments made were observed to be similar to those made in asthma sufferers. For this reason, the dyspnea resulting at high ambient pressures was concluded to be mechanical rather than chemical in nature.

Wood and Bryan (1978) tested two subjects with a graded exercise regime on a bicycle ergometer at five ambient pressures ranging from 1 to 10 ATA (dry environment). An oesophageal balloon was used to record intrapleural pressures. Maximal testing (900 kg.m.min^{-1}) had to be discontinued at ambient pressures above 4 ATA due to severe dyspnea. The reduced aerobic capacity was related to the limitation of expiratory flow, due to dynamic airway compression. The authors suggested that the decrease observed in the maximum expiratory flow at depth was the direct result of the raised resistance caused by an increased gas density. Analogous to Spaur *et al.* (1977), Wood and Bryan observed a resemblance between healthy subjects breathing dense air and patients with obstructive lung disease.

Similar findings were observed by Thalmann *et al.* (1979) when three divers performed submerged exercise in the prone position, at depth. The purpose of the study was to investigate the effects of static lung loading and increased gas density on the submerged exercising subject breathing air. Dry control testing was executed at 1.45 ATA; immersed testing occurred at ambient pressures ranging from 1.45 ATA to 6.76 ATA. With immersion, the divers were subjected to a hydrostatic gradient effect and thus increased gas density could not be concluded as the only factor for increased respiratory resistance. Dyspnea proved to be the limiting factor at maximum work rate occurring during maximal exercise at 6.76 ATA with some but not all static lung loads. At a load of +10 cm.H_2O, less dyspnea resulted and subjects were able able to perform maximally at this density level for five minutes. The only parameter that correlated directly with the severity of the dyspnea was maximum oxygen consumption, V_{O_2} *max.*. Thalmann *et al.* (1979) noted that work-limiting dyspnea was observed at lower levels of oxygen consumption in the studies of Dwyer *et al.* (1977) and Spaur *et al.* (1977). The two earlier studies were conducted at greater ambient pressures (43.4 ATA and 49.5 ATA, respectively), which would induce a seven-fold increase in breathing gas density. Hence, Thalmann and his co-workers concluded dyspnea to be primarily related to hydrostatic imbalance.

2.4 Mechanical Ventilation as a Research Technique

Otis *et al.* (1950) were the first to develop the technique of mechanical ventilation as a research method, to measure volume changes in the chest in response to changes in pressures at normal breathing frequencies. The subjects voluntarily relaxed and allowed a sinusoidal pump to ventilate them. A Drinker Respirator enclosed the subjects to the level of their necks while in the supine position. The pressure within the tank was subsequently altered, while the pressure gradient between the respirator and the mouth of the subjects was recorded. The elasticity of the lungs and chest wall, as well as the resistance to breathing, acted as the focus of analysis of the pressure-volume data. Inertia was calculated, but found to be negligible at the forcing frequencies employed. (Values calculated for work against elastic and resistive forces have been given earlier in this chapter. Refer to table 2.1.) From the experimental data, the authors derived equations that give an approximate description of the mechanics of human respiratory apparatus.

DuBois and Ross (1951) employed the Drinker Respirator apparatus at higher frequencies in an attempt to highlight the inertial factor. DuBois continued research in this area with his co-workers, and in 1956 reported the frequency response characteristics of the airways, lungs, and chest at oscillations in the range of 2 Hz to 15 Hz. The motion of the chest wall was measured through the creation of an electromagnetic field while the surface motion of the abdomen was also monitored under the assumption that if the diaphragm moved distally, the abdominal wall moved outward. The authors noted that air flow of the normal respiratory pattern was not necessarily sinusoidal, and hence could effect the data obtained.

The total work of breathing in obese men was assessed by Sharp *et al.* (1964) using a method incorporating a tank respirator, similar to the apparatus of Otis *et al.* (1950). Sharp and his co-workers noted that the validity of the method depended on the subject's ability to relax the respiratory muscles and to allow the tank respirator to perform the respiratory work. To test for complete subject relaxation throughout the procedure, they employed independent methods for measuring compliance and total resistance. Twenty-two subjects participated, fourteen of whom were obese and eight of whom were from the normal population. Tidal volumes, flows, and tank respirator pressures were recorded. Three breathing frequencies were used: 12, 20, and 30 breaths per minute, and the pressure in the tank was varied between $\underline{+}70$ cmH_2O and -70 cmH_2O. The resultant values of compliance and resistance were compared to the values obtained from the independent measures. Table 2.3 summarizes these values.

With respect to compliance, the values obtained by the two methods were very similar, and thus highly correlated. With respect to resistance, the tank respirator values were two to three times the values obtained using an independent oscillation technique (control method #2). The difference in volume displacement employed by the two methods was of the magnitude of twenty times. Such an increase in the volume change would imply greater passive stretching of the respiratory tissues with the respirator technique, possibly producing different tissue resistances. Total respiratory work in kg.m.L^{-1}, at a respiratory rate of twenty breaths per minute, averaged 0.073 in normal subjects, 0.095 in obese subjects, and 0.212 in obese subjects with obesity hypoventilation syndrome. The difference in values between the normal group and the obese groups were significant ($p<0.05$).

Table 2.3: **Comparison of data of two measurement techniques for normal (N) and obese(O) men. (Sharp** *et al.*, 1964)

	TOTAL COMPLIANCE $L.cmH_2O^{-1}$		TOTAL INSPIRATORY RESISTANCE $cmH_2O.L^{-1}.sec$	
	Tank Respirator	Control Method #1	Tank Respirator	Control Method #2
N:	0.104 ± 0.005	0.109 ± 0.009	4.80 ± 0.50	1.90 ± 0.07
O:	0.081 ± 0.007	0.072 ± 0.010	7.70 ± 0.72	3.10 ± 0.35

2.5 Objectives

The objectives of this study were

1. To investigate the technique of artificial respiration in the measurement of total respiratory work and respiratory resistance.
2. To measure elastic and resistive components of respiratory work under dry conditions, during immersion with and without hydrostatic pressure compensation of breathing gas, and at raised ambient pressure.
3. To determine the relationship between the energy cost of ventilation ($J.L^{-1}$) and the rate of ventilation ($L.min^{-1}$) in each condition.

2.6 Hypotheses

It was hypothesized that:

1. Upright immersion to the neck without hydrostatic pressure compensation of breathing gas, decreases lung relaxation volume and expiratory reserve volume, *ERV*. This effect will decrease the total respiratory compliance and increase airway resistance, causing an increase in the elastic and flow resistive work of breathing.
2. Provision of hydrostatic pressure compensation of breathing gas to lung centroid pressure (equal to the mean external pressure acting on the thorax) will return total respiratory compliance and airway resistance towards normal, resulting in reduced respiratory work during immersion.
3. At increased ambient pressure, an increase in respired gas density will cause an increase in the work of breathing due to the turbulent nature of airflow in the airways.

CHAPTER 3

METHODS

The most commonly used method for the determination of work of breathing involves the use of an oesophageal balloon to estimate intrapleural pressure. This technique is invasive, requires careful calibration, and involves a certain degree of discomfort to the subject. The validity of this technique exists on the assumption that oesphageal pressure closely approximates mean pleural pressure. It does not measure the total work of respiration as it omits the flow-resistive and elastic forces exerted by the chest wall. Hence, the oesophageal technique estimates only the pulmonary elastic and pulmonary flow-resistive components of respiratory work.

Mechanical ventilation has been used infrequently to measure respiratory work, although widely used for clinical purposes in the medical setting. In the latter situation, mechanical ventilation provides gas exchange that adequately supplies the tissues with oxygen when a patient's lungs can no longer do so independently (Chalikion and Weaver, 1984). Although the mechanical ventilation technique measures the total work of the respiratory system, it relies on the assumption that the subject can maintain total relaxation of the respiratory muscles during passive ventilation (Sharp *et al.*, 1964). Hence, it may be said that both techniques are subject to a source of potential error which may be either random or systematic in nature.

3.1 Apparatus

All experimental procedures took place at the Environmental Physiology Unit, in the School of Kinesiology at Simon Fraser University. The hypo/hyperbaric chamber was the site of mechanical ventilation trials, while the general laboratory area was used for lung function tests.

3.1.1 Hypo/Hyperbaric Chamber

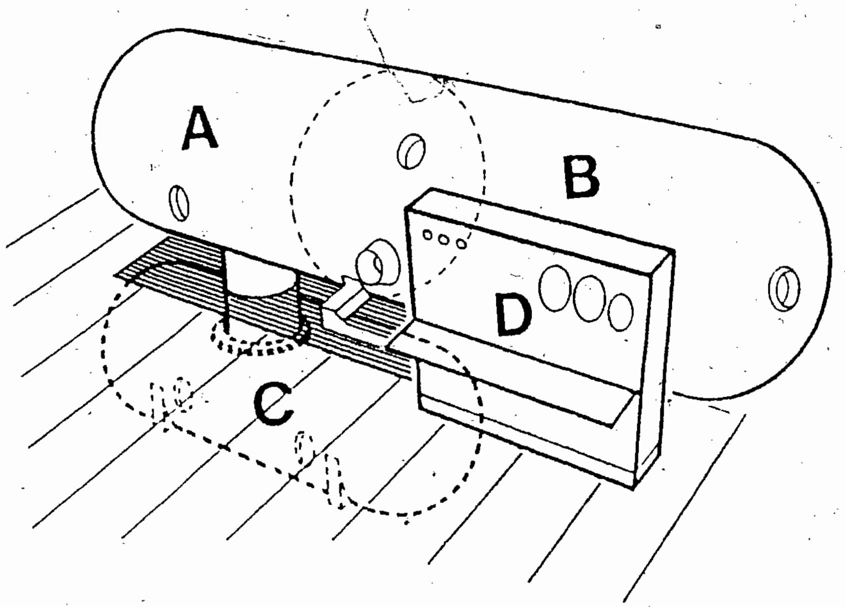
The hypo/hyperbaric chamber consists of three locks; the entry lock, the main lock, and the wet lock. The wet lock is situated below the entry lock, and may be filled with water for experimental purposes. Figure 3.1 illustrates the three locks of the hypo/hyperbaric chamber. Verbal communication ports are wired for both the entry and main locks, while visual contact with all three locks is maintained at the control consul via television cameras. The chamber operators observe the experimental procedures on a video monitor connected to the three cameras. Emergency oxygen is available in the main lock and emergency air is available in all three locks.

3.1.2 Mechanical Respirator

The mechanical respirator consists of three major components in series. An electric motor, controlled by a radiotrol variable frequency controller, powers a hydraulic pump. In turn, the hydraulic pump powers a hydraulic motor which drives the respirator at the selected frequency when the system is engaged. The electric motor, controller, hydraulic pump and reservoir are situated outside the chamber and connected through the chamber wall via hydraulic connectors to the respirator within. Figure 3.2 illustrates the serial components of the respirator system, while Figure 3.3 depicts the breathing machine alone.

Air is pumped with a sinusoidal flow pattern through a combination of rigid and flexible tubing to the mouthpiece and back to the point of origin. One-way valves ensure unidirectional airflow to and from the subject. Between periods of mechanical ventilation, the subject breathed via a valve which connected the breathing circuit to either a demand regulator or the air within the wet lock. During trials, the valve would be shut allowing the subject to be ventilated from the closed circuit of the respirator. Figure 3.4 depicts the breathing circuit.

The changes in volume of the respirator were measured by a rectilinear potentiometer model HLP190 type FS (Llybrid Technology), and amplified by a Daytronic model 9010 mainframe system with a model 9163 analog input module. Pressure changes at the mouthpiece, P_{ao}, were monitored by a Validyne model



A — ENTRY LOCK
B — MAIN LOCK
C — WET LOCK
D — CONTROL CONSOLE

Figure 3.1: The hypo/hyperbaric chamber: entry lock, main lock, and wet lock.

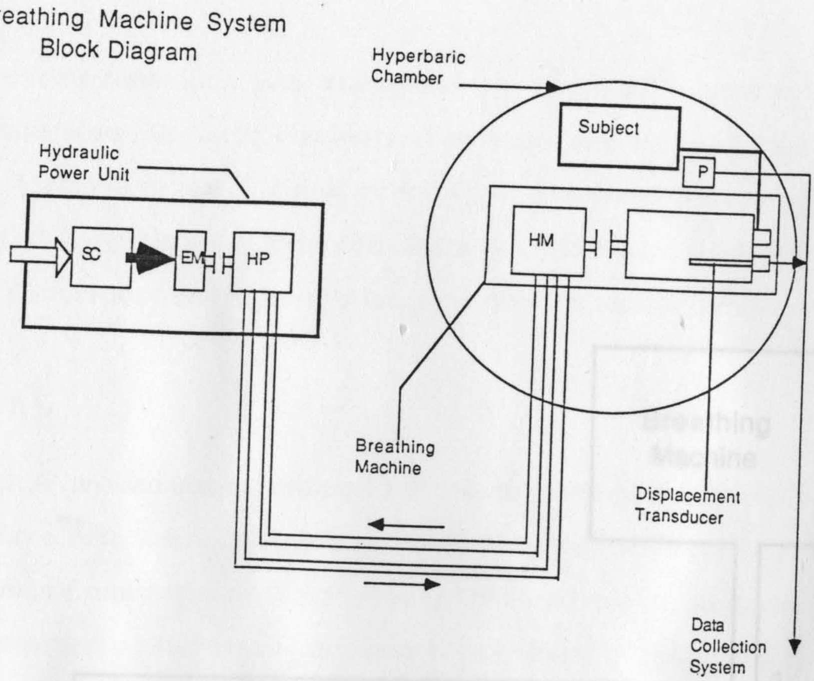


Figure 3.2: Serial components of the respirator system. (Redrawn from the work of G. Morariu, 1983).

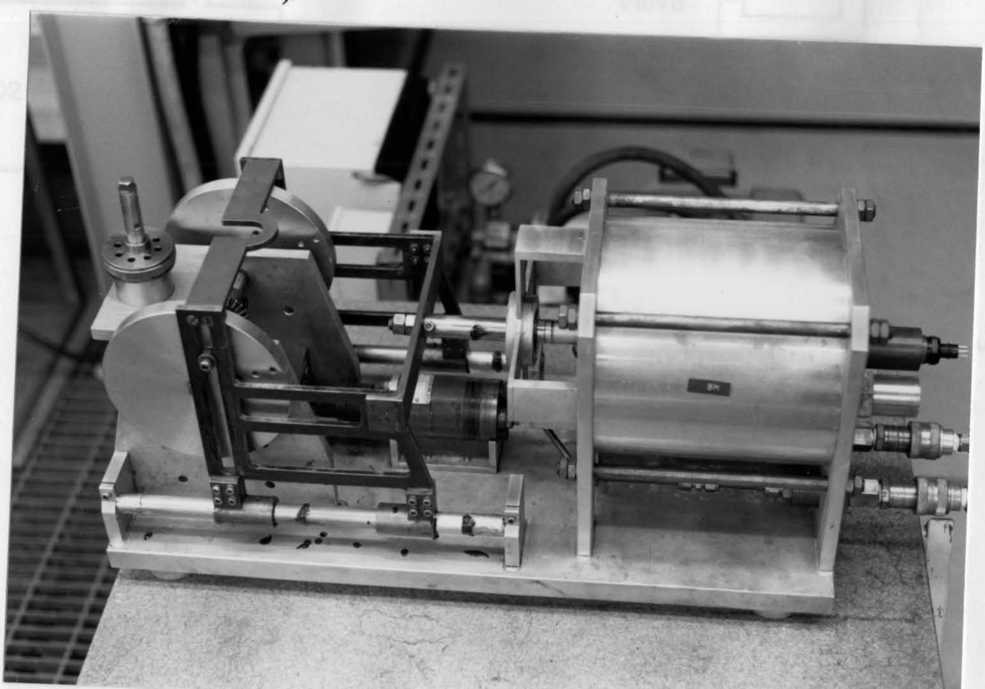


Figure 3.3: The breathing machine (Designed by G. Morariu, 1983).

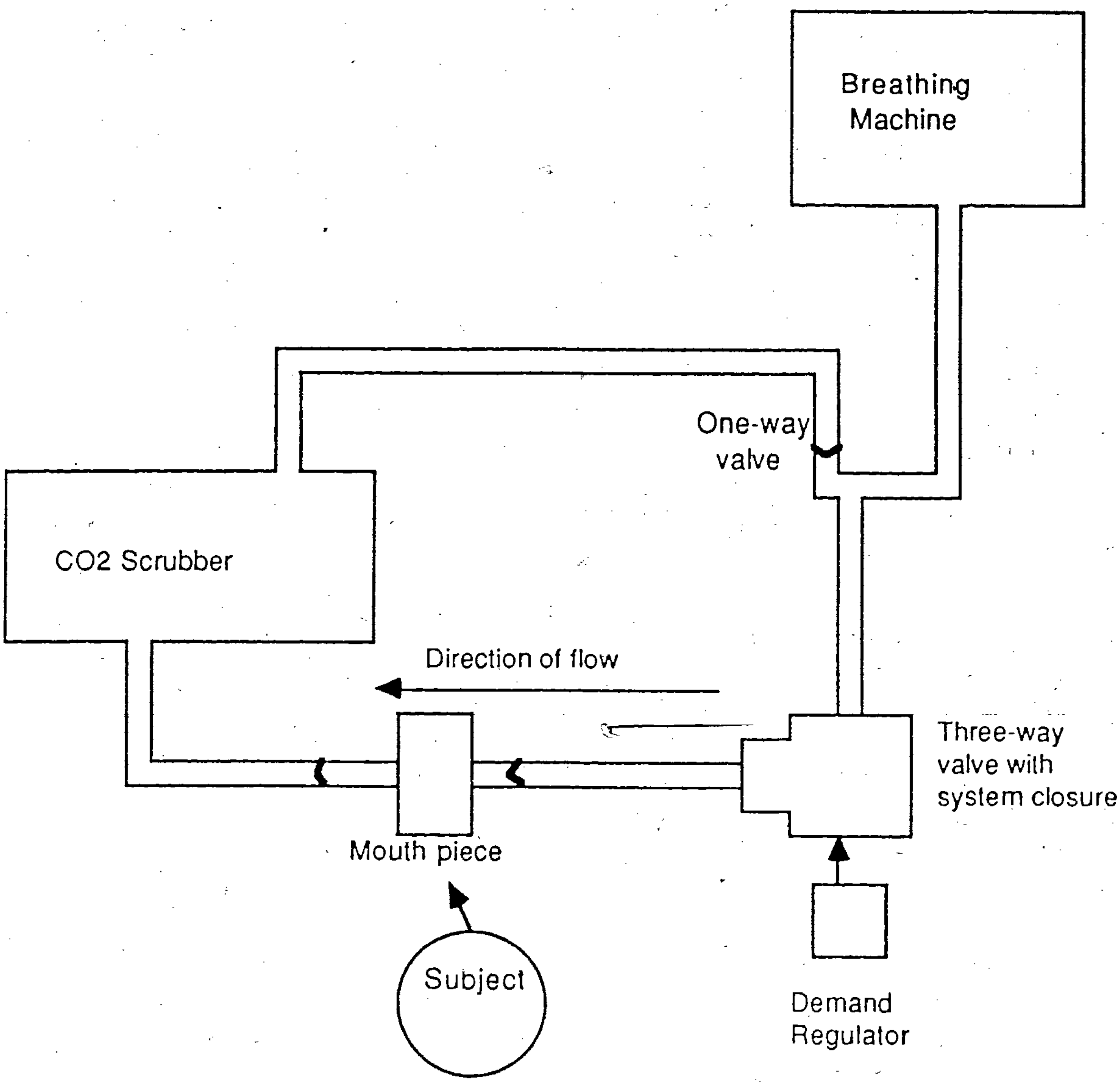


Figure 3.4: Breathing circuit within the wet lock of the hypo/hyperbaric chamber.

DP215-52 pressure transducer and amplified by a model 9130 input module on the Daytronic mainframe. Both the volume and pressure signals were low pass filtered at 5 Hz by a Rockland model 432 dual HI/LO filter (Rockland Systems Corporation). Once filtered, the signals were recorded digitally at 50 Hz by a Tecmar A/D converter and an IBM PC microcomputer, controlled by a data collection program.

3.2 Procedures

The experimental procedures developed for the collection of respiratory pressure-volume data were approved by the Environmental Physiology Unit and the University Ethics Committee at Simon Fraser University, prior to commencement of the study. Informed consent was obtained by all subjects following an information session regarding the approved protocol.

