National Library of Canada
Bibliothèque nationale du Canada

Canadian Theses Service
Services des thèses canadiennes

Ottawa, Canada
K1A 0N4

CANADIAN THESES

THÈSES CANADIENNES

NOTICE

The quality of this microfiche is heavily dependent upon the quality of the original thesis submitted for microfilming. Every effort has been made to ensure the highest quality of reproduction possible.

If pages are missing, contact the university which granted the degree.

Some pages may have indistinct print especially if the original pages were typed with a poor typewriter ribbon or if the university sent us an inferior photocopy.

Previously copyrighted materials (journal articles, published tests, etc.) are not filmed.

Reproduction in full or in part of this film is governed by the Canadian Copyright Act, R.S.C. 1970, c. C-30.

AVIS

La qualité de cette microfiche dépend grandement de la qualité de la thèse soumise au microfilmage. Nous avons tout fait pour assurer une qualité supérieure de reproduction.

S'il manque des pages, veuillez communiquer avec l'université qui a conféré le grade.

La qualité d'impression de certaines pages peut laisser à désirer, surtout si les pages originales ont été dactylographiées à l'aide d'un ruban usé ou si l'université nous a fait parvenir une photocopie de qualité inférieure.

Les documents qui font déjà l'objet d'un droit d'auteur (articles de revue, examens publiés, etc.) ne sont pas microfilmés.

La reproduction, même partielle, de ce microfilm est soumise à la Loi canadienne sur le droit d'auteur, SRC 1970, c. C-30.

THIS DISSERTATION HAS BEEN MICROFILMED EXACTLY AS RECEIVED

LA THÈSE A ÉTÉ MICROFILMÉE TELLE QUE NOUS L'AVONS REÇUE

NL-339 (r. 86/06)

Canada

EFFECT OF BREATHING GAS PRESSURE ON RESPIRATORY MECHANICS OF IMMERSED MAN.

by

NIGEL A.S. TAYLOR

Dip.T., Brisbane, 1973

B.H.M.S., University of Queensland, 1977

B.H.M.S.(Hons), University of Queensland, 1983

M.Sc., University of London 1984.

THESIS SUBMITTED IN PARTIAL FULFILLMENT OF

THE REQUIREMENTS FOR THE DEGREE OF

DOCTOR OF PHILOSOPHY

in the School

of

KINESIOLOGY

© NIGEL A.S. TAYLOR 1987

SIMON FRASER UNIVERSITY

March 1987

All rights reserved. This work may not be reproduced in whole or in part, by photocopy or other means, without permission of the author.

Permission has been granted to the National Library of Canada to microfilm this thesis and to lend or sell copies of the film.

The author (copyright owner) has reserved other publication rights, and neither the thesis nor extensive extracts from it may be printed or otherwise reproduced without his/her written permission.

L'autorisation a été accordée à la Bibliothèque nationale du Canada de microfilmer cette thèse et de prêter ou de vendre des exemplaires du film.

L'auteur (titulaire du droit d'auteur) se réserve les autres droits de publication; ni la thèse ni de longs extraits de celle-ci ne doivent être imprimés ou autrement reproduits sans son autorisation écrite.

ISBN 0-315-36367-3

APPROVAL

Name: NIGEL A.S. TAYLOR

Degree: DOCTOR OF PHILOSOPHY

Title of thesis: EFFECT OF BREATHING GAS PRESSURE ON RESPIRATORY MECHANICS OF IMMERSED MAN.

Examining Committee:

Chairman: Prof.M.V. Savage

Prof.J.B. Morrison
Senior Supervisor

Dr.T.J. Smith

Dr.D.K. Stirling

Dr.J.Road
Faculty of Medicine
University of British Columbia

Prof.N.R. Anthonisen
Faculty of Medicine
University of Manitoba
External Examiner

Date Approved: March 6, 1987

PARTIAL COPYRIGHT LICENSE

I hereby grant to Simon Fraser University the right to lend my thesis or dissertation (the title of which is shown below) to users of the Simon Fraser University Library, and to make partial or single copies only for such users or in response to a request from the library of any other university, or other educational institution, on its own behalf or for one of its users. I further agree that permission for multiple copying of this thesis for scholarly purposes may be granted by me or the Dean of Graduate Studies. It is understood that copying or publication of this thesis for financial gain shall not be allowed without my written permission.

Title of Thesis/Dissertation:

EFFECT OF BREATHING GAS PRESSURE ON RESPIRATORY MECHANICS OF IMMERSED MAN.

Author:

(signature)

NIGEL A.S. TAYLOR

(name)

12 March, 1987

(date)

ABSTRACT

It was hypothesised upright, whole-body immersion, without hydrostatic pressure compensation, would displace the respiratory compliance curve, reduce functional residual capacity (FRC), and increase respiratory work. Lung *centroid* pressure (P_{LC}) was defined as the pressure necessary to return immersed static respiratory mechanical attributes to levels observed in air. It was hypothesised that breathing air supplied at P_{LC} would restore normal lung volume subdivisions, return elastic and flow-resistive work towards normal, and permit greater workload tolerance with less respiratory distress.

Lung *centroid* pressure was determined from changes in lung relaxation volume accompanying immersion (N=17). Upright immersion caused a mean displacement of the respiratory compliance curve of +1.33kPa (SEM=0.106), relative to the hydrostatic pressure at the sternal notch. Prone immersion produced a mean displacement of -0.69kPa (SEM=0.124) (N=13). Lung *centroid* was deemed to be 13.6cm inferior, and 7.0cm posterior to the sternal notch.

Static respiratory mechanics were investigated in air (control) and water (N=10). Upright immersion produced no significant changes in lung, chest or total respiratory compliance, measured at control relaxation volume. Immersed subjects were provided with air supplied at four hydrostatic pressures: mouth pressure, (P_{LC} -0.98)kPa, P_{LC}, and (P_{LC} +0.98)kPa. Immersion with air supplied at mouth pressure reduced expiratory reserve volume by 47% ($p<0.05$) and increased inspiratory muscle work three-fold ($p<0.05$). At the other three supply pressures, inspiratory work and lung volume compartments were returned towards control status ($p<0.05$).

Dynamic pulmonary mechanics were investigated under identical conditions. When immersed, inspiratory and expiratory pulmonary work, and pulmonary resistance increased significantly ($p<0.05$), with air supplied at mouth pressure. Air provision at greater pressures returned variables towards control status sequentially with pressure increments.

Ten subjects performed graded, immersed exercise at ambient pressures of 1 and 6 ATA, breathing air supplied at mouth and lung *centroid* hydrostatic pressures. Minute ventilation, end-tidal CO_2, oro-nasal pressure, heart rate and perceived respiratory comfort were recorded. At 6 ATA subjects experienced significant bradycardia and hypoventilation ($p<0.05$). When breathing air at P_{LC} subjects found respiration to be more comfortable ($p<0.05$), while achieving a significantly greater minute ventilation ($p<0.05$). Subjects also reported being less affected by narcosis.

ACKNOWLEDGEMENTS

The author gratefully acknowledges the contributions of the following people.

Professor J.B.Morrison, without whose expertise, dedication and continual guidance, this project would not have been possible.

Mrs.E.A.Taylor for supervision of data acquisition during the immersion experiments, and for typing this manuscript.

Dr.D.Hedges for medical supervision during all invasive and hyperbaric experiments.

Messrs I.Wood, T.McColl and G.Caldwell for guidance with the computer programming.

Mr.R.Village for research assistance during the diving experiments.

Professor E.W.Banister, Dr.D.R.Stirling and Dr.T.J.Smith for their editorial comments.

Messrs R.Maskell and I.Wood for assistance with the instrumentation.

Dr.D.R.Stirling for instruction in the art of oesophageal ballooning.

Messrs V.Stobbs and G.Morariu for controlling the hyperbaric chamber during all the experiments conducted under hyperbaric conditions.

Professors B.Bake (University of Göteborg, Sweden), G.O.Dahlbäck (Chalmers University, Sweden), P.T.Macklem (McGill University, Canada) and J.Milic-Emili (McGill University, Canada) for their personal communications.

DEDICATION

TO

Liz and Skye

for their love, continual support,

understanding, encouragement and assistance,

and to the memory

of

John Francis Webb.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

CHAPTER 1

INTRODUCTION

The respiratory system has been described as a reciprocating bellows pump (Mead and Milic-Emili 1964) consisting of two concentric walls, with the outer wall housing the respiratory muscles, which power the pump. The work done, and the degree of efficiency of this pump are of prime importance. In man, respiratory work does not normally impede physical power (Asmussen and Neilsen 1960, Stubbing *et al.* 1980). Pulmonary patients have shown that power may be impaired by ventilatory inadequacy (Roussos and Macklem 1978, Bye *et al.* 1983). It has been suggested that under extremes, respiratory work may consume enough oxygen to limit that available for useful external work in normal subjects (Shephard 1966, Hesser *et al.* 1981).

Respiratory work (W_{res}[1]) is carried out to overcome intrapulmonary forces (Figure 1.1), which include elastic recoil of the lungs and rib cage. Lungs recoil towards zero volume so that inspiratory effort must exceed recoil pressure to cause inflation. The rib cage recoils towards a volume approaching 60% of the vital capacity (VC) (Agostoni and Mead 1964). Inspiration to volumes ≥60%VC requires work against rib cage recoil pressure, however, at volumes ≤50%VC, recoil is in the inspiratory direction, aiding inspiration.