3.2.1 Subject Selection Procedure

Subjects selected to participate in the research experiment were required to meet certain criteria:

1. Participants of the study were required to be certified and experienced as divers.
2. Potential candidates performed respiratory maneuvers to produce static compliance curves. Repeatable curves indicated that the subject was able to relax with an open glottis against an occluded airway. Respiratory relaxation with an open glottis would allow for mechanical ventilation by the breathing machine.
3. Subjects selected to participate were medically examined and questioned for good health by an attending physician. Identification of acute or chronic pulmonary, cardiac, respiratory, or neurologic conditions or diseases necessitated exclusion from the study.

Five male subjects were selected. All were experienced in pulmonary function testing procedures in addition to meeting specified criteria. The subjects ranged in age from 29 to 45 years, with a mean age of 35 years. Table 3.1 provides anthropometric information for each subject.

3.2.2 *Experimental Protocol*

The technique used to mechanically ventilate the participants was fashioned after the method developed by Otis *et al.* in 1950. The technique is based on the integration of respiratory driving pressure with respect to lung volume change. The subjects were trained to relax and permit a respirator to passively ventilate the lungs. A sinusoidal pump having a controlled tidal volume and frequency was employed to measure respiratory work, dynamic compliance and flow resistance using this technique. Inspiratory and expiratory phases shared equal timing due to the sinusoidal wave form. The rate of respiration was randomly altered in successive trials,

$$10\ bpm < f_b < 30\ bpm,$$

to produce ventilatory increments within the normal range of

$$20\ L.min^{-1} < V_E < 60\ L.min^{-1}.$$

The stroke volume of the respirator remained at a fixed value of approximately 2.0 liters. Carbon dioxide was scrubbed from the system to prevent hypercapnia causing the initiation of voluntary ventilatory effort. Simultaneous measurement of the lung volume and pressure difference between the driving pressure at the mouth or airway opening and the system relaxation pressure (ie. breathing gas supply pressure) allowed the construction of the pressure–volume diagram and hence the analysis of respiratory work.

Measurements were performed in the upright seated position at 1 ATA under the following experimental conditions:

1. Dry.

Table 3.1: **Subject anthropometric information**

Subject Number	Age	Height cm	Weight kg	Vital Capacity L (BTPS)
1	46	182.9	77.5	5.594
2	29	178.0	73.5	5.459
3	29	185.3	75.1	5.756
4	35	184.4	82.3	6.216
5	35	182.1	74.3	5.589

2. Immersed to the chin breathing air at mouth pressure (P_m).

3. Immersed to the chin breathing air supplied at lung centroid pressure (P_{LC}).

4. Immersed to the chin breathing air supllied at lung centroid pressure while wearing a wetsuit.

Trials were also repeated at 2, 4, and 6 ATA with the subject immersed and breathing air delivered at P_{LC}.

3.2.3 Dry Conditions

The trials made at 1 ATA with dry experimental conditions served as both a training session and a control for other experiments at similar ambient pressures. A second dry trial at surface pressure was conducted following all other experimental procedures in order to establish the possibility of serial effects on subject performance.

The subject, seated in the upright position, breathed quietly from the mouthpiece through the breathing circuit with the valve opened to allow chamber air to enter and leave the circuit. The nasal airway was occluded by a nose clip. At the end of a normal expiration, the subject relaxed and signaled the chamber tender to start the trial. The valve was immediately closed by the tender, who simultaneously signalled the experimenter outside the chamber to engage the respirator and start the data collection program. The respirator passively ventilated the subject with full respiratory loops for twenty seconds while the subject relaxed with an open glottis. Following the twenty second data collection period, an audio signal was sounded. At this point the tender opened the valve on the breathing circuit, allowing the subject to ventilate independently. The respirator was disengaged momentarily, then reset to its starting position at end expiration (lowest lung volume). Breathing frequencies of 10, 15, 20, 25 and 30 breaths per minute were used in conjunction with an approximate 2.0 L tidal volume, resulting in five consecutive trials. Balanced design determined the order of the breathing frequencies. Trials were repeated when requested by the subject, or when the experimenter was dissatisfied with the data.

3.2.4 Immersed Conditions at 1 ATA

When immersed, the hydrostatic forces applied to the body create a transrespiratory hydrostatic pressure imbalance between the thoracic surface and the airway opening or mouth (when $P_{ao} = P_m$). The hydrostatic imbalance is countered by elastic forces generated in response to a change (decrease) of lung volume. Lung centroid pressure, P_{LC}, has been defined as the hydrostatic breathing pressure required to return the immersed lung relaxation volume to the level which exists in air (Taylor, 1987). It is theorized that the lung centroid pressure is equal to the mean hydrostatic pressure acting on the outside of the thorax during immersion (ie. $P_{LC} = P_{bs}$). Hence when the static breathing pressure P_{ao} is raised to P_{LC} the hydrostatic imbalance between thoracic surface pressure P_{bs} and airway opening P_{ao} is removed. The location of the lung centroid was identified by Taylor (1987) to be 13.6 cms inferior to the sternal notch when in the upright position and 7.0 cms posterior to the sternal notch when supine.

Seated in the upright position, the subject performed the same experimental procedures as in the dry conditions with two variations of experimental environment. Firstly, water in the chamber reached the subject's chin, creating a hydrostatic pressure imbalance across the thorax. Secondly, the subject performed the trials at both uncompensated (mouth) and compensated (lung centroid) breathing pressures. The water temperature ranged from 30° C to 35° C assuring thermal comfort to the swimming trunk clad subject.

During immersion, the subject breathed from a demand regulator attached to the breathing circuit when not being ventilated by the respirator. When breathing at P_{LC}, the demand regulator and a pressure compensator, a device that offsets the hydrostatic pressure imbalance across the pressure transducer, was lowered to the predetermined level of the subject's lung centroid. In addition, the subject was required to wear a neoprene diving hood in an attempt to compensate for facial pressure gradients when breathing at lung centroid pressure. At high rates of ventilation the hands were also used to further restrict cheek motion by applying direct pressure to the area.

A special case of the immersed – lung centroid trial condition at 1 ATA involved the use of a wetsuit. Each subject donned a 3/8" neoprene wetsuit which included both a full length long-john and a jacket. Weights were attached to hold the subject down in the seated position. Breathing gas pressure was supplied at lung centroid level. It was predicted that the additional compression on the chest wall by the wetsuit would make inspiration noticably more difficult where as expiration would exhibit no difference than with the control lung centroid trials. The wetsuit data is only compared to the immsersed-lung centroid trials in the results section and not to the 1 ATA dry and immersed trials without hydrostatic pressure compensation due to their incompatibility of trial factors.

3.2.5 Raised Ambient Pressures

Test procedures were repeated in the immersed state at raised ambient pressures of 2, 4, and 6 ATA. At 2 ATA and 4 ATA, data was collected only when breathing at P_{LC}. At 6 ATA, test procedures were performed with hydrostatic pressure compensation (P_{LC}) and with uncompensated air supply (P_m). During uncompensated breathing pressure trials, problems were experienced with large negative dynamic pressures (P_{ao}) during the expiratory phase. Presumably, the negative pressures resulted from upper airway constriction. As a result, these large negative pressure transients caused subject discomfort, and made it difficult to maintain complete relaxation.

3.2.6 Post-Dive Procedures

Following the testing procedures at depth, decompression commenced according to the Canadian Armed Forces Diving Tables and Procedures (D.C.I.E.M., February 1986 revision). Both the subject and the tender were required to remain under observation in the laboratory for a further sixty minutes following decompression as a safety precaution against the remote possibility of latent bends.

3.3 Compliance and Pulmonary Function Tests

Vital capacity (VC), maximum inspiratory capacity (IC), expiratory reserve volume (ERV), and forced expired volume ($FEV_{1.0}$) for each subject were measured in dry conditions using standard spirometric techniques. Appendix A contains a table of the subject data. Robertson *et al.* (1978) calculated FRC to be approximately 64% of VC. FRC could not be measured while the subject was coupled to the mechanical ventilator. Hence the FRC of each subject was calculated to be 64% of VC for subsequent use in the building of regression coefficients. As a safety precaution during mechanical ventilation trials, tidal volume was never implemented above 75% of IC. $FEV_{1.0}$ is measured as the volume of air expired with maximum effort in 1.0 second following a maximum inspiration. A standard 9.0 L Collins respirometer model P-900 (W.E. Collins Inc.) was used to collect the desired data.

The static respiratory compliance curve was measured to provide comparison with the dynamic compliance obtained through artificial respiration. The respiratory compliance curve was procured by the performance of a series of static pressure-volume relaxation maneuvers over the volume range from residual volume to total lung capacity and vice versa. While seated in the upright position, the subject was trained to open his glottis against an occluded airway. This training procedure initially was performed at maximal inspiratory capacity, and repeated until consistent values were obtained. The procedure was then repeated at maximal expiratory volumes. When the subject felt ready to do the complete test, he inspired fully followed by expiratory increments of approximately 500 mL to 1000 mL until maximum expiratory capacity, *RV*, was reached. At each level of lung volume the airway was occluded and P_{ao} recorded when stable before the next expiratory effort was made. The procedure was repeated beginning with a maximum expiration followed by inspiratory increments of similar volumes. Several recordings were made by each subject. A third order polynomial was fitted to the data representing the total static compliance curve. All five subjects had produced static compliance curves in the past and thus had previous training.

3.4 Data Analysis

3.4.1 Calculations

The data collected for the six environmental conditions was converted into units of pressure and volume and then smoothed before calculations began. A five point smoothing technique developed by Lancos (1965) according to the equation

$$(-3\cdot x_{i-2} + 12\cdot x_{i-1} + 17\cdot x_i + 12\cdot x_{i+1} - 3\cdot x_{i+2}) \cdot (35)^{-1} \qquad (3.1)$$

was used for this purpose. The work of breathing, respiratory compliance and total respiratory resistance were calculated with the aid of computer programs. Specifically, the work of breathing was calculated from the data for inspiratory work, expiratory work, elastic work, and flow resistive work, as a function of the equation

$$Work = \int P \delta V. \qquad (3.2)$$

Total respiratory work is represented by the summation of inspiratory and expiratory work, which in turn are summations of positive elastic and flow resistive work. Flow resistive work is derived from the integration of total resistive work over a complete respiratory cycle. Four subdivisions of dynamic respiratory work exist:

positive inspiratory *(WIP)*
positive expiratory *(WEP)*
negative inspiratory *(WIN)*
negative expiratory *(WEN)*

$$W_{rs} = (WIP - WEN) + (WEP - WIN) \qquad (3.3)$$

Figure 3.5 illustrates the four flow-resistive work divisions by areas on the pressure-volume diagram.

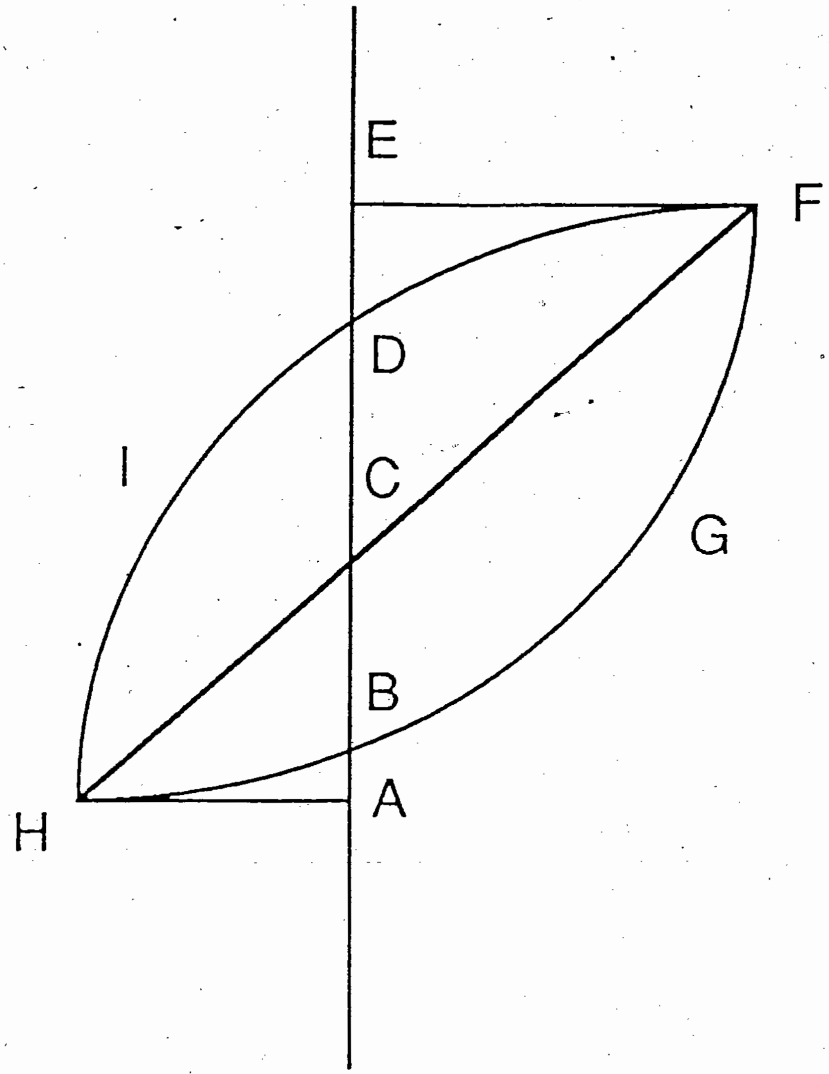


Figure 3.5: Flow-resistive work divisions by area on the pressure-volume diagram.

$WIP = BGFE$

$WEP = DIHA$

$WEN = EFD$

$WIN = AHB.$

Resistive work, inspiratory and expiratory, are illustrated in Figures 3.6 and 3.7, respectively. When end-tidal transrespiratory pressure equals or exceeds relaxation pressure, inspiratory flow resistive work equals positive inspiratory work minus elastic work (Figure 3.6) and expiratory resistive work equals elastic work minus negative expiratory work (Figure 3.7):

$$W_{res(i)} = WIP - elastic\ work \qquad (3.4)$$
$$W_{res(i)} = ABCD - ACD$$

$$W_{res(e)} = elastic\ work - WEN \qquad (3.5)$$
$$W_{res(e)} = ACD - AECD$$

$$W_{el} = 0.5 \cdot (\Delta P \cdot \Delta V) \qquad (3.6)$$
$$W_{el} = ACD.$$

For situations where the end-expired transrespiratory pressure is less than relaxation pressure, different areas on the pressure-volume diagram give the correct results. Inspiratory resistive work equals all inspiratory work minus inspiratory elastic work. Similarly, expiratory resistive work equals all expiratory work minus expiratory elastic work . Figures 3.8 and 3.9 schematically represent these relationships.

$$W_{res(i)} = total\ inspiratory\ work - inspiratory\ elastic\ work \qquad (3.7)$$
$$W_{res(i)} = (BGFE - AHB) - (CFE - CHA)$$

$$W_{res(e)} = total\ expiratory\ work - expiratory\ elastic\ work \qquad (3.8)$$

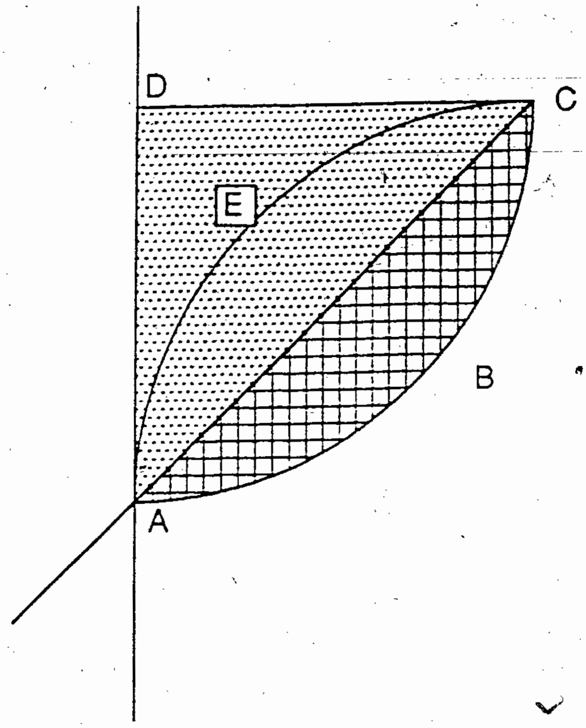


Figure 3.6: Inspiratory resistive work when the end-tidal transrespiratory pressure equals or exceeds relaxation volume.

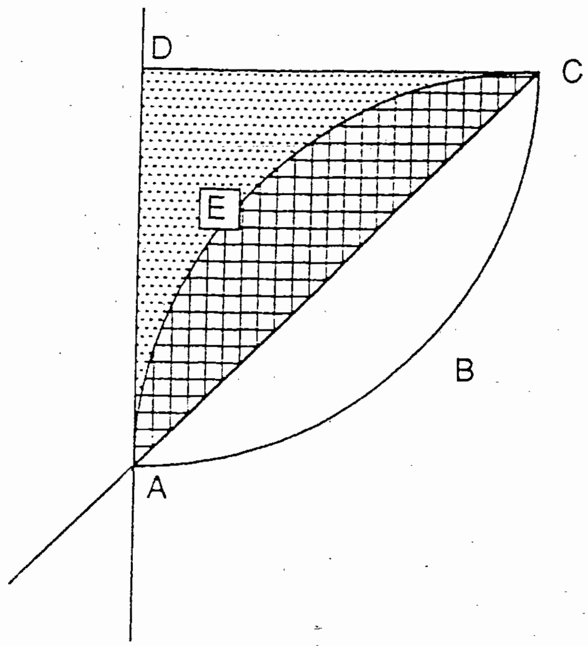


Figure 3.7: Expiratory resistive work when the end-tidal transrespiratory pressure equals or exceeds relaxation volume.

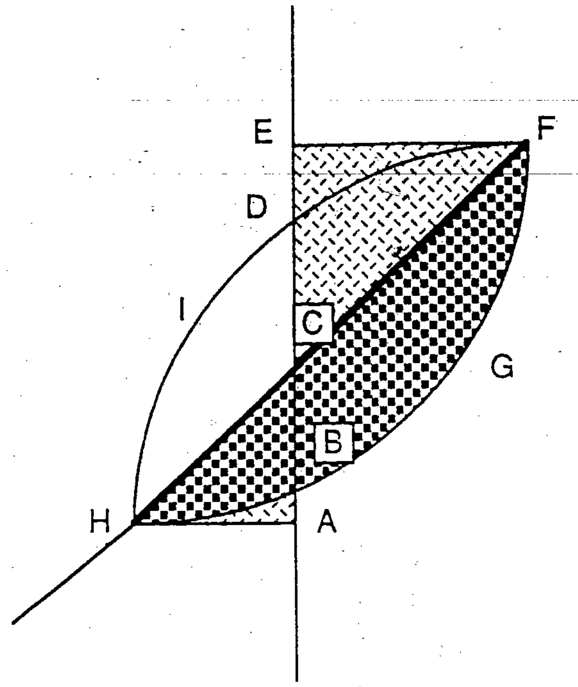


Figure 3.8: Inspiratory resistive work when the end-tidal transrespiratory pressure is less than relaxation volume.

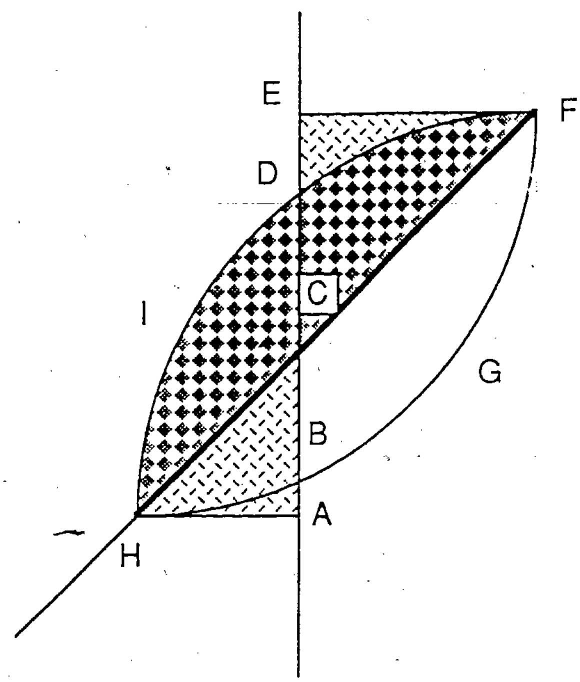


Figure 3.9: Expiratory resistive work when the end-tidal transrespiratory pressure is less than relaxation volume.

$$W_{res(e)} = (DIHA - DFE) - (CHA - CFE)$$

Positive elastic work for a complete breathing cycle

$$W_{el} = 0.5(P_{max} - P_0)(V_{max} - V_0) + 0.5(P_0 - P_{min})(V_0 - V_{min}) \quad (3.9)$$

$$W_{el} = CFE + CHA.$$

Values of respiratory work were normalized to a value of one litre by dividing the work of a particular cycle by the tidal volume. The resultant work was expressed in joules per litre ($J{\cdot}L^{-1}$).

Compliance, the tissues capacity to distend with inflation, was calculated using the following equation:

$$C_{dyn}(rs) = \Delta V \cdot \Delta P^{-1} \quad (3.10)$$

Dynamic compliance is the ratio of tidal volume to pressure change between two points of zero gas flow at either end of the tidal excursion (Mead and Milic-Emili, 1964; Taylor, 1987). Compliance is reported in units of $L{\cdot}kPa^{-1}$.

Resistance at any point in the breathing cycle is calculated by dividing the resistive pressure at point i by the corresponding rate of flow. Resistance is then reported at a specified flow rate, commonly 0.5 $L.sec^{-1}$. In addition to 0.5 $L.sec^{-1}$, resistance was calculated at rates of flow of 1.0 $L.sec^{-1}$ and 2.0 $L.sec^{-1}$. Resistance is reported in units of $kPa.L^{-1}.sec$.

$$Resistance = P_{res} \cdot \dot{V}^{-1} \quad (3.11)$$

Resistive pressure at any point i is calculated by subtracting elastic pressure from transrespiratory pressure at the desired instant. Elastic pressure must be calculated first in order that resistive pressure be determined.

$$P_{el} = P_{min} + (V_i - V_{min}) \cdot (\Delta P \cdot \Delta V^{-1}) \quad (3.12)$$

$$P_{res} = P_{rs} - P_{el} \quad (3.13)$$

3.4.2 Statistical Analysis

Repeated measures analyses of variance were performed on the respiratory work and resistance data to test for the effects of independent factors on the calculated results. With respect to respiratory work, two factors were considered: rate of ventilation and experimental design (condition at 1 ATA, and density with the trials performed at different ambient pressures). Respiratory resistance data was identified as having three factors: flow rate, lung volume, and density. Any interaction between the factors would be indicated by the analyses of variance as well. Significant differences in the effects were subsequently treated with Tukey's HSD test to show the level or levels within the factors where the differences in means occur. A significant difference was assumed to exist when a probability of $p<0.05$ was achieved. Actual values of probabilty are quoted in the results.

Complications arise while attempting to fit explanatory equations to data when interactions between factors occur. Interaction is the condition where the relationship of interest is different at different levels of the extraneous variables (Kleinbaum *et al.*, 1988). Multiple regression analysis is required when interactions exist in order to design an appropriate predictive model. Several predictor variables can be applied, but since physiological knowledge is available, constraints on the variables can be made.

The problem of developing the *best* equation to the present data is confounded by the repeated measures nature of the data. If one chooses to treat all the data as independent measures, a model could be developed and tested. Such a manipulative technique is not entirely sound statistically due to the actual dependence of data points with one another, in addition to the wide variance obtained due to the differences in subjects. An alternative method is to apply a single-outcome regression model with variables that relate theoretically to respiratory work and respiratory resistance. The coefficients for the variables for each subject vary within a normal distribution for the true population. That is, although the coefficients are not expected to be identical for different people, the variance of the coefficients β_0, β_1, and β_2, are normally distributed.

The best estimate for each coefficient β_i, when considering the sample population, is the mean of the estimates, which is β_i and described by the equation below:

$$\beta_i = \Sigma\beta_{ij}/n \qquad (3.14).$$

Similarly, the variance of the mean of the estimate, $Var(\beta_i)$ is calculated to include the average of the individual coefficient variance for each subject in addition to the mean coefficient variance. This is the method that is used to develop the coefficients for the predictor models for both work and resistance.

3.4.3 Respiratory Work Model

One explanatory equation was developed to describe the trend of the experimental data results. The equation is theoretically based rather than a *best fit* mathematical relationship which severely limited the number and type of variables chosen to be in the model:

$$W = \beta_0 + \beta_1 \cdot \dot{V} + \beta_2 \cdot D \cdot \dot{V}^2. \qquad (3.15)$$

In this equation, W represents work in $J.L^{-1}$, V represents ventilation rate in $L.min^{-1}$, and D represents the gas density relative to air at one atmosphere (ie. at 1 ATA, D=1). β_0, the constant coefficient, was included in the theoretical regression equation based on the results of the data collected. A constant coefficient would be required if respiratory hysteresis exists. The concept of respiratory hysteresis is discussed in

Chapter 6. This equation is based on the physiological knowledge that respiratory flow has both laminar and turbulent components. The former is more prevalent with normal atmospheric and physiological conditions. With laminar flow, respiratory pressure varies directly with flow rate. However, when flow becomes turbulent, respiratory pressure varies with the square of flow rate. The degree of confidence of the equation was ascertained by the imposition of limits, based on regression coefficients and the standard error of the estimate.

3.4.4 Respiratory Resistance Model

Respiratory resistance has three physiological factors that must be accounted for in a model: gas flow rate, gas density, and lung volume. The partly laminar and partly turbulent nature of respiratory airflow indicates that the value of resistance will increase as gas flow rate increases. Since turbulent flow has a greater contribution to the total, resistance will increase at a greater rate with raised gas density. The length and diameter of the airways change with lung volume: at low lung volumes, airway resistance should be greater and at high lung volumes, airway resistance should be smaller. DuBois (1964) and McKenna *et al.* (1973) have suggested that resistance is inversely proportional to lung volume. With inspiration, the lungs and chest cavity are expanded. As the size of the lungs increase, the airways increase proportionately (DuBois, 1964). Considering Poiseuille's Law regarding laminar flow (Equation 1.2), the resistance term k_1 is proportional to the inverse of the cube of dimension (L^3, where L represents dimension), or $k_1 \alpha L^{-3}$. Likewise, volume is proportional to dimesion cubed, or ($V \alpha L^3$). Hence, $k_1 \alpha V^{-1}$, and is a linear term. The turbulent component of resistance is not as easily dissected to illustrate its relationship with lung volume as airway geometry is a prevelent factor. From the equation of airway turbulence (Equation 1.4), the term k_2 is proportional to L or

$$k_2 \alpha l \cdot r^{-5} \alpha L^{-4}.$$

As volume varies with L^3, it might be approximated that $k_2 \alpha V^{-1.25}$, assuming the effects of volume on radius, r, and airway length, l are proportional. In this study, lung volume has therefore been modelled with equal contribution for both the laminar and the turbulent components of resistance (ie. $R \alpha V^{-1}$) since the exact relationship is unknown. Briefly, the regression equation developed to describe respiratory resistance applies the laws of laminar and turbulent flow, and Poiseuille's Law as follows:

$$R = (\beta_0 + \beta_1 \cdot \dot{V} \cdot D) \cdot V^{-1} \tag{3.16}$$

where D is density (ATA), $\dot{V}$ is flow ($L \cdot sec^{-1}$), and V is lung volume (L).

CHAPTER 4

RESULTS: RESPIRATORY WORK

Respiratory work is reported in this chapter under two categories. The first is work at 1 ATA ambient pressure only; the second is work at ambient pressures of 1, 2, 4, and 6 ATA. Respiratory work values are reported for five work types: elastic, inspiratory flow-resistive, expiratory flow-resistive, total flow-resistive, and total respiratory work.

4.1 Work at 1 ATA

Three experimental conditions were performed at 1 ATA. First, subjects were mechanically ventilated while seated in a dry environment. This condition will be referred to as "dry" in the remainder of the chapter. Second, the subjects were mechanically ventilated while seated in water to the level of the chin. No hydrostatic pressure compensation was supplied to the subject during this trial. This trial type will be referred to as "immersed - mouth" throughout the chapter due to the breathing gas supply pressure being at the level of the mouth (P_m). Third, the subjects were mechanically ventilated while seated in water to chin level. Breathing gas supply pressure was compensated to lung centroid pressure (P_{LC}) for each subject. Hence, this trial type will be referred to as "immersed - lung centroid".

The theoretically based regression equation relating work with rate of ventilation was not fitted to the data at 1 ATA due to the limited number of data points. Each subject and each condition would be considered separately and thus the equation, with two degrees of freedom, would be fit to a total of five points. Instead, the mean data and subject error are graphically displayed for each condition to illustrate the differences in the conditions. Subject error is the standard deviation of work values between subjects. A best fit curve with positive coefficients is fitted through the data, but no coefficient of correlation is reported.

4.1.1 Elastic Work

An analysis of variance performed on the elastic work data showed no significant difference of elastic work between the tested rates of ventilation. However, a significant difference was found with the condition factor (p<.05). The mean values of elastic work for the three conditions and their associated variances are shown in Table 4.1. Tukey's HSD test revealed that a significant difference does exist between the dry and immersed–mouth trials (p<.05[and between immersed–mouth and immersed–lung centroid trials (p<.05). No significant difference was found between the dry and immersed–lung centroid conditions.

4.1.2 Inspiratory Flow-resistive Work

Analysis of variance revealed a significant difference in flow resistive work between the different test conditions (p<.01) and at different rates of ventilation (p<.005). *Post hoc* analysis indicate that with respect to the three environmental conditions, a significant difference does exist between dry and immersed–mouth trials (p<.05) and between immersed–lung centroid and immersed–mouth (p<.05) trials, but not between dry and immersed–lung centroid trials. With respect to changes in ventilation, paired t–tests indicated that significant differences in work were not found between all adjacent levels of ventilation. However, in general the resistive work was greater (p<.05) at higher rates of ventilation. Figure 4.1 illustrates the mean values and subject error for each ventilation rate.

4.1.3 Expiratory Flow-resistive Work

Unlike the inspiratory flow–resistive work results, analysis of variance found no significant differences between environmental conditions. The rate of ventilation factor produced significant differences in work values (p<.05). *Post hoc* analyses indicated that significant differences in work values did not occur between consecutive minute ventilations, but on a broader scale. The highest rate of ventilation, 60 $L.min^{-1}$, produced expiratory resistive work values significantly different from all other rates of ventilation (p<.05). Similarly, expiratory flow–resistive work at 50 $L.min^{-1}$ differed from work at all other rates of ventilation

Table 4.1: **Mean elastic work at each experimental condition performed at one atmosphere pressure.**

CONDITION	*ELASTIC WORK ($J \cdot L^{-1}$)*
Dry	0.724 ± 0.177
Immersed–Mouth	1.218 ± 0.576
Immersed–Lung Centroid	0.894 ± 0.831

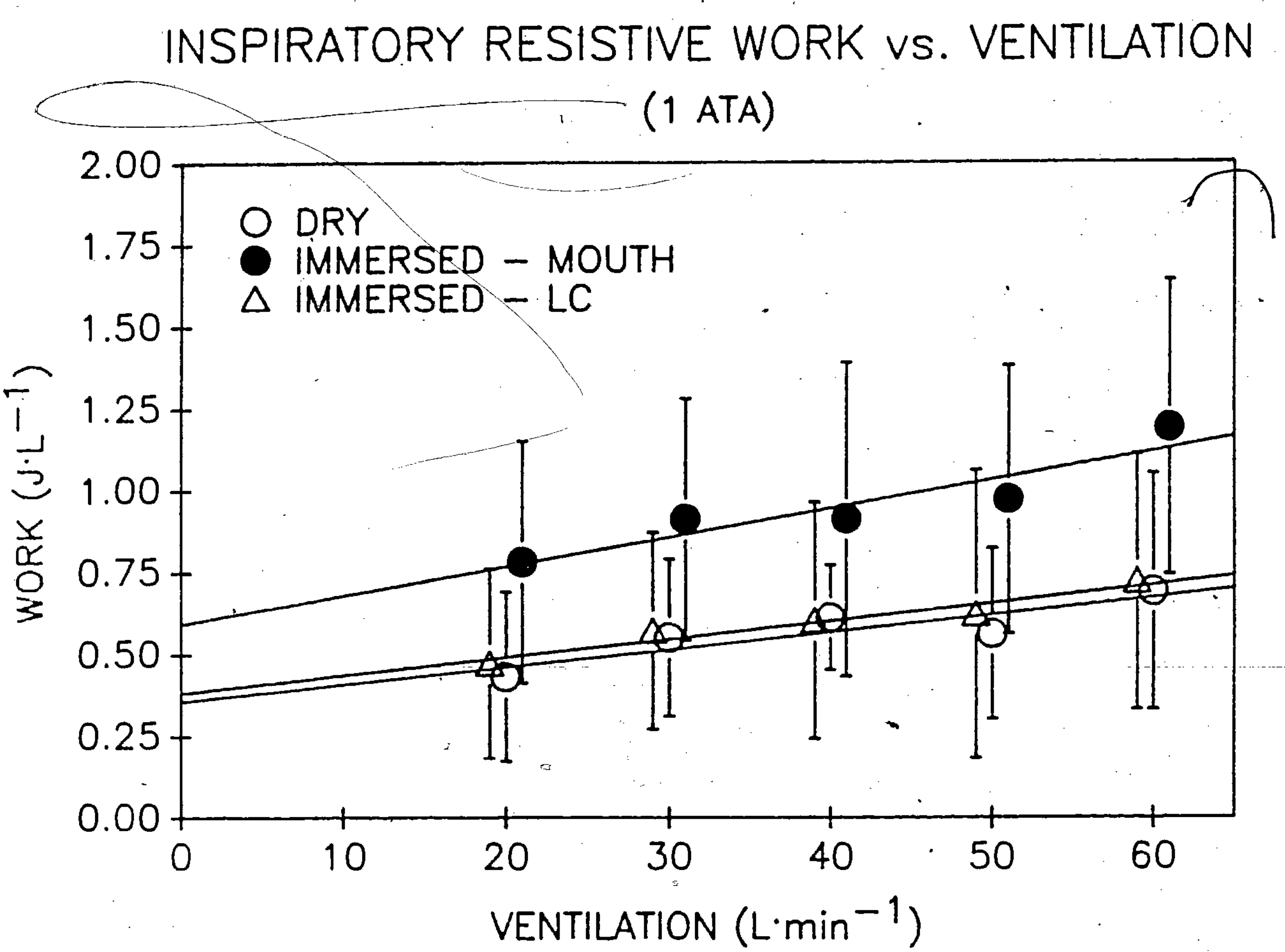


Figure 4.1: Relationship of inspiratory flow-resistive work with increasing minute ventilation at 1 ATA. Dry, immersed-mouth, and immersed-lung centroid experimental conditions.

($p<.05$). Figure 4.2 illustrates the mean work and subject error results. Note that although the mean value of expiratory flow-restive work at each minute ventilation is higher for the immersed-mouth trial than for the remaining two conditions, the variance in the work values reduces the chance of significance.

4.1.4 Total Resistive Work

Analysis of variance showed a significant difference ($p<.05$) with regards to the condition effect. A difference in flow-resistive work values occurred between dry and immersed-mouth trials ($p<.05$), but not between immersed-mouth and immersed-lung centroid or dry and immersed-lung centroid trials.

The rate of ventilation effect on total resistive work was also significant ($p<0.01$). The difference in flow-resistive work due to increasing minute ventilations was not found consistently between consecutive rates but rather in a broader fashion across the entire range of ventilations. All tested rates of ventilation produced total resistive work values significantly less than those obtained at 60 $L.min^{-1}$ ($p<.05$). The differences in the experimental conditons are illustrated in Figure 4.3.

4.1.5 Total Respiratory Work

Analysis of variance showed significant rate of ventilation and environmental condition effects ($p<.001$). No interaction between the two factors occurred. Tukey's HSD test indicated that a difference does exist between total work values obtained during dry and immersed-mouth trials ($p<.05$). Likewise, a difference in respiratory work does exist between immersed-mouth and immersed-lung centroid trials ($p<.05$). No difference was found between total work obtained in dry and immersed-lung centroid environmental conditions. Figure 4.4 graphs the means of the total respiratory work values with increasing minute ventilation.

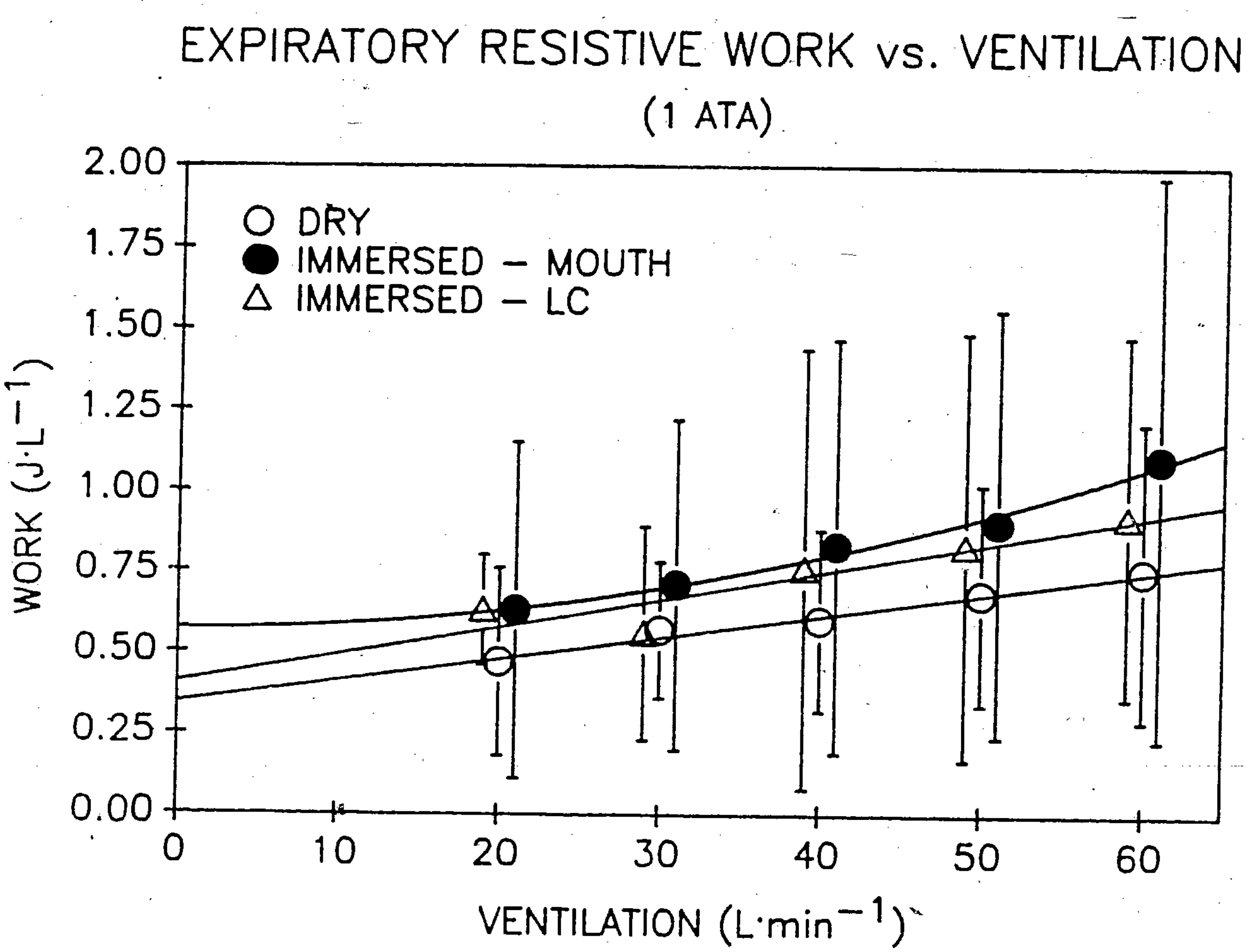


Figure 4.2: Relationship of expiratory flow-resistive work with increasing minute ventilation at 1 ATA. Dry, immersed-mouth, and immersed-lung centroid experimental conditions.