A second category of respiratory forces encountered are the flow-resistive forces created by friction during respiration: airway resistance (R_{aw}), lung tissue resistance (R_{lt}), and chest wall resistance (R_{wt}).

Finally there are inertial forces. Though minor under normal ambient conditions, both gas and tissue inertia may become significant under conditions of raised gas density and elevated thoracic mass. Factors which reduce lung or rib cage compliance, or increase respiratory resistance (R_{pul}), or inertial forces will increase the internal work of respiration (W_{res}).

Standing in air, man is exposed to pressure from the environment. This pressure acts on all internal and external surfaces with which the air is in direct contact. The gradual decline in pressure with altitude renders detection of a vertical air pressure gradient over an upright man, almost imperceptible. In this case all body surfaces are essentially exposed to the same atmospheric pressure. This is not the case when man is immersed in water.

---

[1] Operational definitions and abbreviations are contained in Section 1.2.

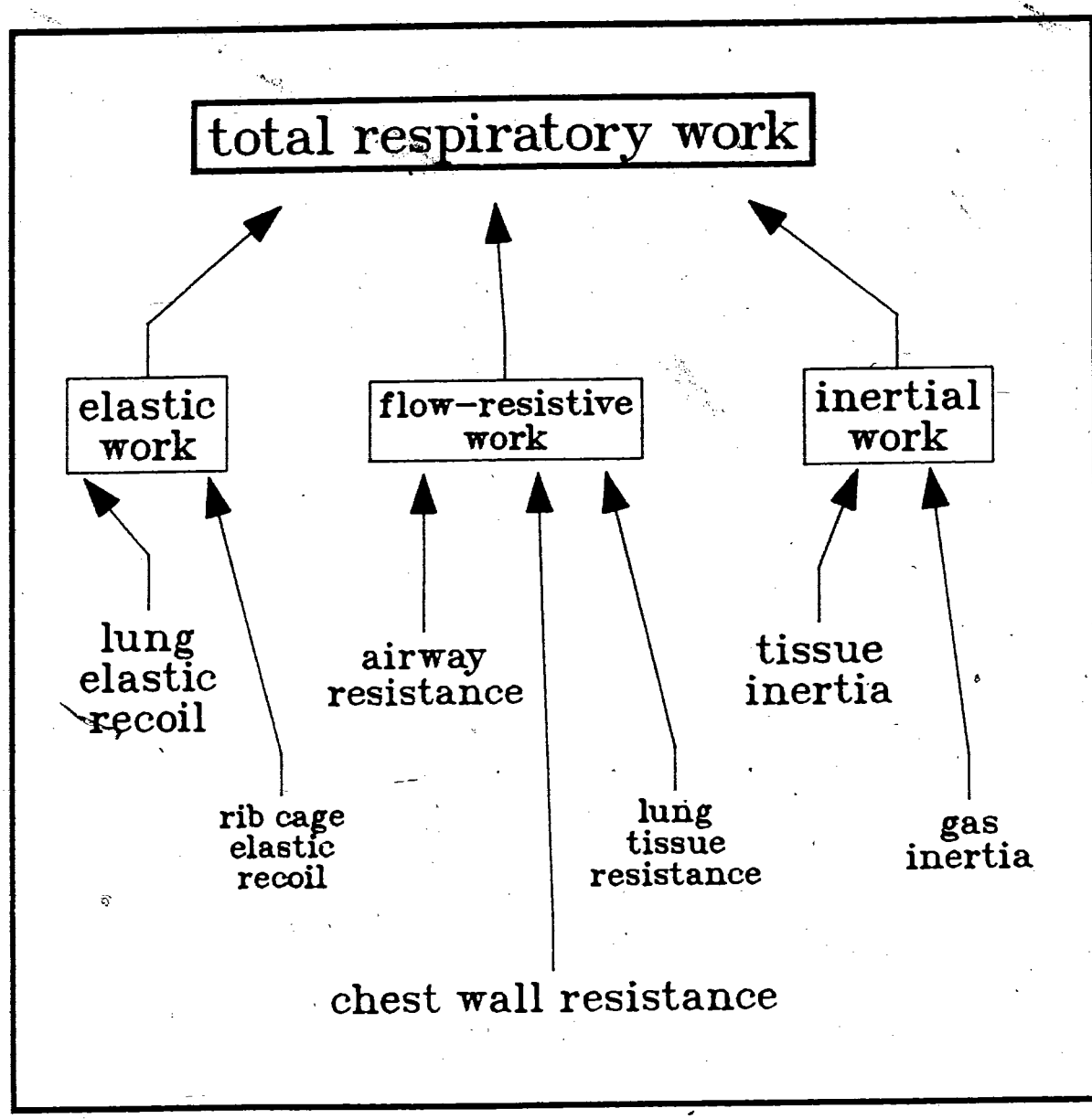


Figure 1.1: Respiratory work and the forces from which it arises.

Water density is about 770 times greater than that of air at sea level. Descent to a depth of only 10 metres will double ambient pressure. Under these circumstances an immersed, upright man will be exposed to an appreciable pressure gradient from head to toe. A man 1.7m tall experiences a total body pressure gradient of 16.67kPa ($170cmH_2O$)[2]. This is equivalent to the pressure difference between sea level and an altitude of 1500m. A second gradient exists between the mouth and diaphragm. Both pressure gradients have been shown to produce respiratory perturbations resulting in altered ventilatory work during head-out immersion (Agostoni *et al.* 1966, Hong *et al.* 1969, Flynn *et al.* 1975, Dahlbäck *et al.* 1979).

Man adopts a variety of postures during recreational and commercial diving. The commercial diver is forced, by the nature of his work, to spend most of his diving hours in an upright posture. In this position the craniocaudal pressure gradient is greatest. When receiving air at a pressure head equal to that of water at mouth level, there exists a hydrostatic pressure imbalance between the external thoracic pressure and mouth pressure, equivalent to about 2.45-2.94 kPa (25-35 cmH_2O), depending upon stature. This is physically similar to negative pressure breathing, which has been shown to reduce lung volumes and elevate airway resistance (R_{aw}) (Ting *et al.* 1960a, Agostoni *et al.* 1966, and Bjurstedt *et al.* 1980).

It has been consistently demonstrated during immersion, that the imposition of a vertical hydrostatic pressure gradient produces a translocation of peripheral blood to the thorax (Arborelius et el. 1972, Farhi and Linnarsson 1977, Löllgen *et al.* 1980, and Krasney *et al.* 1984). Elevated venous return increases cardiac output, possibly due to a preloading effect, to produce a concommitant pulmonary engorgement (Arborelius *et al.* 1972a, Begin *et al.*1976, Löllgen *et al.* 1980, and Hajduczok *et al.* 1984). The effect of this translocation, and mechanical compression of the thoracoabdominal region, is to reduce lung relaxation volumes and total respiratory compliance (Agostoni *et al.* 1966, Hong *et al.* 1969, McKenna *et al.* 1973, Flynn *et al.* 1975, and Craig and Dvorak 1975). These changes elevate elastic respiratory work.

Upright immersion also produces volume-dependent increments in airway and pulmonary resistance (Agostoni *et al.* 1966, Sterk 1970, 1973, Dahlbäck *et al.* 1979, Löllgen *et al.* 1980). Such changes increase flow-resistive ventilatory work.

---

[2] The *Système Internationale d'Unités* (*SI* units) are used throughout this thesis (Appendix One).

Commercial divers respire gas at mouth pressure whilst working upright, and thus will be required to generate this additional elastic and flow-resistive work. At the very least, such increased effort will reduce respiratory comfort and relative respiratory work efficiency ($J.l^{-1}$), and at worst may be frankly dangerous, if the loading is large enough to impose a ventilatory limit on gas exchange during emergency operations.

1.0.1 Statement of the problem.

The general lack of experimental evidence concerning levels of respiratory work divers can tolerate, has resulted in the ventilatory requirements of underwater breathing apparatus remaining largely unspecified. Implications of this omission, for safe and efficient pulmonary function in divers using conventional apparatus, remain unknown. By evaluating the components of respiratory work at various breathing pressures, one may derive optimal physiological specifications for the design of respiratory apparatus necessary to improve work efficiency, respiratory comfort and safety for commercial divers.

Lung *centroid* pressure is defined as the breathing gas pressure required to return static respiratory mechanics to normal. Lung *centroid* (or thoracic centre of pressure) may be considered a spatial location within the lungs, representing the mean hydrostatic pressure acting on the outside of the thorax. It is postulated that gas delivery at pressures approximating lung *centroid* pressure will return respiratory work towards levels which prevail in air.

1.1 HYPOTHESES

1.1.1

Upright, whole-body immersion without compensation of air supply pressure, produces a positive pressure shift of the total respiratory compliance curve, contributing to a reduced lung relaxation volume, and increased elastic work of breathing.

1.1.2

The provision of air supplied at lung *centroid* pressure, during upright immersion, will restore the position of the respiratory compliance curve, and return lung volume subdivisions and capacities towards levels which exist in air, thereby reducing

inspiratory muscle work required to overcome respiratory elasticity. It is further postulated that these work levels will be equivalent to the inspiratory muscular work in air.