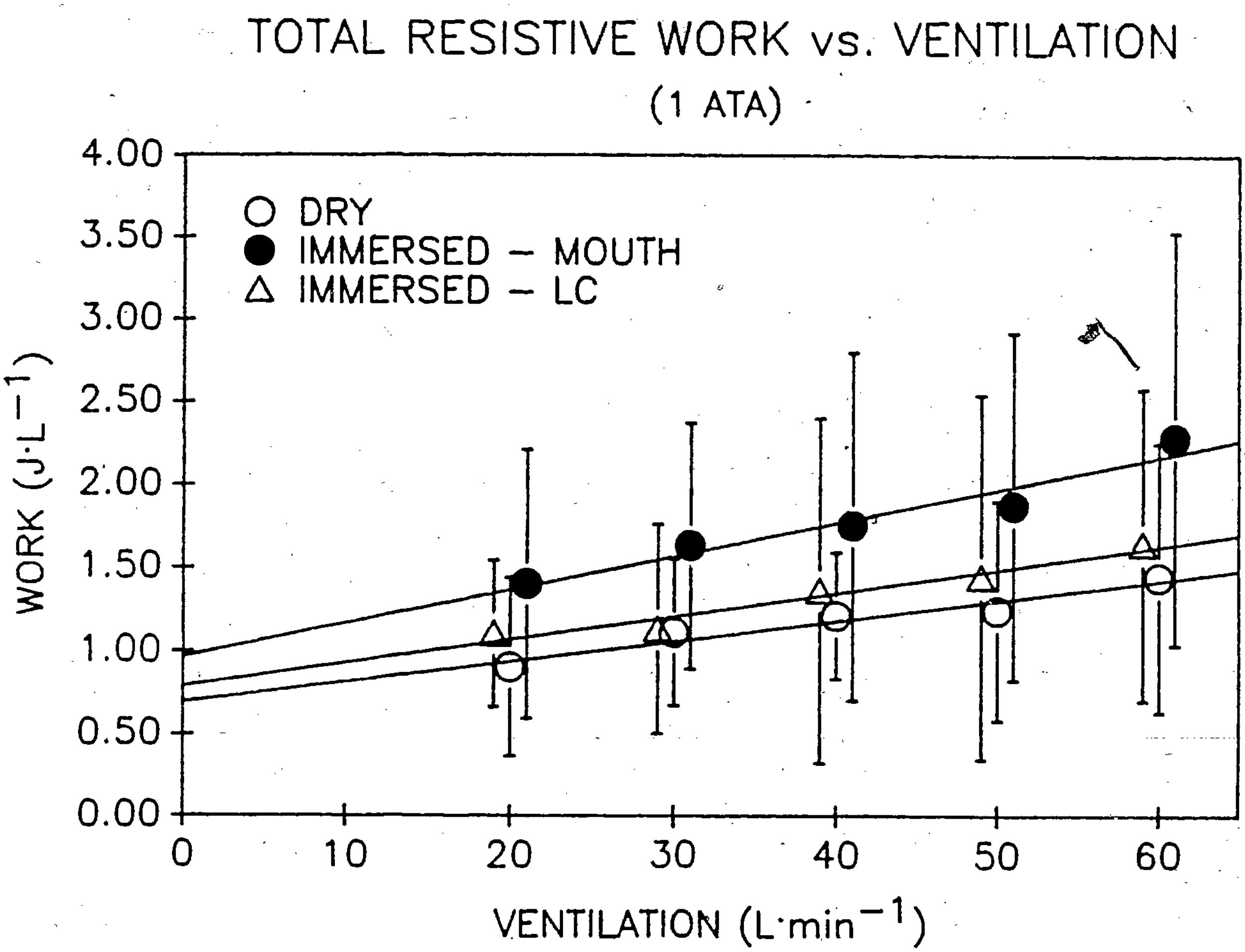


Figure 4.3: Relationship of total flow-resistive work with increasing minute ventilation at 1 ATA. Dry, immersed-mouth, and immersed-lung centroid experimental conditions.

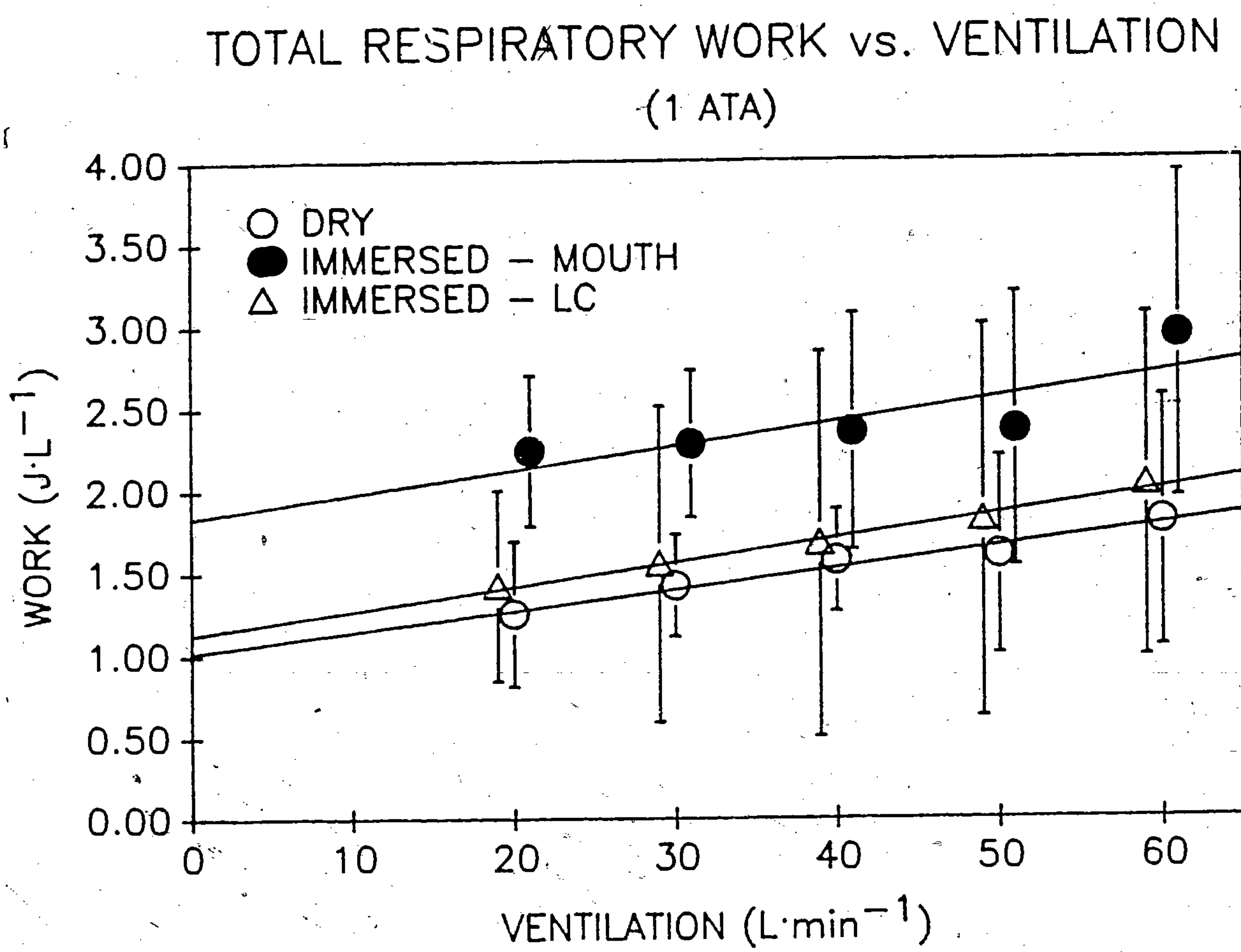


Figure 4.4: Relationship of total respiratory work with increasing minute ventilation at 1 ATA. Dry, immersed-mouth, and immersed-lung centroid experimental conditions.

4.1.6 Analysis of the Five Respiratory Work Components

Individual work components for the immersed - lung centroid condition are plotted against increasing minute ventilation in Figure 4.5. The figure serves to illustrate the relative magnitude of the respiratory work components. Tukey's HSD test indicated a significant difference between inspiratory and expiratory flow-resistive work in the immersed-lung centroid condition ($p<.05$), but not with the other two conditions.

4.1.7 Wetsuit Versus Control Immersed-Lung Centroid Trials

An analysis of variance was performed on immersed-lung centroid data obtained when subjects wore only a swimsuit, and when wearing a wetsuit. Elastic work did not differ significantly having mean values of 0.894±0.831 $J.L^{-1}$ and 1.037±0.221 $J.L^{-1}$ for swimsuit and wetsuit trials, respectively.

Inspiratory resistive work was significantly different with wetsuit and swimsuit conditions ($p<.05$), with work higher while wearing the wetsuit. Figure 4.6 illustrates the relationship for the two conditions at the five tested rates of ventilation. Analysis of variance displayed no differences in expiratory flow-resistive work between the two conditions ($p<.05$). Expiratory flow-resistive work values are displayed graphically in Figure 4.7.

Total resistive work for the two trial conditions to be significantly different ($p<.05$). The relationship of total resistive work with ventilation rate is illustrated in Figure 4.8.

The analysis of variance performed on total respiratory work indicated a condition effect ($p=.05$). The relationships relating total respiratory work to increasing minute ventilation while immersed and wearing a swimsuit and wetsuit are illustrated in Figure 4.9.

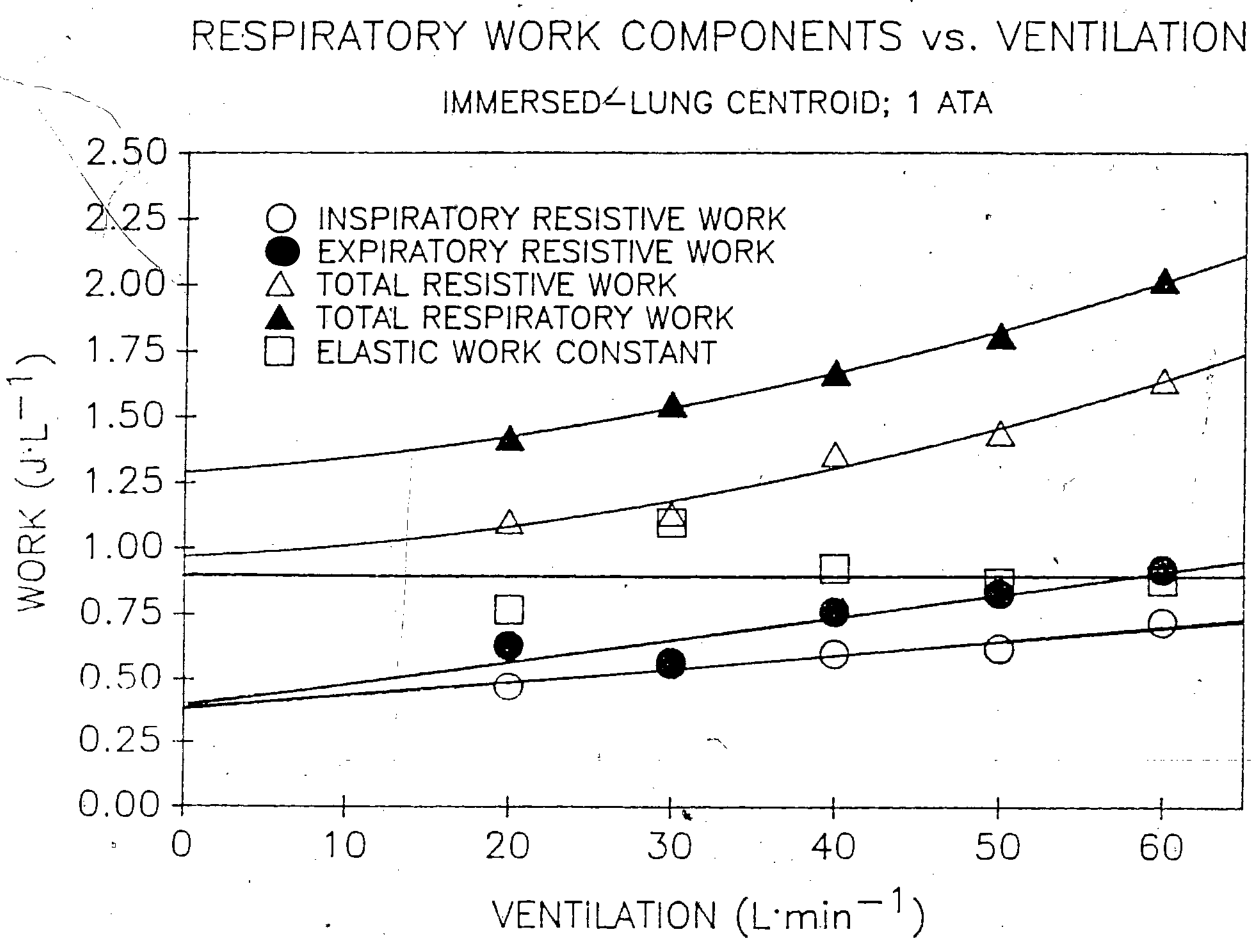


Figure 4.5: The relationship of respiratory work with increasing minute ventilation in the immersed experimental environment with hydrostatic pressure compensation at 1 ATA. (Elastic, inspiratory flow-resistive, expiratory flow resistive, total flow-resistive and total respiratory work values)

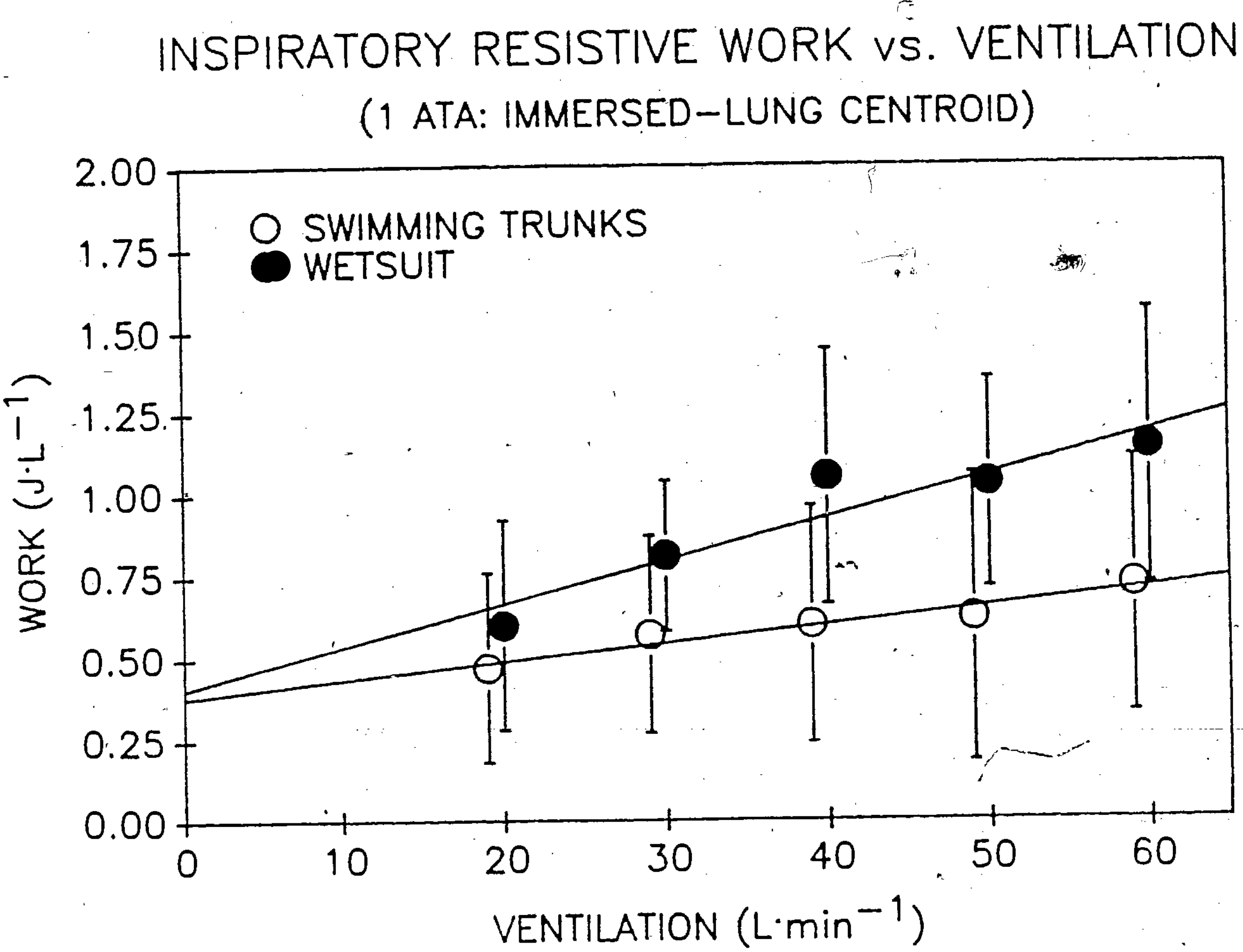


Figure 4.6: The relationship of inspiratory flow-resistive work with increasing minute ventilation while seated immersed and breathing with hydrostatic pressure compensation: wetsuit vs. swimming trunks.

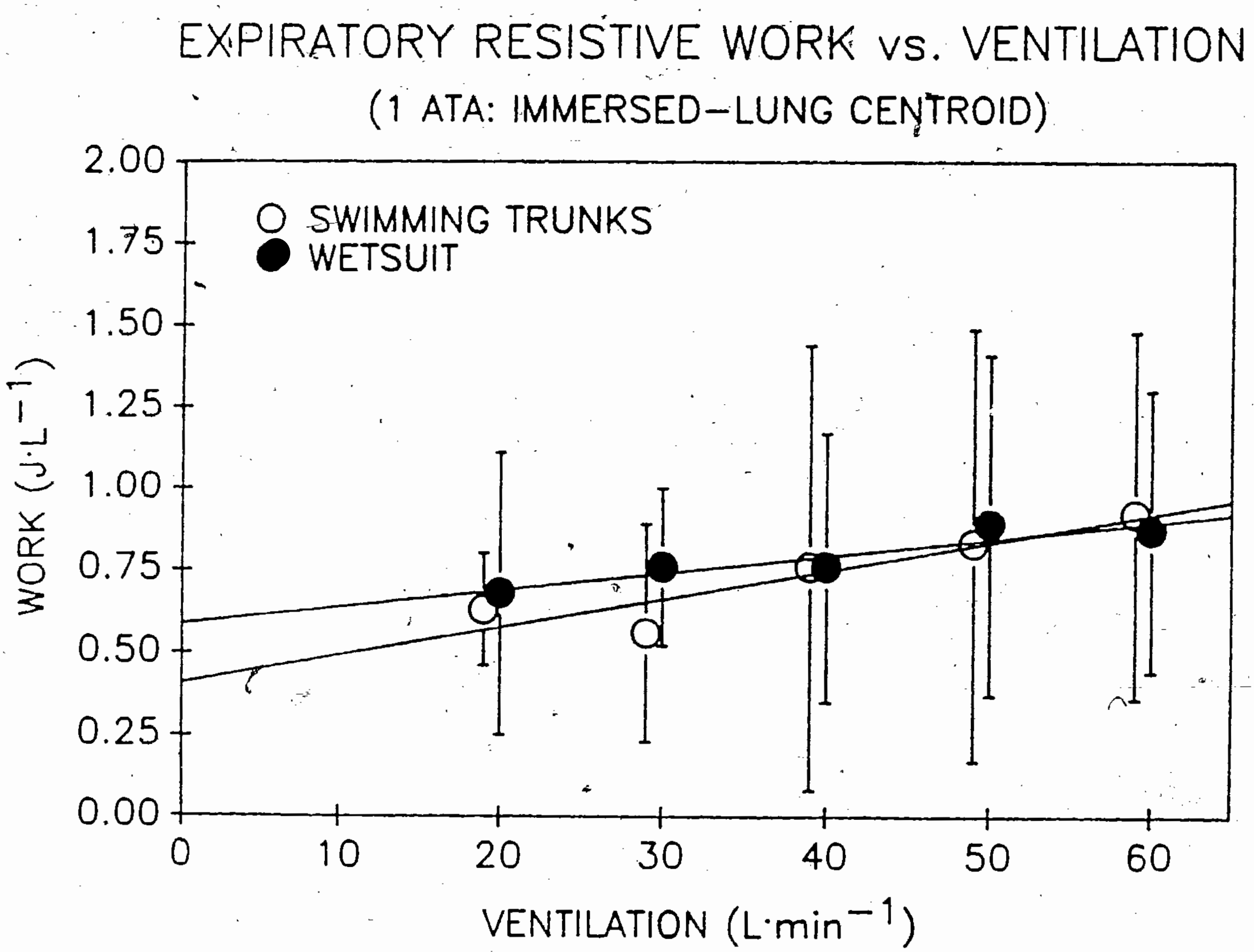


Figure 4.7: The relationship of expiratory flow-resistive work with increasing minute ventilation while seated immersed and breathing with hydrostatic pressure compensation: wetsuit vs. swimming trunks.