1.1.3

Uncompensated upright immersion will elevate the pulmonary flow-resistive work of breathing. The provision of air supplied at lung *centroid* pressure, during upright immersion, will reduce pulmonary flow-resistive work of breathing during immersion, returning it towards values obtained in air.

1.1.4

Increased internal elastic and flow-resistive pulmonary work, accompanying uncompensated, upright immersion will cause respiratory discomfort during severe exercise and will limit physical power. It is hypothesised that provision of air at lung *centroid* pressure during exercise, will: (a) reduce respiratory effort, (b) improve respiratory comfort, and (c) permit divers to tolerate greater workloads.

1.2 ABBREVIATIONS AND DEFINITIONS OF TERMS

The definitions and symbols used within this thesis were abstracted from the recommendations of Pappenheimer (1950), the International Union of Physiological Scientists (Bartels *et al.* 1973), and the American College of Chest Physicians and the American Thoracic Society (1975)[3].

1.2.1 Lung volumes.

The lungs may be divided into four non-overlapping, primary compartments. Summation provides total lung capacity (Boren *et al.* 1966). When volume is expressed per unit time, it becomes a measure of gas flow ($\dot{V}$).

1. **Expiratory reserve volume** (ERV): The volume of air which may be expired from the point of end tidal expiration to maximal expiration ('reserve air', Hutchinson 1846).
2. **Inspiratory reserve volume** (IRV): The volume of air which may be inspired from the point of end tidal inspiration to maximal inspiration (Boren *et al.* 1966).
3. **Residual volume** (RV): The volume of air remaining in the lungs following the fullest possible expiration (Davy 1800).
4. **Tidal volume** (V_T): The volume of gas expired or inspired during the respiratory cycle ('breathing air', Hutchinson 1846).

1.2.2 Lung capacities.

Lung capacities are composed of combinations of two or more primary volumes. There are four standard lung capacities (Boren *et al.* 1966).

1. **Functional residual capacity** (FRC): The volume of air remaining within the lungs at the end of a normal expiration (Lundsgaard and Schierbeck 1923). FRC = ERV + RV.
2. **Inspiratory capacity** (IC): The volume of air inspired from the point of end tidal expiration to maximal inspiration ('complemental air', Hutchinson 1846). IC = V_T + IRV.
3. **Total lung capacity** (TLC): The total volume of gas within the lungs at the end of a maximal inspiration. It is composed of all four lung volumes (Boren *et al.* 1966).
4. **Vital capacity** (VC): The volume of air expelled from the lungs during expiration, from the point of maximal inspiration to the point of maximal expiration (Hutchinson 1846). VC = ERV + IC.

[3] Reference citations have, where possible, been traced to primary sources.

1.2.3 Ventilatory variables.

1. **Breathing frequency** (f_b): The number of breaths per unit time (usually b.min^{-1}) (Pappenheimer 1950).
2. **Forced expired volume** (FEV): The volume of gas expired in a given time during the forced vital capacity manoeuvre (*e.g.* $FEV_{1.0}$) (Boren *et al.* 1966).
3. **Forced vital capacity** (FVC): The volume of gas expired during performance of the vital capacity manoeuvre using maximal expiratory effort (Boren *et al.* 1966).
4. **Maximal voluntary ventilation** (MVV): The maximal volume of air (per unit time) that is voluntarily breathed without extrinsic control of tidal volume or frequency of breathing. The term maximal breathing capacity is reserved for corresponding values driven involuntarily (*e.g.* exercise, or carbon dioxide driven) (Boren *et al.* 1966).

1.2.4 Respiratory work.

1. **Work of breathing**(W_{res}): Classically physicists define work as a scalar product of the vectors force (F) and displacement (s). When a force applied to a body causes it to move in the direction of the force application, then work is performed on that body. Thus:

$$W = \int F \cdot ds$$

The product of Force (newtons) and displacement (metres) has the metric units of work: N.m (1J = 1 N.m). Since the 'bellows-action' of the thorax pumps air, force may be substituted by the pressure difference (P) required to produce gas flow and alveolar duct cross-sectional area (a):

$$F = P \cdot a$$

Displacement may be substituted by lung volume change (V) divided by the alveolar duct cross-sectional area (a):

$$s = V / a$$

Substitution of the last two expressions into the first (and simplifying) provides a means for deriving the flow-resistive work of the respiratory pump:

$$W_{res} = \int P \cdot dV \qquad [\text{units}=N.m^{-2} \cdot m^3 = N.m = \text{Joule}]$$

Classically lung pressure is taken as dynamic transpulmonary pressure (pressure at the airway opening [mouth] minus pleural pressure) and oesophageal pressure is used as an approximation of average pleural surface pressure, Otis 1964. Volume is obtained from respired gas measured at the mouth. This technique fails to account for flow-resistive work performed on the tissues of the thorax and abdomen, which may account for about 25% of respiratory work during exercise (Opie

et al. 1959, Otis 1964, Goldman *et al.* 1976), and the work of antagonist muscles opposing each other during breathing (Milic-Emili *et al.* 1981). W_{res}, derived using spontaneous pressure-volume relationships, is defined as work performed to overcome airway and lung tissue resistance. Under these circumstances W_{res} should more precisely be named **pulmonary work** (W_{pul}). This nomenclature has been adopted.

2. **Specific respiratory work and power**: Both work and power may be expressed in absolute units, or in terms relative to the volume of gas moved. This later expression permits normalisation of data to account for inter-subject, intra-subject and inter-trial variations in tidal volume.
3. **Power** ($\dot{W}_{res}$) : A time derivative of the scalar quantity work, power is work performed per unit time:

$$\dot{W}_{res} = dW_{res} / dt$$

1.2.5 Respiratory pressures.

All pressures are expressed relative to an anatomical or physical location by use of subscript abbreviations. Anatomical subscripts include: airway (aw), airway opening or mouth (ao), alveolar (alv), oesophagus (oes), gastric (ga), pleural (pl), lung (l), lung tissue (lt), diaphragm (di), rib cage (rc), body surface (bs), total respiratory system (rs), chest wall (w) (which is a composite of the rib cage and abdomen (ab)) (Agostoni and Mead 1964), transthoracic (tth), and transpulmonary (tp) (Goldman *et al.* 1976). These subscripts have also been used to describe the components of the work of breathing (*e.g.* W_{aw}) and flow-resistance (*e.g.* R_{lt}). Physical subscripts include: ambient pressure (A).

Pressures are measured statically (zero gas flow), quasi-statically or dynamically and have the respective subscripts 'st', 'q-s' and 'dyn'. Location subscripts follow measurement subscripts where both are present, and in such cases the former appears in parentheses (e.g. $P_{st(l)}$). The following expressions define and display pressure inter-relationships.

$$P_{st(rs)} = P_l + P_w = P_{ao} - P_{bs} = P_{trs}$$

$$P_{st(w)} = P_{pl} - P_{bs}$$

$$P_{st(l)} = P_{ao} - P_{pl} = P_{tp}$$ **Pulmonary pressure**

$$P_{st(w)} = P_{st(rc)} = P_{tab} + P_{tdi}$$

(*If the rib cage and abdomen- diaphragm operate in parallel while the diaphragm and abdomen operate in series*)

$$P_{tp} = P_{alv} - P_{pl}$$

$$P_{tth} = P_{pl} - P_{bs}$$

1.2.6 Compliance.

Lung compliance (C) measures lung tissue capacity to be distended during inflation (elastance = 1/C) (Radford 1964). Compliance may be measured in static (zero flow and zero volume acceleration), quasi-static (slow flow or transient occlusion) and dynamic (flow and/or volume acceleration not equal to zero) conditions (C_{st}, C_{q-s} and C_{dyn}) and may be used to refer to the lungs (*e.g.* $C_{st(l)}$), chest wall, or the entire respiratory system (including lungs and chest wall *e.g.* $C_{st(rs)}$). Compliance may be derived at any point along the pressure–volume curve, and is given by the curve slope:

$$C = dV / dP$$

Dynamic compliance is the ratio of tidal volume to the pressure change between the two points of zero gas flow at either end of the tidal excursion (Mead and Milic-Emili 1964). [4]

1.2.7 Resistance.

1. **Airway resistance** (R_{aw}): The *opposition* to air movement created by friction, and resulting in the loss of mechanical energy as heat. To compensate for resistance, respiratory muscles perform more work to move a given gas volume [5].
2. **Pulmonary resistance** (R_{pul}): Frictional resistance is also imposed by lung tissue moving across itself (R_{lt}). Pulmonary resistance is the summation of airway and lung tissue resistances.

$$R_{pul} = R_{aw} + R_{lt}$$

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

$$R_{rs} = R_{pul} + R_{wt}$$

[4] Compliance measurements are sensitive to two variables: (a) lung volume excursion immediately prior to measurement, and (b) the time taken to produce the volume change (i.e. static versus dynamic measurement).

[5] Airway resistance will vary with lung volume and flow rate.

1.2.8 Miscellaneous terms.

1. **Eupnoeic pressure**: Derived by Paton and Sand (1947) to refer to the air delivery pressure subjectively determined as most comfortable during immersion.
2. **Lung centroid** (P_{LC}): Derived from the phrase *'centre of pressure of the thorax'*, and used initially by Paton and Sand (1947). Lung *centroid* pressure (P_{LC}) is defined as the pressure required to return the immersed lung relaxation volume back towards that level which exists in air. The *centroid* is a spatial location within the thorax which represents the mean hydrostatic pressure acting on the outside of the thorax during such an 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 base of the lungs.
3. **Oxygen consumption** ($\dot{V}O_2$) : The quantity of oxygen consumed by an organism or tissue. Since oxygen forms vital links within the energy metabolism processes, $\dot{V}O_2$ has been used to express energy production, using the thermal equivalent for the non-protein respiratory quotient. $\dot{V}O_2$ is expressed in absolute ($L.min^{-1}$) and relative units ($ml.kg^{-1}.min$).
4. **Relaxation pressure** (P_R): Intrathoracic pressure obtained during complete respiratory relaxation against an occluded airway, with the glottis open.
5. **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 equilibrated to atmospheric pressure, but to the delivery pressure of the apparatus. When the airway is open to the atmosphere, when seated in air, the relaxation volume is usually equivalent to the functional residual capacity.

CHAPTER 2

DETERMINATION OF THE THORACIC CENTRE OF PRESSURE

Historians accredit Becker developing the first helmet-hose diving apparatus in 1715 (Davis 1962). Augustus Siebe initiated 'hard-hat' diving in 1837 with the 'closed' suit and vented helmet (Davis 1962). According to Haldane (1907), the Siebe apparatus was still used by the Royal Navy in 1907. A vent valve located on the helmet, permitted divers to regulate the volume (and hence pressure) of air inside the suit. With the valve fully open, pressure acting on a given body part equalled atmospheric pressure, plus that of a water column equal to the difference in depth between the body part and valve. Regions below the helmet were at greater pressure, with the exception of alveolar air which, due to its communication with helmet air, was at helmet pressure when the glottis was open and airflow absent (Pascal's principle).

Using this apparatus, Haldane (1907) reported the first experiments on the respiratory consequences attending disparities between air delivery pressure and external pressures. Attachment of a 60cm tube at the vent valve permitted divers to regulate helmet pressure by positioning the tube end at various depths. With the valve 7-10cm above the helmet ventilation became almost impossible, due to an inspiratory limitation. Increasing valve depth progressively reduced respiratory effort. Respiration was easiest with the valve at shoulder level (deeper depths resulted in excess helmet lift and loss of neutral buoyancy).

Head-out, or whole-body, upright immersion results in an elevation of external thoracic pressure relative to alveolar pressure. Thus alveolar pressure (P_{alv}) becomes negative relative to the hydrostatic forces acting on the thorax, abdomen and legs. Furthermore, the resultant transrespiratory pressure (P_{trs}) is not equally distributed over the thorax, as in air (Figure 2.1). Resting thoracic volumes, which determine respiratory mechanics, are regulated by lung and chest wall elasticity, transrespiratory pressure gradients, gravity, intrathoracic blood volume and pressure differences between parietal and visceral pleural circulation (Rahn *et al.* 1946, Fenn 1951, Agostoni and Mead 1964, Mead and Agostoni 1964, Otis 1964, Agostoni 1972, Miserocchi *et al.* 1981, Denison 1983). Upright immersion has been shown to adversely alter respiratory mechanics (Paton and Sand 1947, Hong *et al.* 1960, Jarrett 1965, Agostoni *et al.* 1966, Hong *et al.* 1969,Craig and Dvorak 1975, Flynn *et al.* 1975, Dahlbäck 1975, 1978, Prefaut *et al.* 1979, Minh *et al.* 1979, Löllgen *et al.* 1980). Such mechanical perturbations explain the breathing difficulty experienced by the divers in Haldane's (1907) study.

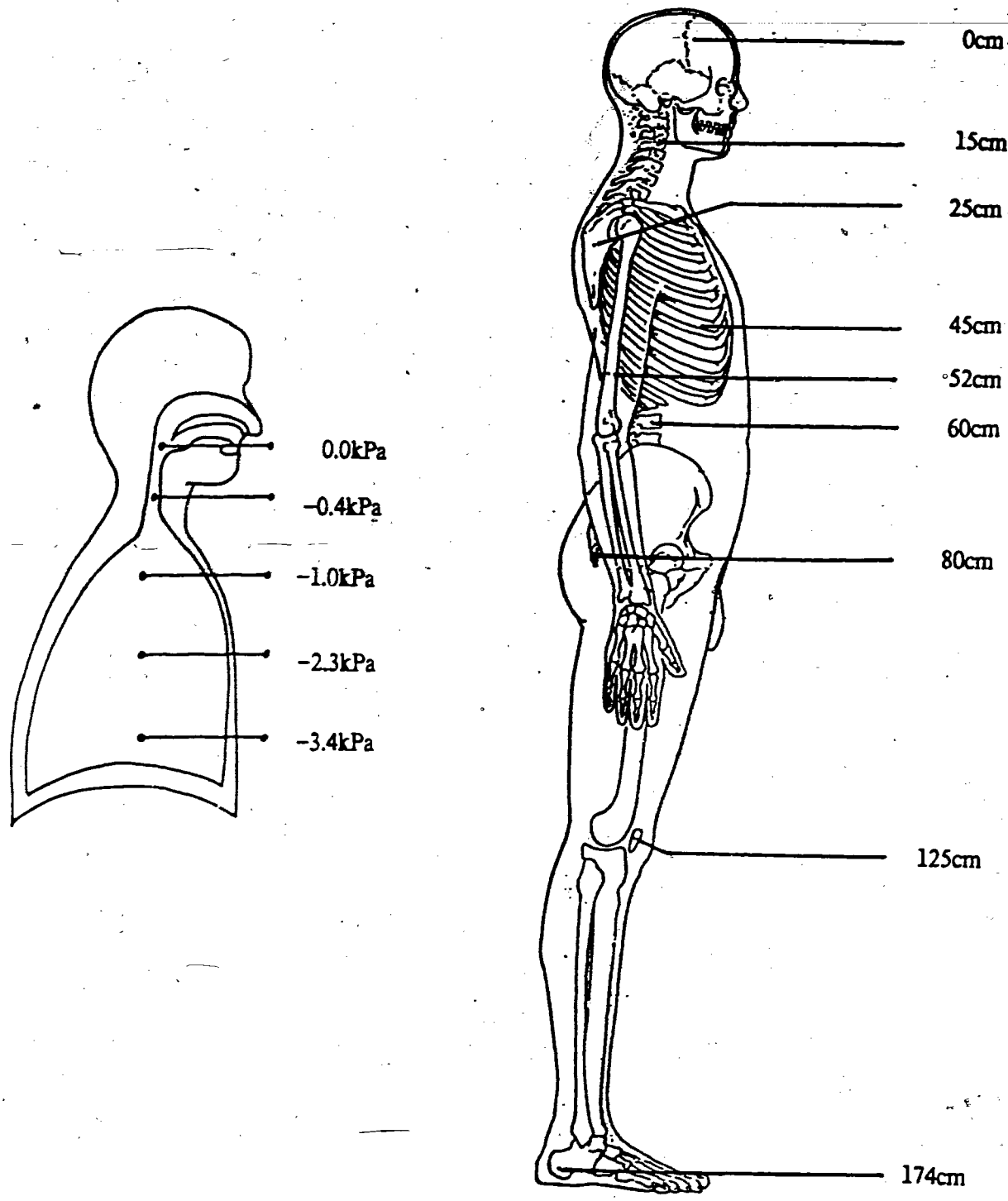


Figure 2.1: **Hypothetical hydrostatic and transrespiratory pressures during whole-body, upright immersion.** If a man 1.74m tall is immersed upright, he will experience a vertical hydrostatic pressure gradient from head to toe, of approximately 17kPa. Over the thorax there will exist a negative transrespiratory pressure, which becomes more negative with increased depth.

Paton and Sand (1947) introduced the terms *eupnoeic* and *centroid* pressure to describe the ideal gas delivery pressure. They defined *eupnoeic* pressure as the air supply pressure producing greatest respiratory comfort, and equated it with the pressure evenly diffused in the lungs, which matched the mean of the external hydrostatic pressure gradient. The difference between air supply pressure and *eupnoeic* and/or lung *centroid* pressure[1] reflects the hydrostatic imbalance between P_{alv} and mean thoracic surface pressure (Figure 2.1).

Jarrett (1965) extended the concept of thoracic centre of pressure, or lung *centroid* pressure, by recognising that it is moved by the phase of respiration and by the posture adopted. Because lungs lack radial symmetry, and because they behave as flaccid balloons within a semi-rigid container, the thorax cannot possess centrosymmetry (a fixed *centroid locus*), and thus, the *centroid* is applicable only to the posture and respiratory phase from which it was derived.

During upright immersion, the negative transrespiratory pressure encountered means that divers are faced with a breathing pressure imbalance. The pressure acting on the thorax is greater than P_{alv}. If the magnitude of this imbalance is known, underwater breathing apparatus may be modified to enable provision of air at slightly higher pressures. In the Siebe suit, this would mean closing the vent valve to increase helmet pressure, since divers breathe from the helmet and suit volume without intervening connections. Self-contained underwater breathing apparatus (SCUBA) provide air at the ambient pressure of the second stage regulator, which is usually positioned at mouth depth. Hydrostatic imbalance between *centroid* pressure and the pressure at mouth level, incurred by this apparatus, may be minimised by supply of air at positive pressure, relative to mouth pressure.