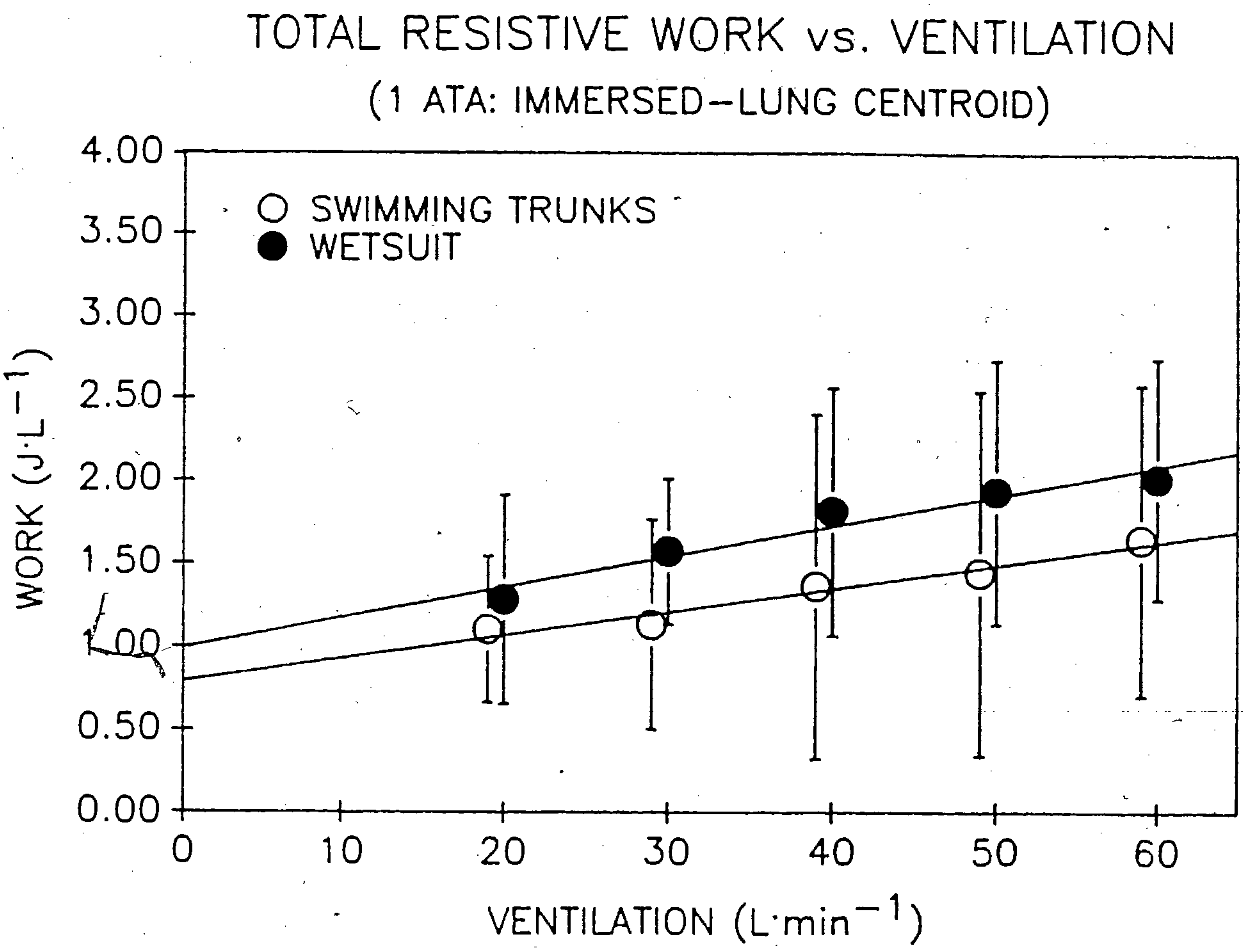


Figure 4.8; The relationship of total flow-resistive work with increasing minute ventilation while seated immersed and breathing with hydrostatic pressure compensation: wetsuit vs. swimming trunks.

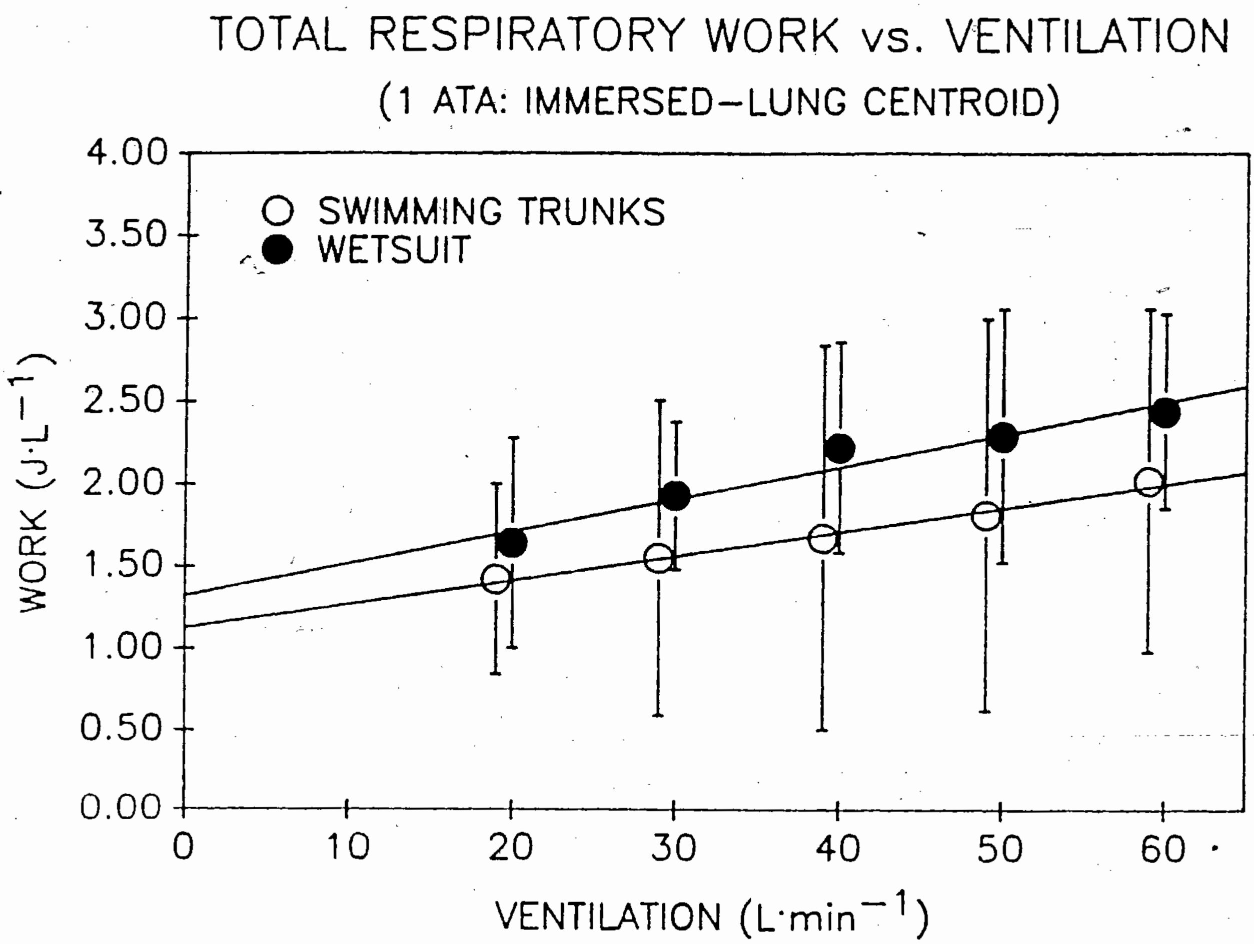


Figure 4.9: The relationship of total respiratory work with increasing minute ventilation while seated immersed and breathing with hydrostatic pressure compensation: wetsuit vs. swimming trunks.

4.2 Respiratory Work at 1, 2, 4, and 6 ATA

One experimental condition, immersion to the chin with hydrostatic pressure compensation of the breathing gas delivery pressure to lung centroid level, termed "immersed–lung centroid", was tested in this particular set of experimental procedures. Four pressure environments were applied to the testing condition. These ambient pressures were 1 ATA (surface), 2 ATA (10 m.s.w.), 4 ATA (30 m.s.w.), and 6 ATA (50 m.s.w.). The calculated data was subjected to analyses of variance to test for significant differences among the ventilatory and density factors. Tukey's HSD test further identified the differences within the factors. The theoretically developed regression equations accounting for density–ventilation interaction were fitted to the data based on the theory of laminar and turbulent air flow components in respiration. Individual regression equations and coefficient variance for the five work types are presented in Appendix B.

4.2.1 Elastic Work

Repeated measures analysis of variance found no significant differences of elastic work to exist within either the density or the ventilation factors. Values of mean elastic work at each atmospheric pressure are shown in Table 4.2.

4.2.2 Inspiratory Flow-Resistive Work

The analysis of variance indicated that inspiratory flow–resistive work differed significantly with different ambient pressures ($p<.05$) and at different rates of ventilation ($p<.001$). Furthermore, a density–ventilation interaction ($p<.005$) was identified. Tukey's HSD test applied to the density effect identified significant differences in inspiratory flow–resistive work between 1 ATA and 4 ATA ($p<.05$); 1 ATA and 6 ATA ($p<.05$); and 2 ATA and 4 ATA ($p<.05$).

Application of Tukey's test to the ventilation effect revealed that differences in inspiratory flow–resistive work values occurred at all levels of ventilation with the highest rate of ventilation, 60 $L.min^{-1}$ ($p<.05$). A general trend is indicated with higher rates of ventialtion associated with higher work values ($p<.05$), although adjacent

Table 4.2: **Mean elastic work at each atmospheric pressure.**

PRESSURE (ATA)	*ELASTIC WORK* ($J \cdot L^{-1}$)
1	0.894 ± 0.831
2	0.686 ± 0.429
4	0.664 ± 0.367
6	0.745 ± 0.692

ventilations did not necessarily result in significant differences.

The theoretical regression polynomial fitted to the data produced the following result:

$$W = 0.186 + 0.0083 \cdot \dot{V} + 0.000032 \cdot D \cdot \dot{V}^2 \qquad (4.1)$$

$$r_a^2 = 0.58 \qquad (r_a = 0.76)$$

Figure 4.10 displays the mean inspiratory flow–resistive work for the four densities at each rate of ventilation; with the theoretical regression equations superimposed on the data.

4.2.3 Expiratory Flow-Resistive Work

Analysis of variance showed that expiratory flow–resistive work in common with inspiratory flow–resistive work is affected by both ventilation ($p<.001$) and density ($p<.001$) factors. An interaction between the two factors was also identified ($p<.001$). Tukey's test indicated real differences in work values at 1 ATA and 4 ATA ($p<.05$); 1 ATA and 6 ATA ($p<.051$); 2 ATA and 4 ATA ($p<.05$); 2 ATA and 6 ATA ($p<.05$); and 4 ATA and 6 ATA ($p<.05$).

The theoretical polynomial regression curve fitted to the data produced the following result:

$$W = 0.394 + 0.0048 \cdot \dot{V} + 0.000061 \cdot D \cdot \dot{V}^2 \qquad (4.2)$$

$$r_a^2 = 0.72 \qquad (r_a = 0.85)$$

Figure 4.11 illustrates the mean expiratory flow–resistive work means for the four densities at each rate of ventilation The theoretical regression curves are superimposed on the mean data.

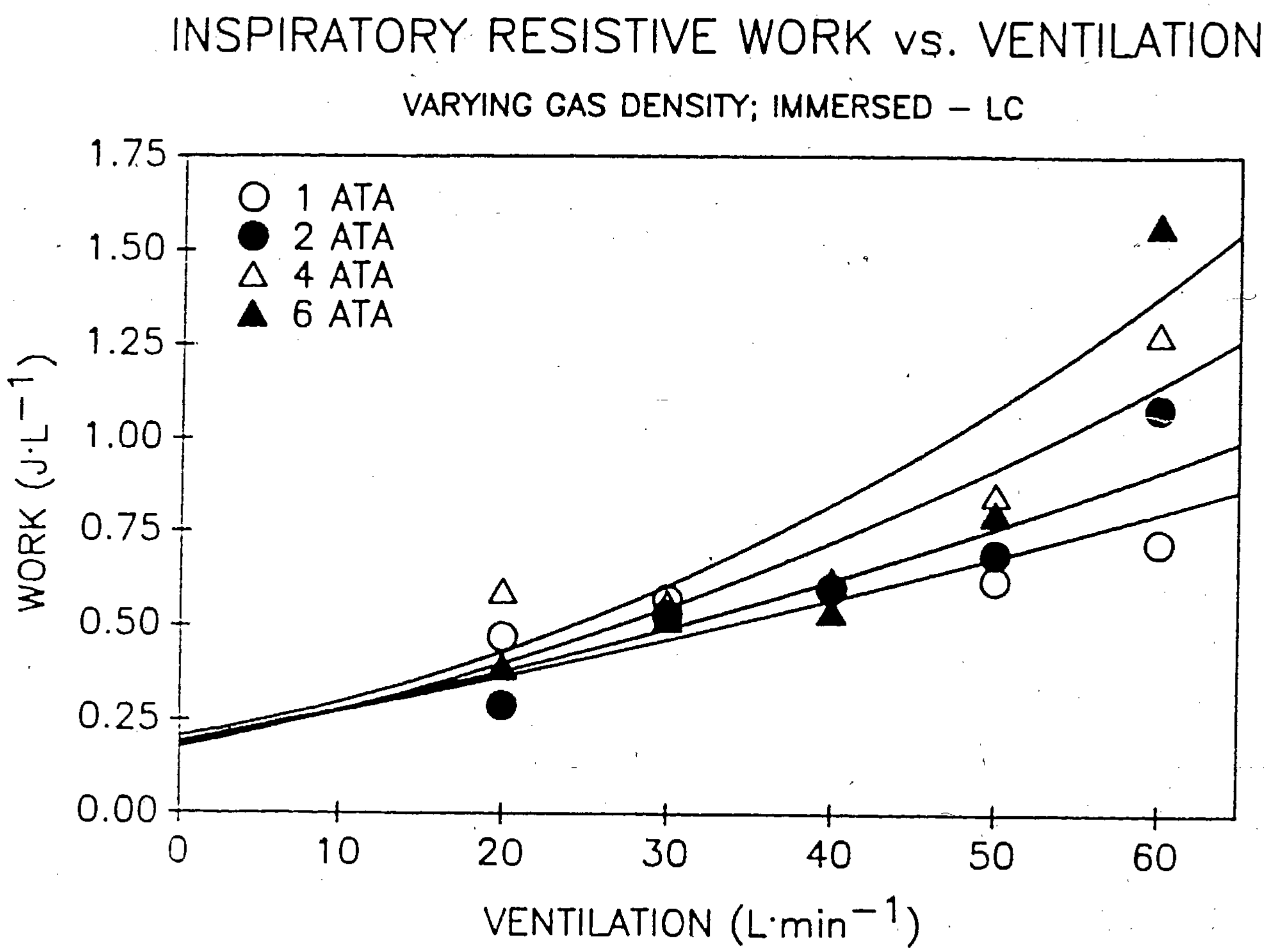


Figure 4.10: Inspiratory flow-resistive work with increasing minute ventilation at 1, 2, 4, and 6 ATA. Immersed – lung centroid.

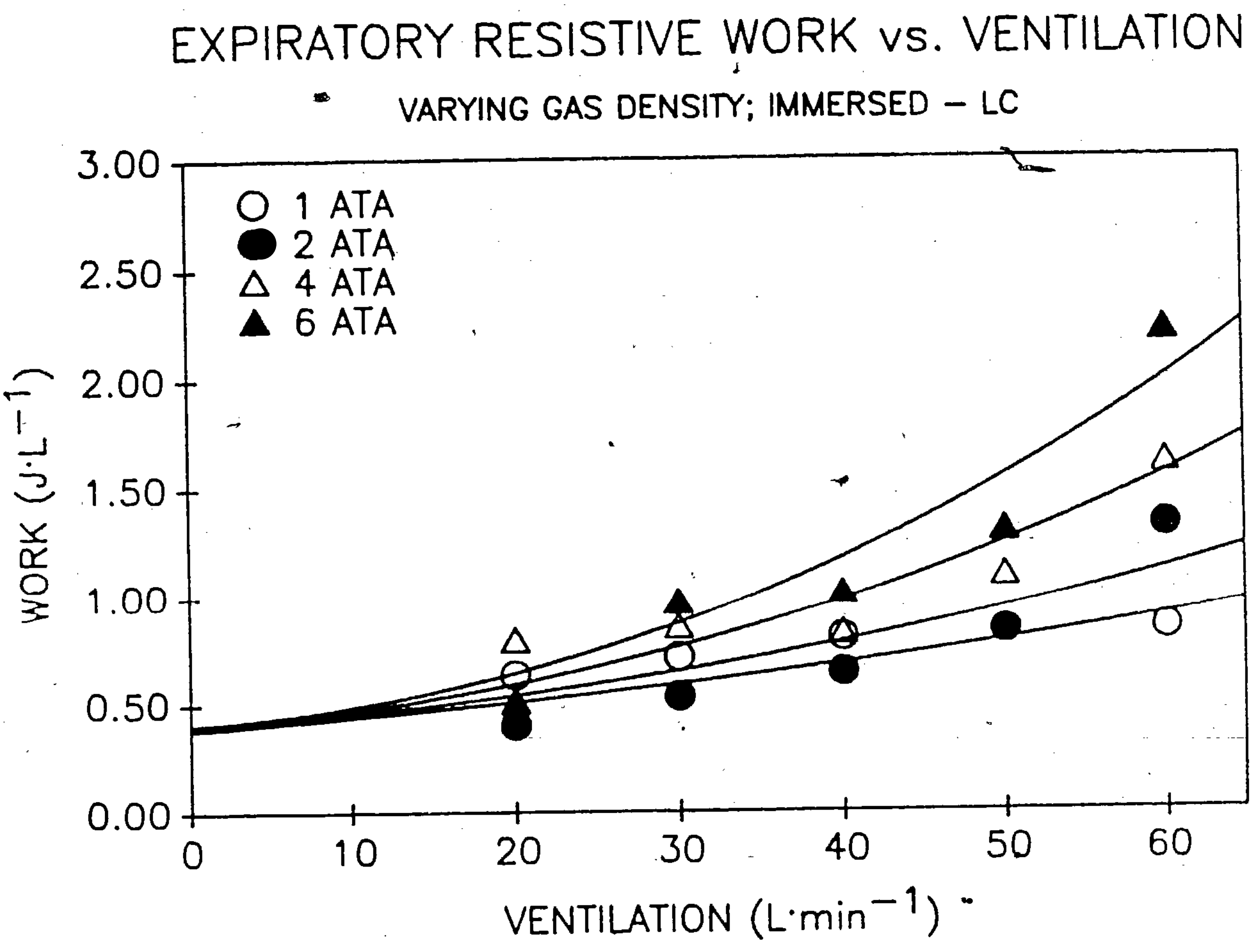


Figure 4.11: Expiratory flow-resistive work with increasing minute ventilation at 1, 2, 4, and 6 ATA. Immersed – lung centroid.

4.2.4 Total Flow-Resistive Work

Since total flow-resistive work is the summation of inspiratory and expiratory flow-resistive work, the statistical analysis and the graphed data would therefore summarize both components. According to the analysis of variance, a significant difference is found among resistive work values with respect to both density ($p<.001$) and the ventilation ($p<.001$) factors. An interaction between the two factors is also apparent ($p<.001$). With regards to density, significant differences were identified between 1 ATA and 4 ATA ($p<.05$); 1 ATA and 6 ATA ($p<.05$); 2 ATA and 4 ATA ($p<.01$); 2 ATA and 6 ATA ($p<.05$); and 4 ATA and 6 ATA ($p<.05$)

The theoretical regression curve fitted to the data gave the following result:

$$W = 0.577 + 0.0129\cdot\dot{V} + 0.000094\cdot D\cdot\dot{V}^2 \qquad (4.3)$$

$$r_a^2 = 0.75 \qquad (r_a = 0.87)$$

Figure 4.12 displays the mean total flow-resistive work means for the four densities at each rate of ventilation. The theoretically based regression equations are superimposed on the data.

4.2.5 Total Respiratory Work

Total respiratory work is a function of both elastic and flow-resistive work components. Analysis of variance exhibited significant differences ($p<.0001$) within the density factor and the ventilation factor. Additionally, it is highly probable ($p<.0001$) that an interaction between the two factors occurs. Tukey's HSD test identified significant differences in total respiratory work between 1 ATA and 4 ATA ($p<.05$); 1 ATA and 6 ATA ($p<.05$), 2 ATA and 4 ATA ($p<.05$), and 2 ATA and 6 ATA ($p<.05$).