In upright immersion, Paton and Sand (1947) reported subject preference for delivery pressures more positive than *eupnoeic* pressure rather than more negative. Subsequent investigators reported restoration of normal lung mechanics during immersion by use of positive mouth pressures (Beckman *et al.* 1961, Flynn *et al.* 1975, Thalmann *et al.* 1979). In accordance with these observations, and the works of Paton and Sand (1947), and Jarrett (1965), lung *centroid* pressure (P_{LC}[2]) is defined, in this thesis, as the pressure required to return the immersed thoracic relaxation volume to levels which exist in air. When immersed upright, delivery of air at P_{LC} represents a positive pressure at the mouth, with a

---

[1] *Centroid* is a term used in physics and mathematics to describe the centre of an area, volume, inertia or mass.

[2] Chapter One

progressive decline of pressure gradient moving upstream to the alveoli. Apical alveoli will be under very slight positive transrespiratory pressure (P_{trs}), if the tracheal bifurcation is located above P_{LC} depth, while basal alveoli will experience a slightly negative P_{trs}.

Three methods have previously been employed to derive P_{LC} : (1) subjective pressure selection by immersed divers (*i.e. eupnoeic* pressure), (2) immersion with airway occluded at the relaxation volume obtained in air, to measure the change in P_{alv} (non-steady state measurement), and (3) analysis of static respiratory pressure-volume relaxation curves for air and immersed states, to measure pressure change at the respiratory relaxation volume (V_R).

Using the psychophysical method of limits, Paton and Sand (1947) asked subjects to determine *eupnoeic* pressures at various depths and orientations. [3]. One would expect *eupnoeic* and P_{LC} to be equivalent, since greatest comfort should coincide with greatest respiratory mechanical efficiency. However, the authors reported mean vertical, supine and prone *eupnoeic* pressures of 8.73±1.1cm below the auditory meatus (above the thorax), about 10cm and 10±3.5cm dorsal to the sternal notch[4] respectively.

Paton and Sand (1947) suggested oral sensation dictated pressure selection to some degree, that immersion elevated the lung's centre of volume, and that thoracic compression may elevate chest recoil and pleural pressure (P_{pl}). If the resting thoracic volume is determined by the algebraic sum of chest wall and lung recoils, then the relaxation pressure (P_R) of the respiratory system may be expressed as the difference between internal and external expansive and compressive forces. Paton and Sand (1947) suggested that an increase in chest recoil and P_{pl} (*i.e.* P_{pl} becomes less negative) with immersion, will reduce the upright lung *centroid* pressure. This explanation was offered to justify their positioning of the *centroid* above the thoracic cage.

Lung *centroid* position in the horizontal plane was subsequently verified (Hong *et al.* 1960, Jarrett 1965, Craig and Dvorak 1975), however, it appears implausible that the upright lung *centroid* would be located above the thorax. Jarrett (1965) proposed that this may have been created by additional muscular effort required to hold the mouthpiece in place at high lung volumes and delivery pressures. Thompson

---

[3] *Eupnoeic pressure* was measured in the units: cmH_2O ($1cmH_2O$ = 0.09803kPa). Such measurements equal the hydrostatic pressure at a given water depth (in centimetres), thus it is convenient to express both *eupnoeic* and P_{LC} as a water depth, relative to the ambient pressure at an anatomical reference point, which has a well defined relation to the chest.

[4] Also known as: supra-sternal notch, sternal angle, jugular notch or interclavicular notch.

and McCally (1967) suggested that large transpharyngeal pressures would produce local discomfort before P_{LC} was reached, and subjects would consequently select pressures below P_{LC}.

Thompson and McCally (1967) subsequently used facial and pharangeal counter-pressure in a replication of the Paton and Sand (1947) study, to further investigate the problem. They found *eupnoeic* pressure was ~8cm H_2O (0.78 kPa) greater with, than without pharangeal counter-pressure. With facial counter-pressure, mean *eupnoeic* pressure was still only 4.0±8.6cm below the sternal notch, though subject variability was high (range = −5 to +20cm). It is concluded by the current author that *eupnoeic* techniques are inadequate to provide an accurate appreciation of the upright P_{LC}, possibly because the poor kinesthetic sensation of the lung tissue makes it difficult to detect changes in P_{alv}.

Hong *et al.* (1960, 1969) implemented non-steady state determinations of P_{LC}. It was postulated that P_{alv}, during airway occlusion at a known thoracic relaxation volume, would be elevated as the body was lowered into the water. The elevation magnitude would be proportional to the mean external thoracic pressure gradient (*i.e.* P_{LC}). They reported an average P_{alv} rise of 0.84kPa (8.6cm H_2O), relative to atmospheric pressure (P_A). This was half the pressure change observed following several minutes at the same immersion depth. The discrepency was attributed to involuntary maintenance of inspiratory muscle tone by the subjects. It is also possible that transient glottal closure prevented the true P_{alv} from being recorded.

Minh *et al.* (1977) determined FRC (N_2 washout technique) in upright anaesthetised dogs in air. The airway was occluded and the dogs were immediately immersed upright to the midneck level while P_{alv} was monitored (relative to P_A). P_{LC} was 3.18 ± 0.12 kPa (32.4±1.2 cmH_2O)[5]. These data, showing a much greater increase in P_{alv} during immersion when inspiratory muscles are anaesthetised, support the inadequacy of the non-steady state technique for derivation of P_{LC} in non-anaesthetised subjects. The relevance of this P_{LC} to human studies is unknown. It must be noted that *centroid* depth is not able to be expressed relative to the sternal notch. Because P_{LC} was determined relative to P_A , *centroid* depth can only be expressed relative to the water surface (*i.e.* at some mid-neck point). Lung *centroid* in these dogs must have been located about 32cm below mid-neck level.

Jarrett (1965) attempted P_{LC} measurement from respiratory pressure-volume relaxation curves, obtained in air and underwater. Subjects (N = 3) performed static inspiratory manoeuvres during

[5] The investigations of Hong *et al.* (1960, 1969), Craig and Dvorak (1975) and Minh *et al.* (1977) did not specifically address the issue of P_{LC} measurement. However, they did obtain data from which P_{LC} could be obtained.

head–out, upright and supine postures. Lung relaxation volumes were derived from upright control curves obtained in air. P_{LC} was obtained by computing regression equations for alveolar pressure as a function of lung volume for head–out immersed subjects, and calculating the air pressure required to reproduce the upright control relaxation volumes (V_R). Thus, Jarrett (1965) obtained a pressure shift of the control V_R along the pressure axis[6] (Figure 2.2).

Classical description of the respiratory compliance curves was first provided by Rahn *et al.* (1946). The total respiratory curves represent a combination of lung tissue and chest wall pressure–volume characteristics. Static pressure exerted by the total respiratory system equals the sum of lung and chest wall static recoil pressures (*i.e.* $P_{st(rs)} = P_{st(l)} + P_{st(w)}$[7]). The mechanical zero of the system (which corresponds with the respiratory V_R) falls between the relaxation volumes of the lung and chest. The total respiratory compliance curve is sigmoidal, with the lower curvature being attributed to the serial collapse of smaller airways, alveolar closure (Velasquez and Farhi 1964, Labadie and van Eenige 1969), and chest wall stiffness. The upper asymptote is due to attainment of near maximal lung expansion. Conventionally such curves are reported with pressure on the abcissa, however, pressure is the dependent variable. The slope of the conventional curve is termed compliance, and describes the ability of the lung system to distend with internal pressure increments[8]. Elastance is the inverse of compliance and describes resistance to distention.

From analysis of total respiratory compliance curves, Jarrett (1965) found P_{LC} to be 19cm inferior and 7cm dorsal to the sternal notch. These results have been used for a variety of applications by numerous investigators (Thompson and McCally 1966, 1967, Sterk 1970, 1973, Craig and Dvorak 1975, Minh *et al.* 1977, 1979, Sawka *et al.* 1978, Milne and Morrison 1979, Weltman and Katch 1981, Ostrove and Vaccaro 1982, Grismer and Goodwin 1983), yet these data were obtained from only three subjects. No subsequent studies have been found by the author which deal specifically with P_{LC} measurement.

Several investigators (Table 2.1) have produced data permitting the calculation of P_{LC}. Inconsistencies between protocols render some of the data contradictory to an upright P_{LC} of 19cm below the sternal notch, as reported by Jarrett (1965). Yet the authors have suggested their evidence is

---

[6] This technique uses a constant lung volume (or isovolume) with which to compare control and immersion compliance curves. Pressure displacements are determined from pressure changes recorded at this isovolume.

[7] Chapter One.

[8] Chapter One.

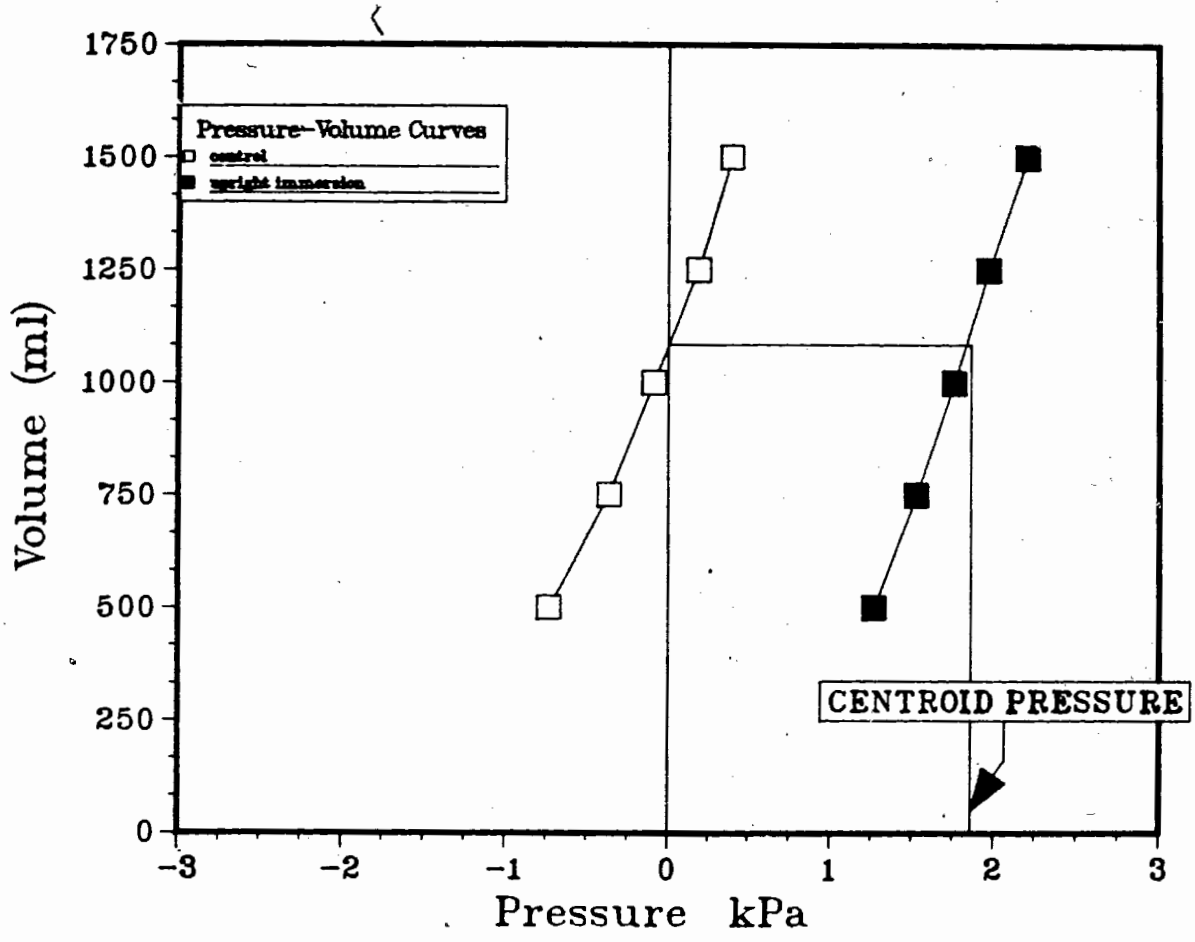


Figure 2.2: **Displacement of respiratory relaxation volume along pressure axis during head-out, upright immersion.** [Redrawn from Jarrett 1965. Curves represent mean values for three subjects.].

supportive of Jarrett's observations. The major inconsistency is the use of different anatomical references. Lung *centroid* may only be referenced to an anatomical structure if transpulmonary pressures are expressed relative to the ambient pressure at that point. Most studies subsequent to Jarrett (1965) report isovolume, compliance curve shifts relative to P_A. Since various immersion depths have been investigated, lung *centroid* pressure may be referenced only to the water surface, and hence the anatomical site at that level. This point has been overlooked by most authors. In order to allow cross study comparisons, data must be referenced to a common anatomical point. Data from previous investigations has been adjusted to permit lung *centroid* pressure to be referenced to the sternal notch (Table 2.1).

Re-analysis of the adjusted data from Table 2.1 permits computation of a weighted average lung *centroid* (according to subject numbers) relative to the sternal notch. *Centroids* averaged 12.7cm and 7.8cm for upright and horizontal positions respectively (*eupnoeic* data not included). Horizontal positioning agrees well with the original work of Paton and Sand (1947) and Jarrett (1965), however, it appears 19cm may be too deep for the upright *centroid* .

2.0.9 Purpose of the chapter.

The objective of this study was to re-evaluate lung *centroid* position in the erect (seated) and prone postures using the isovolume, compliance curve technique used by Jarrett (1965). The latter posture was chosen since it represents swimming posture, while the former (within constraints of the diving chamber) best represents the posture of a working commercial diver. A sample size of 20 was chosen to ensure that measurements derived could be utilised with a degree of confidence, to evaluate possible benefits of gas delivery at pressures approximating P_{LC}.

Table 2.1: **Lung centroid derivation using data from previous investigations.**

REFERENCE	Method	Subject number	Immersion depth	Water temp.	Reported P_{LC}	Corrected P_{LC}
Paton and Sand 1947	eupnoeic	8	total (U)	36.1–37.8° C	8.73 (E)	---
		5	total (Su)		14.2 (F)	--
		4	total (P)		10 (S)	—
Hong *et al.* 1960	P–V (T)	3	total (Su)	28° C	6 (S)	8.1
Beckman *et al.* 1961	PPV	6	neck (U)	34.4° C	8–14 (W)	2–8
Jarrett 1965	P–V (T)	3	SN (U)	32–34° C	19 (SN)	19
		3	S (Su)		7 (S)	7
Agostoni *et al.* 1966	P–V (C)	2	neck (U)	?	21 (W)	14.5
Thompson & McCally 1967	eupnoeic	7	ear (U)	33–34° C	4 (SN)	4
Hong *et al.* 1969	P–V (T)	4	shoulders (U)	25–30° C	16 (W)	11
McKenna *et al.* 1973	P–V (T)	1	?	?	20 (W)	<20?
Craig & Dvorak 1975	P–V (T)	6	neck (U)	33–34° C	25 (W)	17.8
		6	total (Su)		6 (W)	8
Flynn *et al.* 1975	P–V (T)	1	neck (U)		19 (W)	12

Abbreviations: C = chest wall, E = ear, F = tank floor, PPV = positive pressure ventilation to obtain air lung volumes, P–V = compliance curve analysis, P = prone, S = sternum, SN = sternal notch, Su = supine, T = total system, U = upright, W = water level. **Reported P_{LC}** = cm. below or above (–) anatomical reference (in parenthesis). **Corrected P_{LC}** = cm. above or below (–) sternal notch (upright) or plane (horizontal), and recalculated as per Jarrett (1965).

2.1 METHODS.

2.1.1 Subjects.

Twenty male non-smokers (including 13 divers), screened by questionnaire for normal lung function history, participated in this investigation. All received subject information packages and signed informed consent releases.

Subjects were required to perform a series of static pressure-volume relaxation manoeuvres over the volume range from residual volume (RV) to total lung capacity (TLC). Commencing with a relaxation at RV, subjects slowly inspired a variable volume of air, as determined by the experimenter, then relaxed for four to six seconds with the glottis open, against an occluded airway. Trials were performed upright and horizontal in air (control), and while totally immersed. Subjects unable to provide reproducible static pressure-volume curves in air were excluded and replaced. Twenty-nine subjects were tested before twenty suitable subjects were identified.

2.1.2 Apparatus

Lung volumes were measured using a pneumotachograph (Fleisch #4) coupled with a differential pressure transducer (Validyne DP103 ± 0.25kPa). Alveolar pressure was measured at the mouth, with the airway occluded and glottis open, using a differential pressure transducer (S.E. 1150 ± 6.2kPa).

Output from both transducers was amplified (S.E. 423/1E amplifier demodulator), flow signals were low pass filtered at 5Hz (Rockland model 432 dual high/low filter), and both signals passed to an IBM(PC), via an analog/digital converter (Tecmar Labpac), for storage. Data were sampled at 50Hz. McCall *et al.* (1957) demonstrated the peak frequency content of tidal volume and vital capacity manoeuvres to be 3.5 and 4Hz respectively. According to *sampling theorem* (Berson 1970), 50Hz sampling results in minimal signal distortion and information loss.

System linearity was evaluated [9] using a U-tube water manometer in parallel with the mouth pressure transducer. A step series of known pressures (±0.02kPa) was applied to the system over the

[9] Since it was crucial to know the system response (i.e. from transducer to computer), and since system components act in series (causing system amplitude response to be the product of component responses - Fry 1960), it was decided to evaluate the complete system, rather than linearity of separate components.

range – 5.9kPa to +4.9kPa. This range was deemed to include mouth pressures anticipated during static pressure–volume relaxation manoeuvres. The system was linear ($r > 0.999$), and remained so throughout the investigation.

Accuracy of volume and pressure measurement was confirmed by repeated application of known volumes and pressures to each system. The standard error for dry volumes was 0.007 litres (using 0.961 litre syringe injector as a standard), and for immersed volumes it was 0.016 litres (using 5.921 litre syringe standard). Five pressure standards were used (−1.96, −0.98, 0, +0.98, +1.96kPa (−20, −10, 0, +10, +20cmH_2O)). Pressures standards were applied with an accuracy of ±0.02kPa (±0.2cmH_2O). Standard errors for the five pressure standards were 0.006, 0.008, 0.005, 0.005, 0.009kPa respectively.