The theoretical polynomial regression curve fitted to the data produced the following result:

$$W = 0.920 + 0.0121\cdot\dot{V} + 0.000084\cdot D\cdot\dot{V}^2 \qquad (4.4)$$

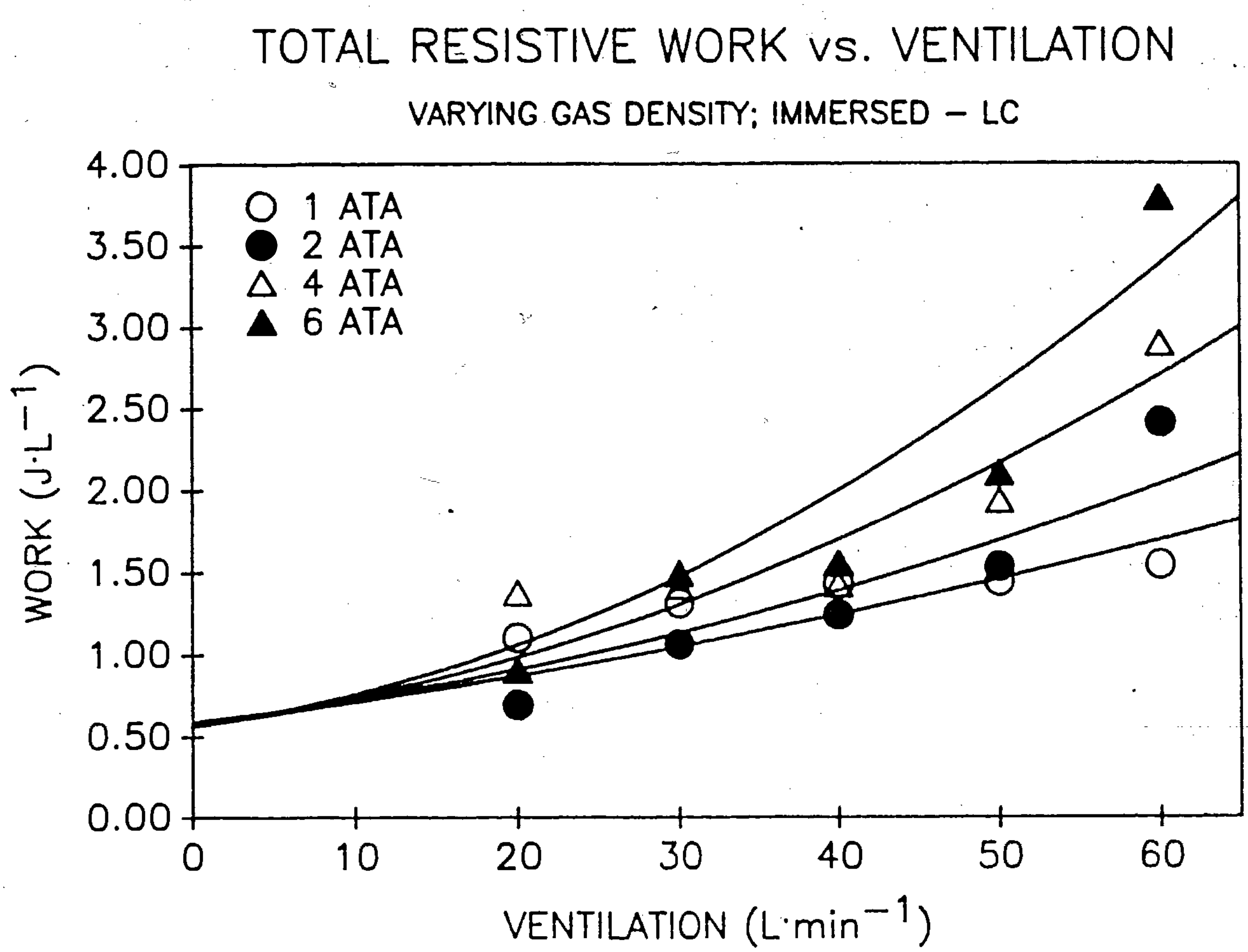


Figure 4.12: Total flow-resistive work with increasing minute ventilation at 1, 2, 4, and 6 ATA. Immersed - lung centroid.

$r_a^2 = 0.68 \qquad (r_a = 0.83)$

Figure 4.13 illustrates the mean of total respiratory work for the four densities at each rate of ventilation with the theoretcal regression curves superimposed.

4.2.6 Analysis of the Five Respiratory Work Components

Individual work components for the ambient pressures of 1 ATA and 6 ATA are plotted against increasing minute ventilation in Figures 4.14 and 4.15, respectively. The data points represent the theoretical relationship derived for the empirical data of the previous figures. The present figures serve to illustrate the relative magnitude of the respiratory work components at each gas density. Total resistive work is clearly marked as the summation of both inspiratory and expiratory resistive work values. The addition of the elastic work component to the total resistive work component is also evident with the total respiratory work values.

The only two work components for which *post hoc* analysis is appropriate are inspiratory and expiratory resistive work values. The results of the analysis indicates that significant differences do exist between the two work components at all four air densities ($p<.05$). Significant differences between inspiratory and expiratory flow-resistive work also exist within the ventilation factor at all rates of ventilation ($p<.05$).

4.3 Ventilatory Power

Ventilatory power represents respiratory work performed over a unit of time (second). Power was calculated from the present results by multiplying the corresponding rate of ventilation to each work value (since work is presented in Joules per Litre). The power was then plotted versus increasing minute ventilations. Figure 4.16 illustrates the flow-resistive respiratory powers obtained at all four tested ambient pressures.

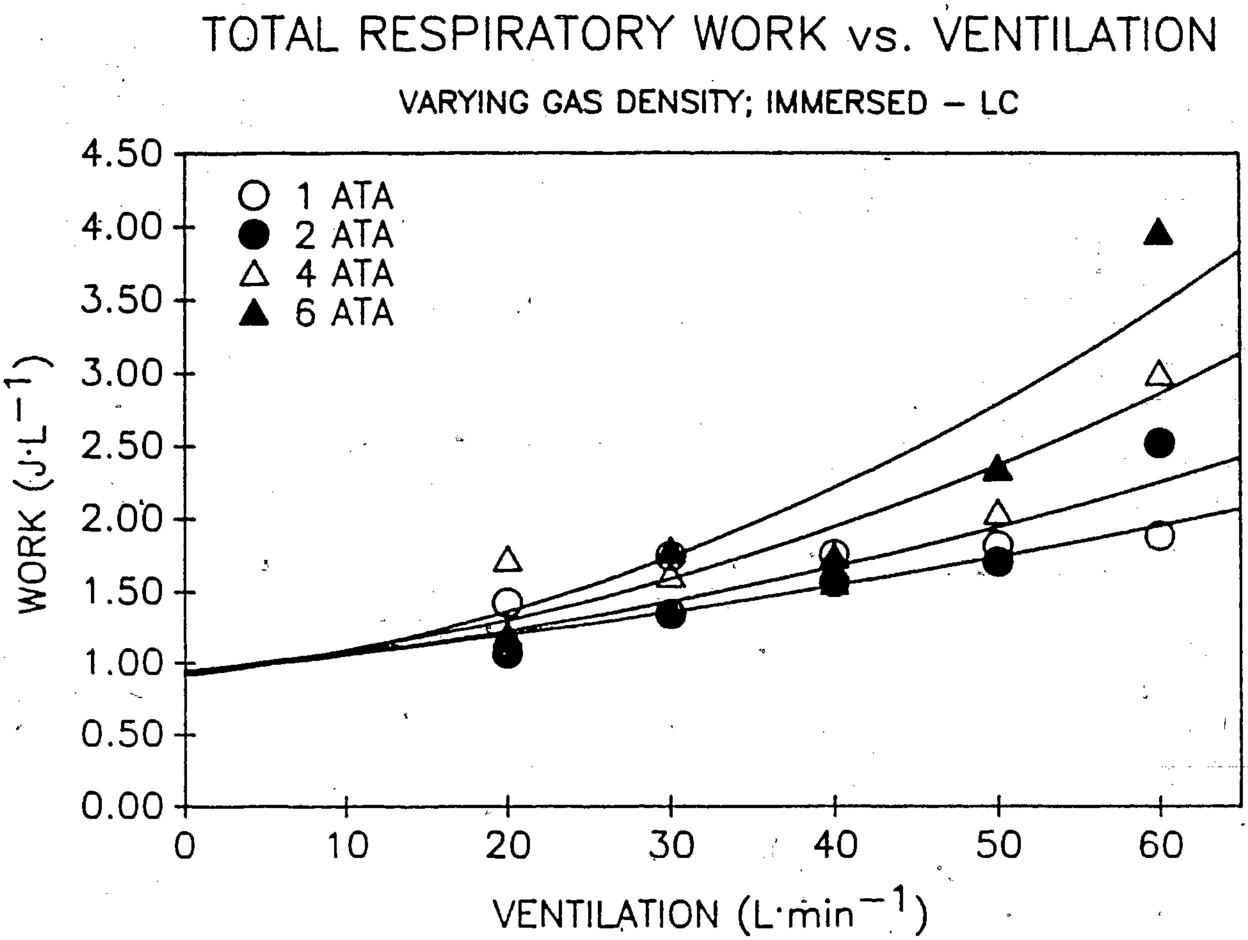


Figure 4.13: Total respiratory work with increasing minute ventilation at 1, 2, 4, and 6 ATA. Immersed – lung centroid.

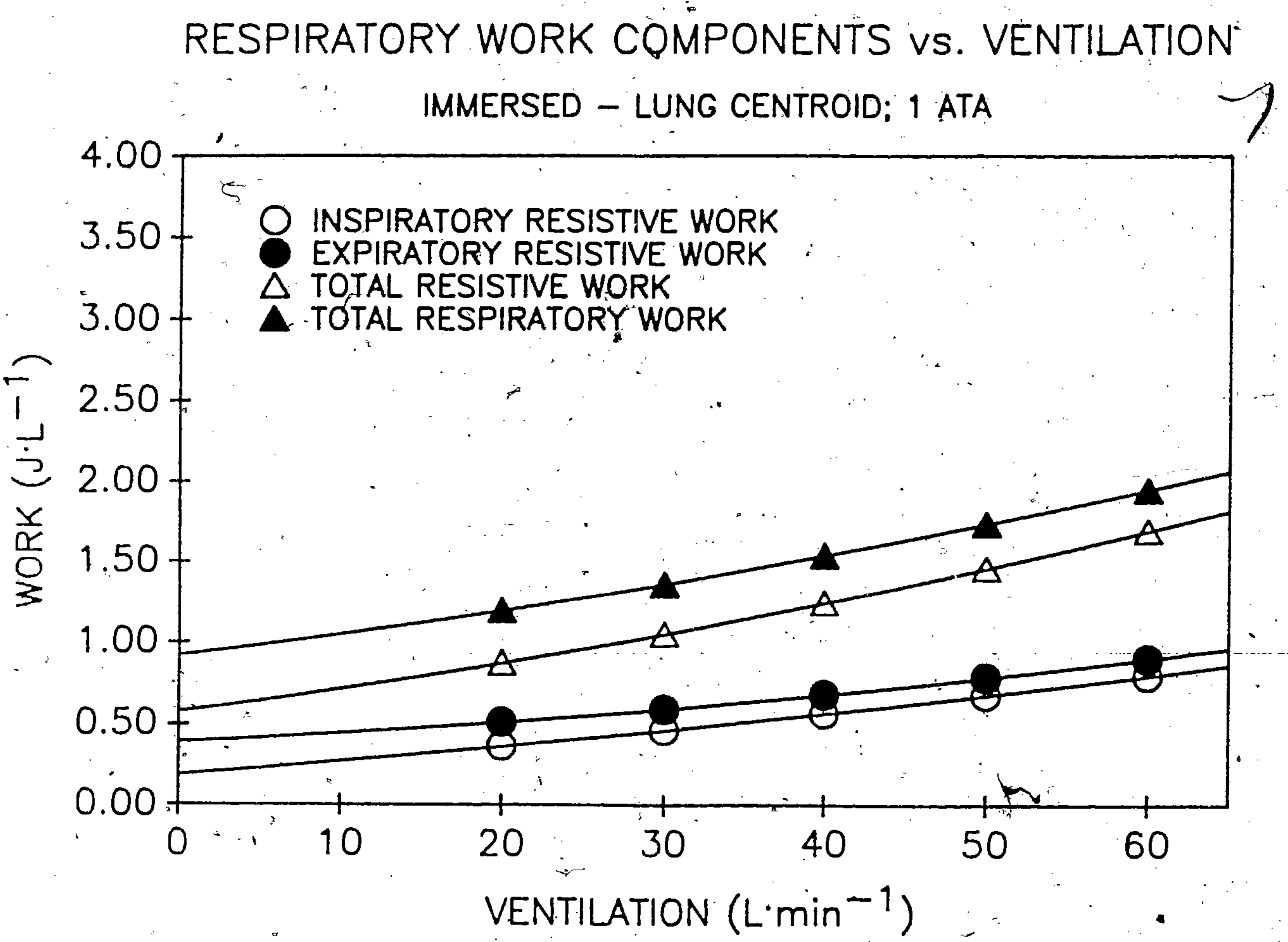


Figure 4.14: Respiratory work components with increasing minute ventilation at 1 ATA: immersed-lung centroid. Theoretical relationship derived from the experimental data.

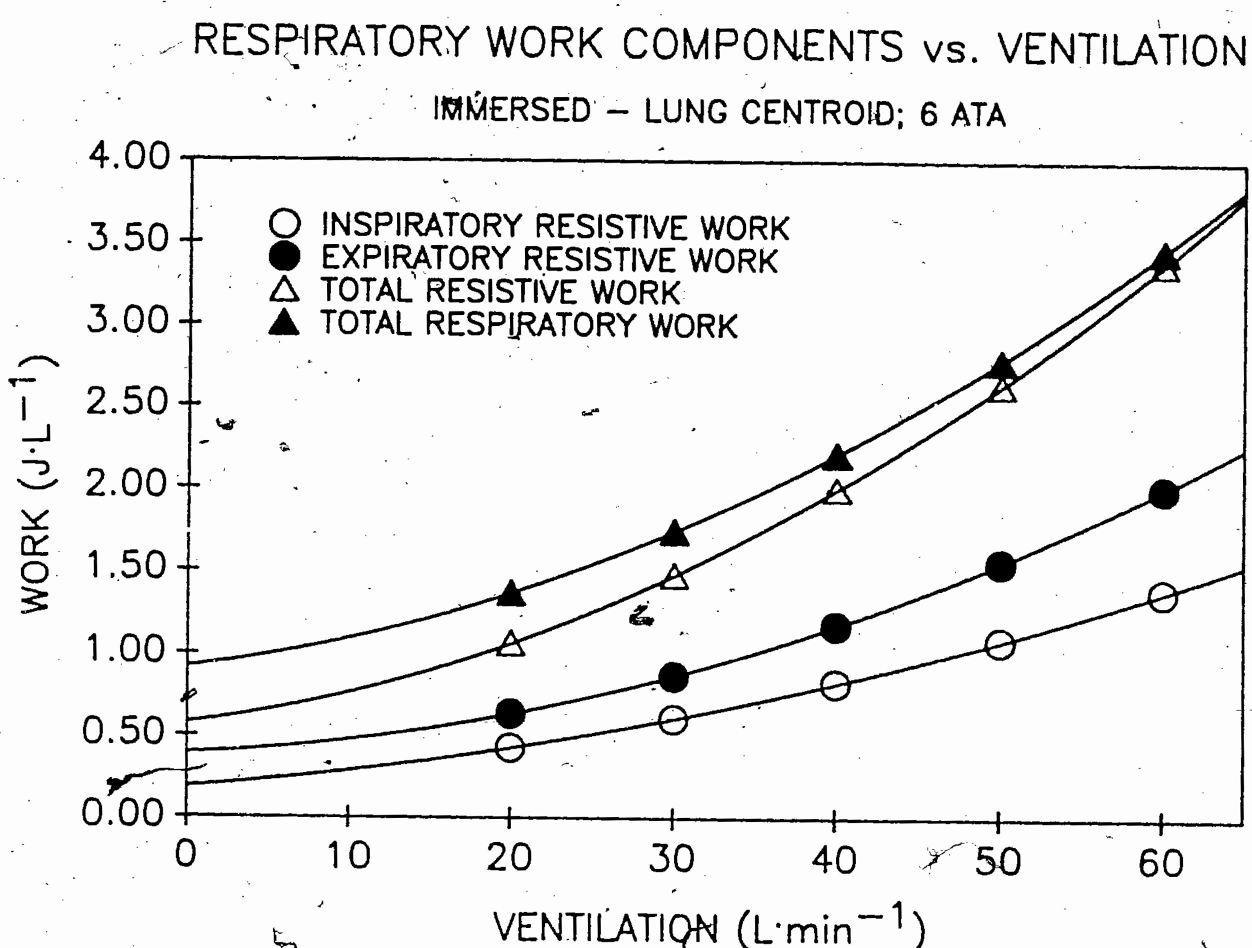


Figure 4.15: Respiratory work components with increasing minute ventilation at 6 ATA: immersed-ling centroid. Theoretical relationship derived from experimental data.

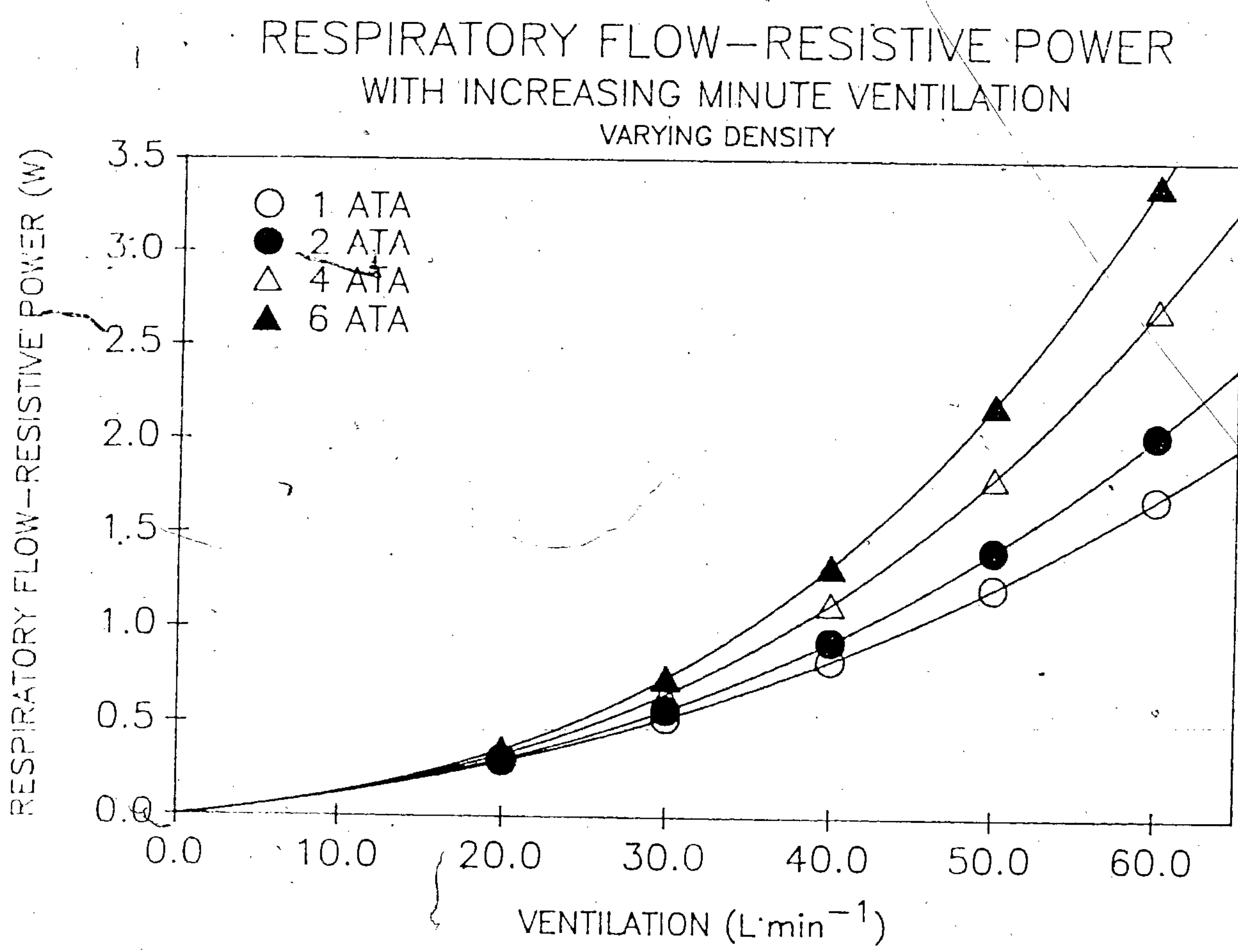


Figure 4.16: Flow-resistive power with increasing minute ventilation at 1, 2, 4, and 6 ATA.

CHAPTER 5

RESULTS: RESPIRATORY RESISTANCE AND COMPLIANCE

5.1 Respiratory Resistance

Resistance was calculated from the experimental data and its variance examined with three variables: gas flow rate, lung volume, and gas density. The data of only four of the five subjects was used to calculate the results due to the non-physiological behavior of the resistances provided by one subject.

Respiratory resistance is the result of resistive pressure divided by the rate of flow. Flow rates of ±3.0 $L.sec^{-1}$ are not uncommon with respect to human respiratory function, at moderate to high levels of work and exercise. Since resistance values at high flow rates were not observed for all conditions of density and lung volume for all subjects, only those values measured at flows of ±1.0 $L.sec^{-1}$ and ±2.0 $L.sec^{-1}$ were analyzed.

5.1.1 Inspiratory Resistance: Density Trials

Three separate analyses of variance were run on the respiratory resistance data calculated at 1, 2, 4, and 6 ATA in the immersed-lung centroid condition. The first was with respect to density itself, the second to lung volume and the third to rate of flow. With density as the only factor, the analysis of variance indicated that a difference in resistances did not exist with the different densities at $p<.05$.

Four lung volumes, 1000 mL, 500 mL, 0 mL, and -500 mL, were used in the second analysis of variance since these were the only four volumes that had data for all subjects across all densities and selected flow rates. The results of the analysis of variance indicate a significant effect of lung volume on values of resistance ($p<.0005$). *Post hoc* analysis showed high levels of lung volume to have significantly lower resistances than low lung volumes ($p<.05$), but no difference was shown between lung volumes of 0 mL and -500mL or 500mL and 1000mL. The volume of 0 mL is the immersed unoccluded relaxation volume and actually represents an absolute lung volume of approximately 3700 mL.

The third analysis of variance compared the two rates of flow and found no difference to exist between them, when all densities and lung volumes were collapsed into a single cell.

A fourth analysis of variance tested the effects on resistance of all three factors simultaneously. A main effect in resistance with lung volume was found ($p<.075$). The other two factors, flow and density, were non-significant when considered alone, but showed significance as an interaction ($p<.05$). An interaction between all three variables was also identified ($p<.10$), suggesting that the variables do not effect resistance independently.

The results of the grouped analysis of variance lends support to the theoretically modelled regression equation which incorporates all lung volumes and flow rates for which data was collected. Individual subject regression coefficients are found in Appendix C. When grouped together, the coefficients for the predictor variables were calculated as follows:

$$R = [0.987 + (0.071 \cdot \dot{V} \cdot D)] \cdot V^{-1} \qquad (5.1)$$

$$r_a^2 = 0.79 \qquad (r_a = 0.89)$$

Figure 5.1 illustrates the theoretically based interactive relationship of inspiratory resistance with gas density, lung volume, and air flow rate at a selected flow rate of 1.0 $L.sec^{-1}$.

5.1.2 Expiratory Resistance: Density Trials

The three variables, density, flow rate and lung volume were treated with three separate analyses of variance in the same manner as the inspiratory resistance data. Values of resistance were subject to a density effect ($p<.0001$). *Post hoc* analysis indicate that significant differences exist between resistances at 6 ATA and 1 ATA, and 6 ATA and 2 ATA ($p<.05$). The second analysis of variance also showed a

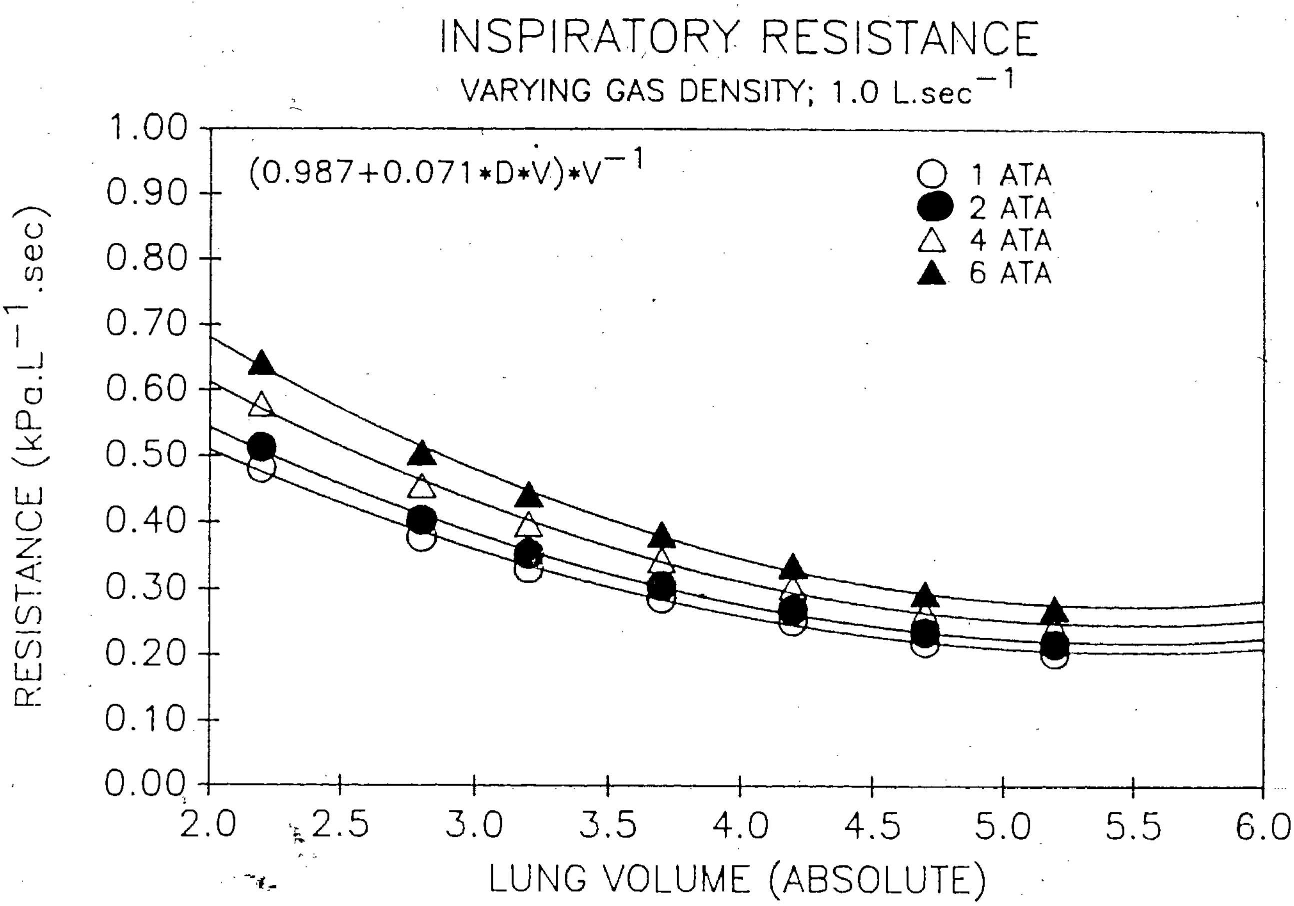


Figure 5.1: The theoretically based interactive relationship of inspiratory resistance with gas density, lung volume and air flow rate at a flow rate of 1.0 $L.sec^{-1}$.

significant effect in terms of differences in resistance values with lung volume changes ($p<.0001$). *Post hoc* tests showed that resistance tends to increase as lung volume decreases, with a significant difference found between resistances at lung volumes of 1000mL and -500mL. As with the inspiratory resistive data, flow rate when viewed as an independent variable did not produce significant differences in the resistance values. A fourth analysis of variance applied to a full model incorporating all three variables exhibited significant main effects on resistance with both lung volume ($p<.05$) and flow ($p<.005$), but not with respect to density. Further, an interaction between the two significant variables was indicated ($p<.05$). Applying the theoretically derived regression equation to all expiratory resistance data to obtain estimates of the coefficients of the predictor variables resulted in the following:

$$R = [1.258 + (0.122 \cdot \dot{V} \cdot D)] \cdot V^{-1} \qquad (5.2)$$

$$r_a^2 = 0.78 \qquad (r_a = 0.88)$$

Figure 5.2 illustrates graphically the theoretical relationship of expiratory resistance with air flow, gas density, and lung volume.

A paired t-test was run on the resistance data to test whether the observed differences with inspiration and expiration were significant. The results of the test indicate that a significant difference does exist between the two phases of respiration with regard to resistance ($p<.05$). Table 5.1 displays the mean resistances at a rate of flow of 1.0 $L.sec^{-1}$. Resistances at three lung volumes are presented at 1 ATA and 6 ATA respectively.

5.1.3 Inspiratory and Expiratory Resistance at 1 ATA

The four experimental conditions investigated at an ambient pressure of 1 ATA were analysed for differences between conditions with inspiratory and expiratory resistance as the dependent variables. An analysis of variance was performed on inspiratory and expiratory data in which two flow rates (1.0 $L.sec^{-1}$ and 2.0 $L.sec^{-1}$) at

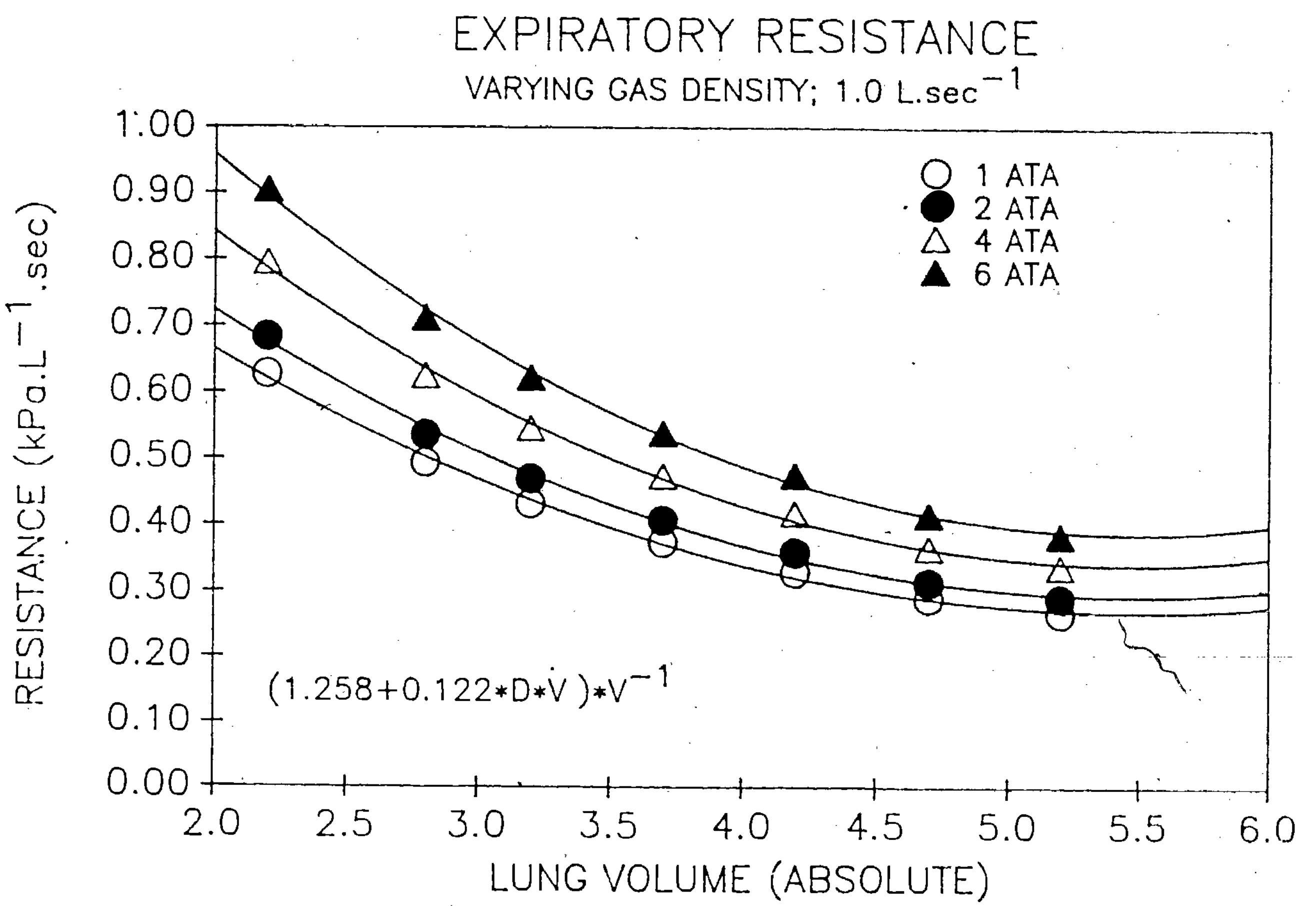


Figure 5.2: The theoretically based interactive relationship of expiratory resistance with gas density, lung volume and air flow rate at a flow rate of 1.0 L.sec^{-1}.

Table 5.1: **Mean subject resistances at 1.0 L.sec**$^{-1}$.

LUNG VOLUME (Absolute)	1 ATA Inspiration	Expiration	6 ATA Inspiration	Expiration
4.3 L	0.11	0.12	0.20	0.14
2.8 L	0.37	0.65	0.47	1.03

three lung volumes (0 mL, 500mL, and 1000mL) of four subjects composed each condition cell. Significant differences were found for both inspiratory resistance ($p<.005$) and expiratory resistance ($p=.05$).

During the inspiratory phase, *post hoc* analysis performed on the data indicated that differences in resistance existed between the dry condition and immersed-mouth ($p<.001$) condition. Dry and immersed-lung centroid conditions exhibit no significant difference. A difference was found between immersed-lung centroid and immersed-mouth conditions ($p<.005$). No significant difference was found when comparing the wetsuit condition to the immersed-lung centroid (swimsuit) condition.

Expiration produced less distinct differences between the four experimental conditions. The wetsuit condition tended to exhibit higher expiratory resistances than the immersed-lung centroid (swimsuit) condition ($p<.10$).

5.2 Respiratory Compliance

The values of dynamic compliance, calculated for each subject at the four experimental conditions (1 ATA), are presented in table 5.3. The static compliance collected via spirometric techniques in the dry condition is also included. An analysis of variance performed on the dynamic compliance data (not including wetsuit trials) indicated that significant differences do exist with compliance between conditions ($p<.01$). An *post hoc* test of the data showed that these differences exist between dry and immersed-mouth trials ($p<.01$), and immersed-mouth and immersed-lung centroid trials ($p<.05$). No significant difference was found with the dry and immersed-lung centroid trial comparisons. Comparing the two immersed-lung centroid cases, swimming trunks and wetsuit, the analysis of variance indicates that a significant difference does exist with the dynamic compliances ($p<.05$).

Figure 5.3 illustrates the mean values of dynamic compliance for the four conditons at 1 ATA in addition to the mean static compliance value.

Table 5.3: **Static and Dynamic Compliance at 1 ATA.**

		DYNAMIC COMPLIANCE			STATIC COMPLIANCE
SUBJECT NUMBER	DRY	IMMERSED MOUTH	IMMERSED LUNG CENTROID	WETSUIT	SPIROMETER
1	1.01	0.85	0.95	0.88	1.31
2	1.07	0.75	1.13	0.70	1.29
3	1.17	1.03	1.26	0.74	0.91
4	0.98	0.78	1.02	0.62	0.61
5	0.76	0.66	0.66	0.56	0.69
Average	1.00±0.15	0.81±0.14	1.00±0.22	0.70±0.12	0.96±0.33

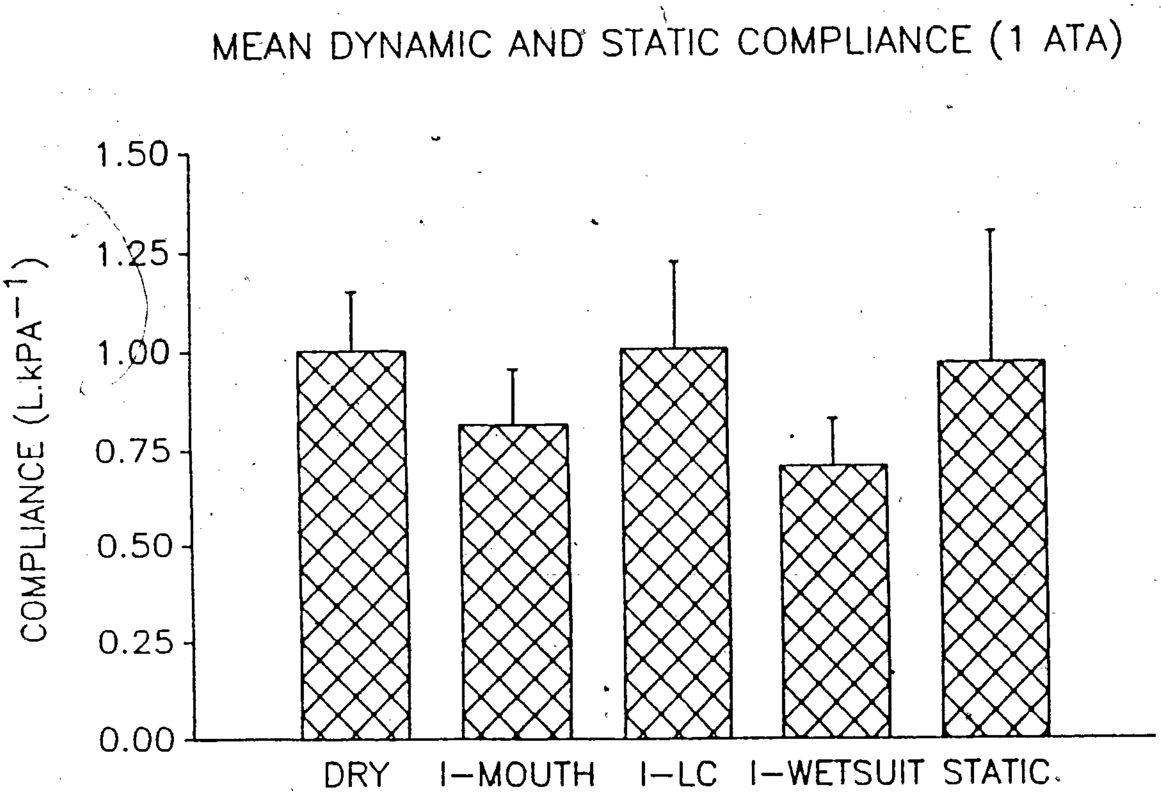


Figure 5.3: Mean dynamic and static compliance at 1 ATA.

5.3 Repetition of Dry Trials at 1 ATA

Trials performed in the dry experimental condition were repeated after the completion of all other test procedures to evaluate if a learning or habituation response did exist. Paired t-tests performed on the two conditions indicated that no significant difference did exist.

CHAPTER 6

DISCUSSION

6.1 Respiratory Work

Analysis of the flow-resistive respiratory work data from the present study showed an interaction to exist between the rate of ventilation and gas density. This lends some support to the theory of laminar and turbulent flow contributions to overall pressure loss across the system.

Regression analyses indicated that the proposed relationship relating flow-resistive work with ventilation and gas density explained approximately 82% of the data, across all work types, when all subjects were considered together. Some of the individual subject data were fitted by the equation better than others, indicating either the need for the inclusion of yet another physiological variable to the predictive model of respiratory work, or a varying contribution of random error in the quality of the subject relaxation.

A constant coefficient was added to the present theoretical model due to the fact that the resistive work data of all subjects appeared to miss the origin when the best fit curve was extrapolated to the ordinate (zero ventilation). In fact, values of work obtained at a ventilation rate of 20 $L.min^{-1}$ were often equal to or greater than those obtained at the adjacent ventilation rate of 30 $L.min^{-1}$. These data forced the regression curves upwards, suggesting the ability to obtain a resistive work value other than zero when no airflow is present.

Why resistive work was raised at the lowest rate of ventilation is not clear. One possibility stems from the slow cyclical rate of the breathing machine. With the tidal volume fixed at two litres, the breathing frequency was adjusted to ten breaths per minute to produce a ventilation rate of 20 $L.min^{-1}$. The frequency may have been too slow for the subjects, causing difficulty in remaining completely relaxed during the mechanical ventilation. Further, hypocapnia effects smooth muscle tone. It is possible that hypocapnia occured at the 20 $L.min^{-1}$ rate of mechanical ventilation. Alternatively, the constant term k_0 may reflect the presence or respiratory hysteresis, or a flow-independent friction component resisting movement of the thorax.

Pressure-volume curves for inflation and deflation are different when constructed over a moderate range of volume (Cotes, 1979). This pulmonary hysteresis, according to Cotes (1979) may be the result of surface tension. He suggests the the linear increase in volume with applied pressure in the middle section of the pressure-volume curve with inspiration represents work done in expanding the surface film which lines the alveoli. The difference in pressure with a given volume during expiration is therefore partially due to reduced surface tension.