Volumes were calculated by integration of flow signals with respect to time. System validation included all components between the pneumotachograph and integration routines. Integrated volumes displayed a non–systematic variance with flow rate ($\bar{x}$ = 1.004±0.042 litres), when a one litre standard was passed through the pneumotachograph at flow rates between 20–40 l.min^{-1}.

Immersion trials were performed in the wet chamber of a hypo–hyperbaric chamber complex able to simulate depths to 300m with temperature and humidity control. Water temperature was regulated to 34.6±0.4° C by a thermostatically controlled heat exchanger. This necessitated the use of long pressure probes to connect the pneumotachograph to its differential pressure transducer. To evaluate the effects of probe length upon volume measurement, trials were performed using known volume standards (0.961 litres) and different probe lengths (0.415, 1.600 and 3.350 metres[10]). Increasing probe length resulted in a systematic but non–significant increment in the integrated volume measurement (0.966±0.034, 0.971±0.004, and 0.978±0.007 litres respectively), representing 0.52%, 1.04% and 1.77% above syringe volume, which itself was determined by repeated water filling.

During the immersion trials the pneumotachograph was kept above water. Preliminary trials revealed leakage around the mouth, causing occlusion of the Fleisch tubes when the latter were below mouth level. Since both subjects and the air supply were immersed (the latter to the depth of the sternal notch), lengths of low resistance tubing were used to link same to the pneumotachograph (Figure 2.3). At either end of this tubing were Mares (MR 12 III) regulators connected to compressed air storage tanks. Subjects were positioned in this circuit, close to the second regulator. Two, two–way taps

[10] Probe lengths corresponded with: the shortest possible probe length, the probe length for dry trials, and the probe length for immersion trials, respectively.

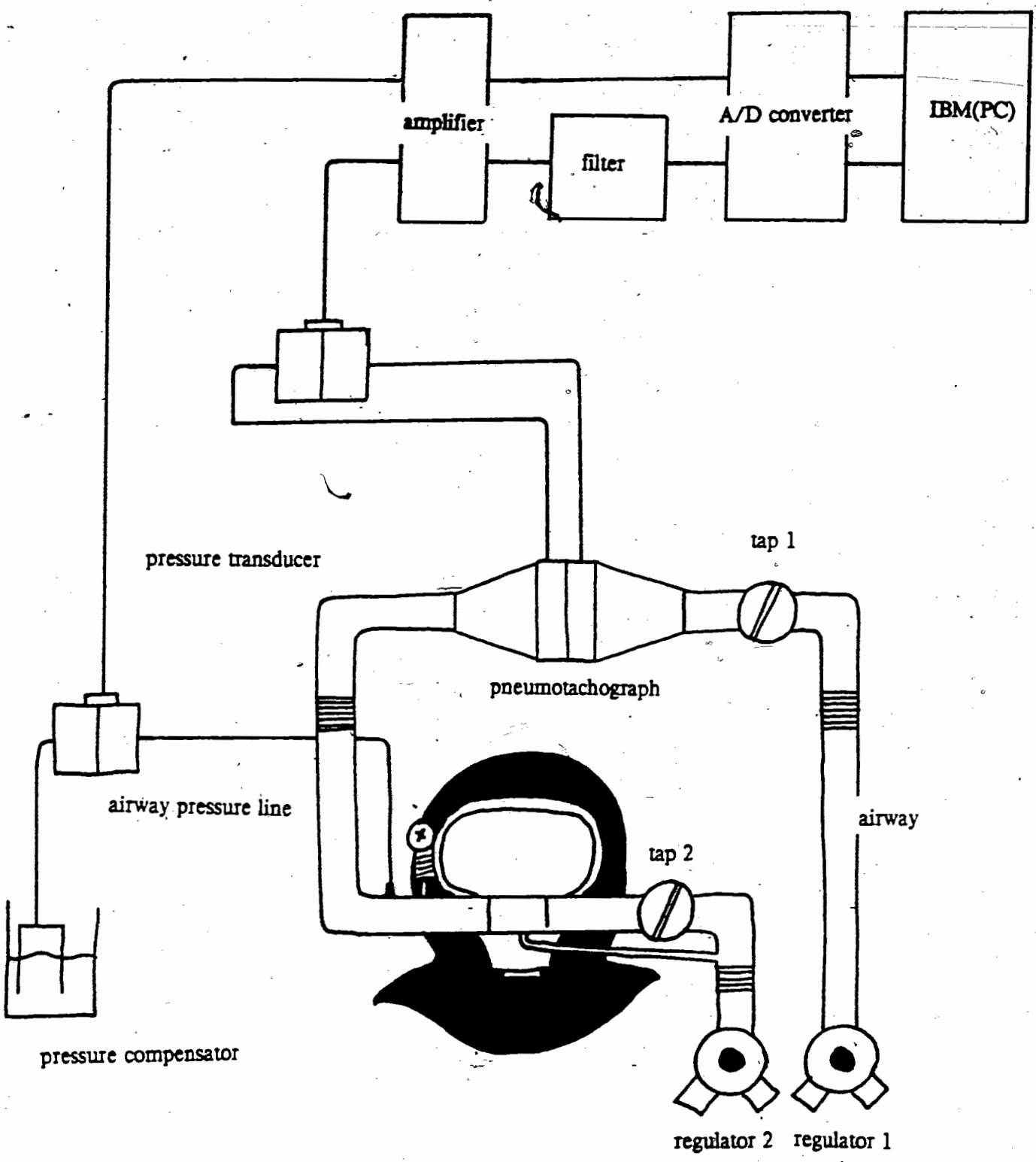


Figure 2.3: Schematic of apparatus for static pressure–volume curve measurement during immersion.

permitted the experimenter to control airflow through the pneumotachograph, and from which regulator the subject breathed. Subjects were instructed on how to operate the taps to obtain air on demand, if required.

Regulator 1 was positioned at sternal notch depth (upright trials), or in the sternal plane (prone trials), and provided air at that hydrostatic pressure. This prevented excessive negative pressure breathing during immersion. This regulator was used only during pressure–volume manoeuvres, because of the large dead space between it and the subject (approximately 2 litres), and to avoid condensation in the unheated pneumotachograph, created by rebreathing. Tap 1 remained closed except during trials.

The second regulator served three purposes:

(1) it provided an easily accessible emergency air supply to subjects,

(2) it permitted subjects to breathe comfortably below the water between trials, increasing their ability to reproduce relaxation pressures, and

(3) it obviated rebreathing through the pneumotachograph.

For control experiments in air, a small length of low resistance tubing (volume = 0.365 litres) connected subjects directly to the pneumotachograph. All other tubing was disconnected, but tap 1 remained. In each set–up there existed a volume of air between the subject and tap 1 (1.591 litre and 0.950 litres – immersion and control respectively). Lung volumes were corrected for gas expansion during negative pressure holds, and for compression during positive pressure holds.

Nine subjects took part in preliminary immersion trials using this apparatus. Subjects wore swimming goggles, a tightly fitting neoprene diving hood (to compress the cheeks) and a nose clip. Hands were used to support cheeks in both control and immersion trials. Data revealed marked divergence from the results obtained by Jarrett (1965), yet pressure–volume curves were highly reproducible. Jarrett suggested subjects were unable to adequately relax unless an oronasal mask was used. To evaluate the significance of this observation, all subjects were retested using a pressure compensated diving hood.

From this point, all 20 subjects performed immersion trials using a Kirby–Morgan band mask (U.S. Divers) (Figure 2.4). A small pressure line connected regulator 2 directly with the air space inside the hood, and pressurised the latter to the pressure at the depth of the regulator. Both regulators were maintained in the same horizontal plane at the level of the sternal notch (or sternum), pressurising cheeks

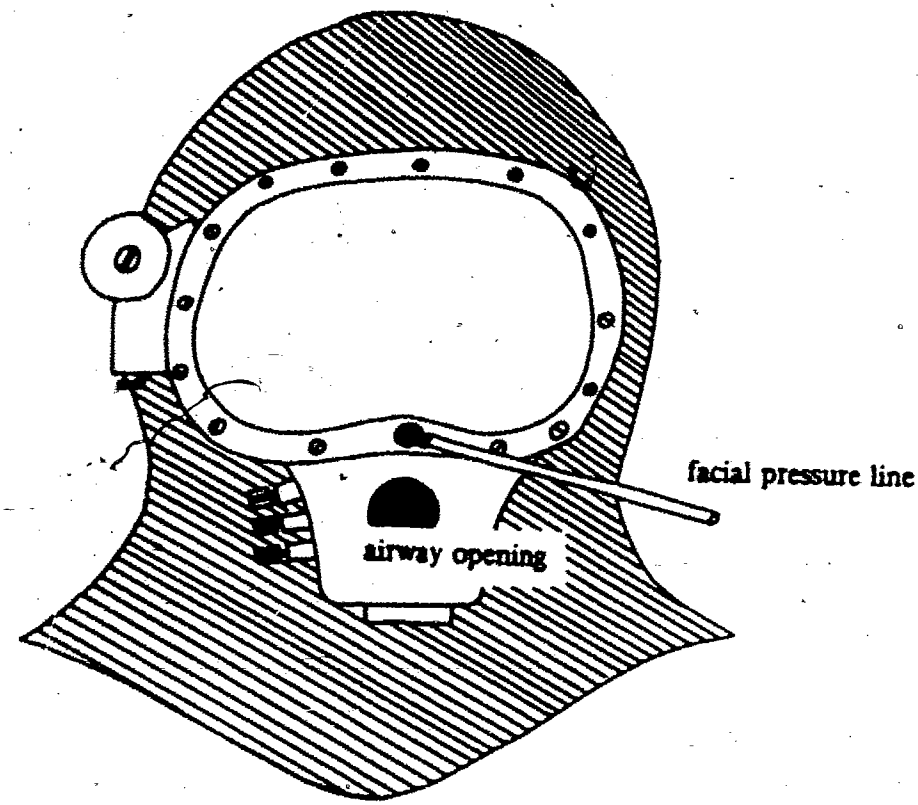


Figure 2.4: Kirby–Morgan band mask used to provide facial counter–pressure during immersion trials.

to reduce bulging at high relaxation volumes. Since the pressure line by-passed tap 2, mask pressure was constantly maintained. The hood provided strong rubber strappings to maintain its position during immersion. These facilitated mouthpiece stability, as it was impossible to physically eject the mouthpiece when strapped in position. Subjects wore a noseclip inside the hood to prevent air losses.