Noting the values of the coefficients for the different work components with the density conditions, it is observed that expiratory flow-resistive work exceeds inspiratory flow-resistive work (Figures 4.14 and 4.15). This finding is similar to that of Taylor (1987) who reported greater pulmonary resistive work with expiration at 1 ATA. Referring back to the resistance findings of the present study, expiratory resistance was shown to have a greater turbulent flow contribution than inspiratory resistance. The energy cost of airflow turbulence is magnified with increased density. Therefore, if expiratory resistance is larger than inspiratory resistance due to increased air turbulence, it would be expected that expiratory flow-resistive work would also exceed inspiratory flow-resistive work.

Total respiratory work in normal men was calculated by Sharp *et al.* (1964) to be approximately 0.073 $kg.m.L^{-1}$ (0.716 $J.L^{-1}$) at rest in dry conditions. The present study's total respiratory work mean is higher than the former investigation, at 1.226 $J.L^{-1}$ with dry conditions (1 ATA), at a rate of ventilation of 20 $L.min^{-1}$.

The contribution of elastic work to total respiratory work, at 20 $L.min^{-1}$, was calculated to be 58% in the dry environment. The remaining 42% is therefore assumed to be flow-resistive work. These values are similar to those reported in the literature. Otis *et al.* (1950), using mechanical ventilation as the measurement technique, reported the contributions of elastic work and flow–resistive work to be 63% and 37% respectively. McIlroy *et al.* (1954) reported a slightly higher elastic work contribution of 70%. Finally, Attinger and Segal (1959) reported flow–resistive work to contribute 38% to the total work of respiration.

With immersion, at rest, inspiratory flow–resistive work increased 81% from 0.43 $J.L^{-1}$ to 0.78 $J.L^{-1}$. Expiratory flow–resistive work increased from 0.47 $J.L^{-1}$ to 0.63 $J.L^{-1}$ with immersion, representing an increase of 34% at rest. The increase in both inspiratory and expiratory flow resistive work is far less than that reported by Taylor (1987). Inspiratory flow–resistive work increased from 0.071 $J.L^{-1}$ to 0.285 $J.L^{-1}$ and expiratory flow–resistive work increased from 0.127 $J.L^{-1}$ to 0.462 $J.L^{-1}$ with immersion. Taylor, however, made his measurements with the subject fully immersed as opposed to neck immersion in the present study. Therefore, the differences in the results would be expected. Secondly, Taylor measured only pulmonary flow–resistive work, whereas the present study included the chest wall. Thus changes in airway resistance would have a relatively smaller effect on the overall flow–resistive work in this study.

The relative increase in inspiratory flow–resistive work is greater than that reported by Hong *et al.* (1969). Hong and his co–workers reported the increase induced by a change from immersion to the xiphoid process to immersion at the neck. The smaller percentage increase of 57.4% can therefore be explained due to the smaller hydrostatic pressure change experienced.

Jarrett (1965), Craig and Dvorak (1975), and Flynn *et al.* (1975) all postulated improvements in resistive work during upright immersion with manipulations of breathing pressure. Taylor (1987) confirmed this hypothesis with breathing gas pressures supplied at P_{LC} and $P_{LC+0.98\ kPa}$. The results of the present study support the findings of Taylor (1987). With breathing gas supply at P_{LC}, significant improvements were found with inspiratory and total flow–resistive work components. Although expiratory flow–resistive work was also expected to improve with breathing gas supplied at P_{LC}, the results indicated a non–significant difference to exist. This non-significance may be due to non-linearity of the compliance curve at low lung volumes with uncompensated immersion. Since compliance was assumed to be linear in the calculations, inspiratory work may have been over-estimated and expiratory work underestimated as a direct result. This theory is supported with the density conditions since expiratory flow-resistive work exceeds inspiratory flow-resistive work at all gas densities with breathing gas supplied at P_{LC}.

6.2 Power

Flow-resistive power has been studied by several investigators, but the exact nature of the relationship with rate of ventilation is not conclusive. Linear, curvilinear, and exponential functions have all been suggested (Otis *et al.*, 1950; McIlroy *et al.*, 1954; Fritts *et al.*, 1959; Holmgren *et al.*, 1973; Taylor, 1987). Figure 6.1 illustrates the respiratory flow-resistive power results of several investigators. Data is available only at 1 ATA. Note that the two studies that include chest wall resistance, Otis *et al.* (1950) and the present study, produce power curves with greater slopes. This suggests that the chest wall resistance is a significant force to be overcome with respiratory work, particularly at high rates of ventilation. From the present study, it may be suggested that 25% to 50% of respiratory power is used to overcome chestwall resistance. The work of Otis *et al.* (1950) suggests a much greater contribution of respiratory power to overcome chestwall resistance. The difference found in the two studies may possibily be accounted for by the methods of mechanical ventilation used. A Drinker respirator encloses the entire body up to the neck while the subject lies supine. The air pressure within the tank is altered, applying forces externally to the body. The breathing machine used in the present study allowed the subject to sit upright while the ventilator mechanically ventilated the lungs and airways. When supine, the lungs possess a lower lung volume. This decreases lung compliance and increases airflow resistance. These effects of posture may partially explain the difference in respiratory power values between the two mechanical ventilation studies.

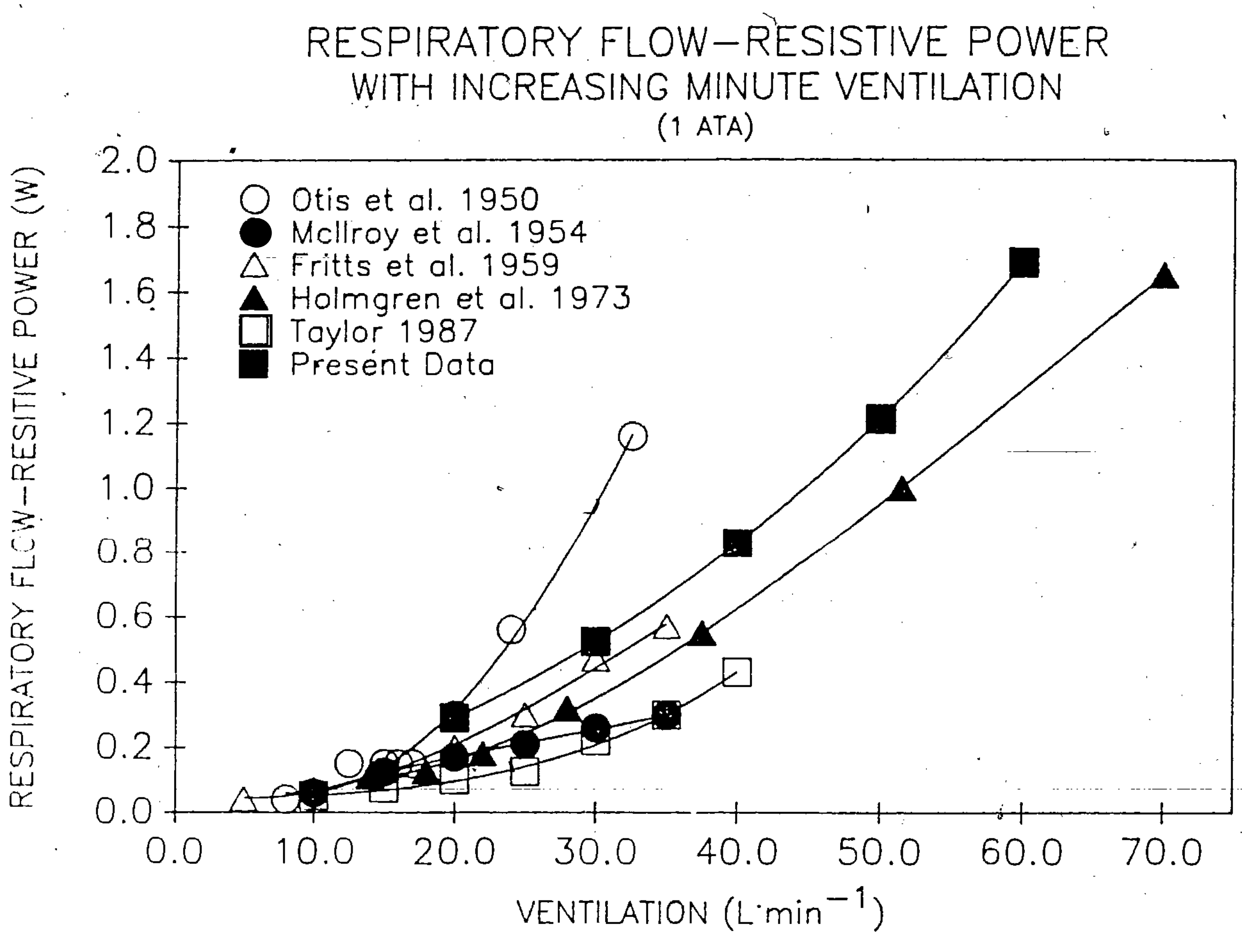


Figure 6.1: Respiratory flow-resistive power with increasing minute ventilation: the results of six studies.

6.3 Respiratory Resistance

The results of the present study support the theoretically derived equation (equation 1.7) relating resistance to gas flow rate, density, and lung volume. Resistance was found vary with the inverse of lung volume. This finding is in agreement with the work of McKenna *et al.* (1973).

Inspiratory and expiratory resistances were shown to differ. The results of the regression analysis indicated expiratory resistance has a larger laminar and turbulent component than inspiratory resistance by 27% and 72%, respectively. This finding, based on the estimated regression coefficients, suggests that turbulent airflow is more prevelant with expiration than with inspiration. Hence, airway diameters are smaller in expiratory flow.

The modelled regression equation fitted to the calculated values of inspiratory and expiratory resistance explained, on average, 89% and 88% of the data, respectively. A better model might be one that allows the effect of lung volume to differ with the laminar and turbulent flow components. A value of V^{n}, where n is constrained between 1.0 and 2.0 may be more appropriate for the turbulent flow component when considering the modelled equation for turbulent flow (Equation 1.4).

With immersion, the average resistance calculated at a flow rate of 1.0 $L.sec^{-1}$ increased 100% during inspiration, but did not change during expiration. Mean respiratory resistance therefore was found to increase by 50%. The relative increase of respiratory resistance with immersion appears to fit well with results reported in the literature (refer to Table 2.2). Most of the studies have reported results for pulmonary resistance only. Lollgen *et al.* (1980), measured total respiratory system resistance and therefore their results may be directly compared with the present results. Lollgen reported average resistance to increase 57.4% with immersion. Taylor (1987) reported inspiratory and expiratory resistances, calculated at a mean flow rate of 0.5 $L.sec^{-1}$ to increase by 125% and 185% respectively with immersion. Total pulmonary resistance increased 155% with immersion. Taylor's resistance values represent pulmonary resistance only, unlike the present study which also includes chestwall resistance. Hence, the large difference in resistances can be attributed to the difference in system compliance.

Sharp *et al.* (1964) measured average respiratory resistance to be 0.49 $kPa.L^{-1}.sec$ in normal men. The present study reports a similar average resistance, at 1.0 $L.sec^{-1}$ and at relaxation volume, of 0.46 $kPa.L.sec^{-1}$. This resistance was obtained by taking the mean of the inspiratory and expiratory resistances with the given conditions. Both investigations employed similar mechanical measurement techniques.

6.4 Elastic Work and Compliance

Elastic work must be expended during inspiration in order to deform respiratory tissue (lung and chest wall). This energy is stored as potential energy and thus is available for assistance during expiration. Moderate to high rates of ventilation, high gas densities and external resistance increase flow-resistive work. When it exceeds inspiratory elastic work, expiration becomes an active process.

According to Morrison and Reimers (1982), when a subject is immersed, a hydrostatic pressure is applied to the thorax, disrupting the equilibrium established in air. A new relaxation volume is found at which the hydrostatic imbalance is compensated by the elastic recoil of the system. As a result, the pressure-volume curve is shifted to a new position. This rightward shift increases the amount of elastic work which must be expended during inspiration due to the non-linearity of the compliance curve (Agostoni *et al.*,1966; Craig and Ware, 1967; Hong *et al.*, 1969; Jarrett, 1965; Morrison and Reimers, 1982). In a later study Morrison *et al.* (1987) suggested that the hydrostatic imbalance is only partially compensated by elastic recoil during immersion, as subjects actively defend an ERV above relaxation volume. This effect limits the extent of change in system compliance.

Robertson *et al.* (1978) emphasized the blood shift into the thorax with immersion due to the compressive effect of water on the blood vessels and the extremities. Both TLC and VC are reduced as a result (Agostoni *et al.*, 1966; Arborelius *et al.*, 1972; Robertson *et al.*, 1978). Provision of breathing gas supply at P_{LC} shifts the pressure-volume curve to the left, returning relaxation volume to its former position in air. Inspiratory elastic work is therfore reduced.

The comparison of dynamic compliance to static compliance (dry) provides a test of validity of the mechanical ventilation technique. If the respiratory muscles were active during the mechanical ventilation, measures of dynamic compliance would be effected. As an example, inspiratory muscle tone at end-inspiration would give reduced compliance, as present in voluntary respiration. Furthermore, the dynamic compliance was not affected by breathing frequency as no change was observed. This finding is in agreement with the study of Woolcock *et al.*, 1969, whose methods did not include the use of mechanical ventilation.

The values of compliance found in the present study agree with the findings of Sharp *et al.* (1964). Both studies found average total respiratory compliance to be approximately 0.10 $L.kPa^{-1}$. Both dry conditions and immersion with hydrostatic pressure compensation of breathing gas supply at P_{LC} showed very similar respiratory compliances, while the uncompensated immersion condition displayed a notably reduced mean dynamic compliance (81% of dry value).

An interesting observation is made when comparing mean dynamic compliance in the wetsuit condition with the uncompensated immersion condition. Compliances for the two conditons are very similar. This suggests that the addition of a wetsuit eliminates the positive effects of pressure compensation on respiratory compliance. The elastic properties of the neoprene compress the chest wall, lowering the relaxation volume to a new point of equilibrium. Inspiratory elastic work should subesquently be increased due to two factors: decreased respiratory compliance and the additional wetsuit compliance. Respiratory flow-resistive work would also be expected to increase as a result of reduced lung volume. Thus, it might be suggested that when wearing a wetsuit, divers should be supplied with breathing gas at a pressure above $\bar{P}_{LC}$ in order to obtain optimum respiratory mechanics. Although a significant difference was not found in elastic work with the wetsuit and swimsuit conditions due to the large variance in the swimsuit work values, the tendency was for the wetsuit elastic work to be higher. Inspiratory and total flow-resistive works were significantly greater while wearing the wetsuit than when wearing only the swimsuit.

CHAPTER 7

CONCLUSIONS

Three objectives and three hypothesis were proposed for the present study at its onset. Through the experimental and analytical processes, the three objectives were achieved and the three hypotheses supported.

Flow-resistive work of breathing increased curvilinearly with increased minute ventilation, according to prescribed theoretical equations. Raised gas density was shown to increase flow-resistive respiratory work in a multiplicative manner within the turbulent flow component. Elastic work was shown to remain constant across minute ventilations and densities. Expiratory flow-resistive work was approximately double inspiratory flow-resistive work.

Immersion without hydrostatic pressure compensation was shown to increase both elastic and flow-resistive work. Dynamic respiratory compliance was reduced by 20%. Provision of breathing gas at a supply pressure of P_{LC} allowed for considerable improvement in respiratory work components moving them back towards their levels in dry conditions. Dynamic respiratory compliance was also restored to its preimmersion state.

A common diving apparel, the wetsuit, appeared to counteract some of the benefits of breathing gas supply at P_{LC} by lowering dynamic compliance to its position with uncompensated immersion. Inspiratory flow-resistive work, total flow-resistive work, and total respiratory work all were significantly larger than the corresponding swimsuit trials.

Respiratory resistance was affected by gas density, flow rate, and lung volume. As gas density and/or flow rate rise, so does resistance. As lung volume was lowered, resistance was increased. The manifestation of inspiratory resistance was shown to be different from expiratory resistance. The latter had a 67% larger turbulent flow contribution than the former, and was therefore more strongly influenced by increases in gas density and flow rate.

The similarity of results to those reported in the literature, in addition to the similarity of dynamic and static compliance, supports mechanical ventilation as a useful research technique. Its non-invasive manner may make its employment as a research tool more common, particularly in alien environments and allow a greater number of divers to be studied in the context of respiratory mechanics and physiology. The greater the acquisition of knowledge, the greater the chance of its application in the design of power assisted breathing apparatus. Resistance and work limits can be employed in the design process, leading to an improvement in the safety of the underwater work environment.

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APPENDIX A

Table A1: Respiratory Capacities, L (BTPS)

SUB #	VC	IC	ERV	FEV_1
1	5.594	3.243	2.657	3.892
2	5.459	3.648	1.757	4.351
3	5.756	3.628	2.564	4.675
4	6.216	4.729	2.660	4.513
5	5.589	4.432	2.387	4.644

APPENDIX B

Table B1: Regression coefficients and variance for inspiratory flow-resistive work.

SUB #	β_0	β_1	β_2	R_a^2
1	-0.062 ± 0.224	0.0116 ± 0.0062	0.000022 ± 0.000014	0.459
2	0.245 ± 0.110	0.0022 ± 0.0032	0.000024 ± 0.000008	0.498
3	0.158 ± 0.104	-0.0002 ± 0.0030	0.000060 ± 0.000008	0.847
4	0.222 ± 0.296	0.0225 ± 0.0088	0.000025 ± 0.000024	0.427
5	0.366 ± 0.105	0.0050 ± 0.0031	0.000030 ± 0.000008	0.690
Ave	0.186 ± 0.326	0.0063 ± 0.0140	0.000032 ± 0.000028	0.797±0.090

Table B2: Regression coefficients and variance for expiratory flow-resistive work.

SUB #	β_0	β_1	β_2	R_a^2
1	0.462 ± 0.127	0.0001 ± 0.0035	0.000067 ± 0.000008	0.871
2	0.350 ± 0.088	-0.0020 ± 0.0026	0.000065 ± 0.000007	0.888
3	0.584 ± 0.163	-0.0049 ± 0.0048	0.000058 ± 0.000012	0.607
4	0.205 ± 0.386	0.0322 ± 0.0115	0.000007 ± 0.000031	0.427
5	0.366 ± 0.195	-0.0016 ± 0.0058	0.000110 ± 0.000015	0.830
Ave	0.394 ± 0.332	0.0048 ± 0.0210	0.000061 ± 0.000051	0.863±0.108

Table B3: Regression coefficients and variance for total flow-resistive work.

SUB #	β_0	β_1	β_2	R_a^2
1	0.397 ± 0.269	0.0118 ± 0.0074	0.000089 ± 0.000017	0.797
2	0.589 ± 0.127	0.0004 ± 0.0038	0.000088 ± 0.000010	0.889
3	0.753 ± 0.236	-0.0059 ± 0.0069	0.000121 ± 0.000018	0.787
4	0.421 ± 0.656	0.0547 ± 0.0196	0.000032 ± 0.000053	0.477
5	0.726 ± 0.280	0.0035 ± 0.0083	0.000140 ± 0.000021	0.814
Ave	0.577 ± 0.479	0.0129 ± 0.0334	0.000094 ± 0.000065	0.880±0.085

Table B4: Regression coefficients and variance for total respiratory work.

SUB #	β_0	β_1	β_2	R_a^2
1	0.692 ± 0.302	0.0077 ± 0.0084	0.000104 ± 0.000019	0.780
2	0.761 ± 0.114	0.0016 ± 0.0034	0.000078 ± 0.000009	0.891
3	1.093 ± 0.258	-0.0059 ± 0.0075	0.000105 ± 0.000019	0.690
4	1.277 ± 0.770	0.0456 ± 0.0230	0.000014 ± 0.000063	0.255
5	0.772 ± 0.277	0.0116 ± 0.0082	0.000116 ± 0.000021	0.798
Ave	0.920 ± 0.597	0.0121 ± 0.0230	0.000084 ± 0.000068	0.683±0.250

APPENDIX C

Table C1: Regression coefficients and variance for inspiratory resistance.

SUBJECT	β_0	β_1	R_a^2
1	1.040 ± 0.111	0.0327 ± 0.0213	0.848
2	0.745 ± 0.160	0.0630 ± 0.0289	0.684
3	0.620 ± 0.155	0.1220 ± 0.0290	0.736
5	1.542 ± 0.129	0.0678 ± 0.0247	0.884
Ave	0.987 ± 0.549	0.0714 ± 0.0632	0.788±0.094

Table C2: Regression coefficients and variance for expiratory resistance.

SUBJECT	β_0	β_1	R_a^2
1	1.572 ± 0.201	0.1319 ± 0.0377	0.829
2	0.389 ± 0.144	0.2040 ± 0.0258	0.838
3	1.038 ± 0.131	0.0679 ± 0.0242	0.836
5	2.031 ± 0.366	0.0839 ± 0.0696	0.616
Ave	1.258 ± 0.917	0.122 ± 0.100	0.780 ± 0.109