During all trials a U-tube water manometer was connected in parallel with the mouth pressure transducer, enabling the experimenter to determine when a stable relaxation pressure was attained[11].

Preliminary immersion trials produced relaxation pressures beyond the differential pressure transducer capacity. To bring pressures back into range a pressure compensator (Figure 2.3), which applied a pressure of about 0.15kPa, was connected to the transducer reference side, which had been open to air during control trials. The compensator consisted of a perspex cylinder, and pressure probe, immersed to known water depths, and connected in parallel with both the manometer and the transducer.

To relate immersion-induced effects found in this study with those reported in the literature, and to a defined anatomical point, relaxation pressures were expressed relative to the hydrostatic pressure at the superior border of the manubrium (sternal notch). Prone relaxation pressures were expressed relative to pressure at the sternal plane (a horizontal line through the anterior sternal border).

2.1.3 Calibration

Calibration was performed daily, or more frequently if the apparatus was shut down between subjects, or when large time gaps occurred between successive testing periods. Pressure calibration was performed at zero and +4.9kPa (+50cmH_2O) using a water manometer (±0.02Pa (±0.2cmH_2O)), utilising previously confirmed system linearity.

Flow calibration was carried out with apparatus *in situ* preceding trials. For control trials a series of six syringe volumes of 0.961 litres were passed through the pneumotachograph at various flow rates. Integrated flow was averaged and matched against volume standards. Immersion trial calibration could not be performed in this manner. With regulator 1 submerged, it was found that calibration by pumping syringe volumes at regulator pressure produced erroneous volumes. Finucane *et al.* (1972) demonstrated that geometry change of tubing upstream from the pneumotachograph had a marked affect upon the pressure differential across the resistive element. Thus flow calibration was performed with apparatus

[11] Relaxation was defined as the absence of inspiratory muscle tone. Since diaphragmatic and/or intercostal electromyograms were not taken, relaxation became a subjectively derived state. The experimenter could determine relaxation stability from manometer movement.

fully assembled, by employing a single inspiritory calibration stroke with a large syringe (5.921 litres)[12]. Calibration was performed with the regulator 10 - 15cm below the surface. Testing revealed no effect of regulator depth on calibration data, and calibration data obtained with the apparatus immersed was now equivalent to control calibrations.

2.1.4 Procedures

Subjects performed static respiratory pressure–volume manoeuvres between RV and TLC. Volume increments were inspiratory, and random in size. At each volume, subjects relaxed totally, with the glottis open, against an occluded airway. Relaxation pauses lasted four to six seconds. Pressures and volumes were recorded for each volume increment. Trials were performed in air (control) and during immersion, using erect (seated) and horizontal postures.

Static pressure–volume manoeuvres were performed in the inspiratory direction to: (a) reproduce protocols of earlier researchers (Jarrett 1965, McKenna *et al.* 1973), (b) to avoid condensation within the pneumotachograph, and (c) to facilitate use of a regulator to reduce the effect of negative pressure breathing during immersion.

Integration of respiratory flow was adopted, rather than the inspiration of a known volume from a spirometer, as used by Jarrett (1965) and McKenna *et al.* (1973). This provided the capacity for subjects to inspire different volumes on successive trials, and to allow total immersion without excessive negative pressure breathing. Jarrett (1965) used inspired volume increments of 0.5 litres, and accepted data where relaxation pressures were consistent. The current procedure lacks the advantage of checking reproducibility till trials were completed, however, it possesses several advantages over the former technique:

(1) pressure–volume reproducibility was verified by total curve consistency, with random volume variations along the curve providing a more acute appreciation of its true shape;

(2) the experimenter could regulate inspired volumes to provide data at given points - this was particularly important in determining pressures around the relaxation volume (V_R);

(3) subjects had no knowledge of the volume inspired, therefore removing subjective bias. Constant inspired volumes may have induced the reproduction of a previously learned relaxation pressure, which may, or may not, be the true relaxation pressure;

---

[12] The calibration pumping error resulted in overestimation of lung volumes during trials due to an underestimation of flow during calibration.

(4) the procedure was technically easier to perform; and

(5) the experimenter was unable to bias subject performance, since he had no knowledge of results.

Subject training consisted of four phases:(1) verbal description of the task and methods of obtaining glottal control and anticipated curve shape; (2) practice at reproducing residual volume (RV) and total lung capacity (TLC) relaxation pressures using visual feedback from a water manometer; (3) practice at performing inspirations and relaxations with visual feedback; and (4) practice performing small inspirations and relaxations without visual feedback. Progression through the stages was controlled by the subject's capacity to consistently reproduce RV and TLC relaxation pressures, and to consistently provide step increments in relaxation pressure without backward pressure movements. Subjects passed all stages prior to commencing the trials in air. Progression varied between subjects (25–90min), and some preferred not to use visual feedback [13]. The four phases were repeated preceding immersion trials, but only a few subjects required practice at phase two.

Agostoni and Mead (1964) stated one in three people were *good relaxers* during the static pressure–volume manoeuvre. In the present study 29 were initially tested to provide 20 successful subjects. Present success was attributed to non–random subject selection designed to provide a high number of divers. Many current subjects had previously participated in other experiments, were comfortable in experimental situations and found it easy to relax. Subjects were rejected for failure in any of the four preliminary phases.

Trials were preceded by 3–5 rapid breaths, at slightly larger than normal tidal volumes, to lower blood CO_2 tension and, reduce respiratory drive. Three normal tidal volume breaths followed this tachypnea to remove volume history influences on compliance (Ferris and Pollard 1960). Expiration to RV followed. During immersion, transition to RV was determined by rapid air expulsion from regulator 2, and RV attainment by cessation of bubbles from this regulator. At RV subjects were isolated from atmospheric air by closing tap 1 (control trials) or from regulator 2 by closing tap 2 (immersion). On verbal command (head taps during immersion) subjects gently inspired when tap 1 was opened and relaxed when it closed. To prevent large inspirations subjects were trained to resist inflow at low relaxation volumes. Inspiration was continued until subjects felt the tap close, at which point they relaxed. To prevent accidental gas loss, tap 1 was not opened till the experimenter observed commencement of inspiratory effort (change in manometer pressure). Using this sequence, subjects were

[13] It was found that subjects with excellent glottal control were those with previously developed motor skills (*e.g.* singing and wind instrument playing).

able to provide 5–10 pressure–volume holds between RV and TLC. Trials were repeated 5–7 times to provide adequate data. Unsatisfactory trials were repeated. An electronic trigger was activiated at each relaxation point and recorded on a separate computer channel to identify points of relaxation.

Subjects performed trials in two postures (seated upright and horizontal) in air and immersed. Seated trials took place on the same chair with the back held vertical and hips flexed at ~90°. Horizontal control trials in air were performed supine to avoid thoracic compression effects upon compliance. Horizontal immersion trials were completed with the subjects prone, and with the regulator in the sternal plane. This produced a situation where subjects were exposed to a continuous positive pressure.

Some supine trials were performed underwater during preliminary studies and prior to using the diving hood. Subjects reported difficulty holding the mouthpiece in place, and resultant static pressure–volume curves were generally not as reproducible as prone data.

To reduce buoyancy, subjects placed a weight belt over the thighs while sitting, and buttocks while prone. The experimenter held their head position constant during trials. This reduced, but did not prevent the chest from rising at high lung volumes. Attempts to secure the chest were deemed unethical and not conducive to measurement of respiratory compliance.

At the commencement of immersion trials, submersion depths of the sternal notch (upright) and sternal plane (prone) were carefully checked and recorded. Regulators were then positioned at that depth.

2.1.5 Calculations

Pressure measurements represented differential pressures between alveolar (measured at the mouth) and atmospheric (control), or atmospheric plus compensator pressure (immersion trials). During air trials the entire body was at atmospheric pressure, but during immersion a vertical pressure gradient (equivalent to atmospheric plus hydrostatic pressure) acted over the body.

Computer programmes were written to perform calculations. Using the electronic trigger points as relaxation markers, the programme integrated flow between consecutive markers, and determined

pressure over a one second period (50 samples) two seconds after each marker point.

Derivation of alveolar pressures relative to anatomical locations during immersion was performed as follows:

$$P_{measured} = P_{mouth} - P_{comp} \qquad \textbf{Equation 1}$$

where:

P_{mouth} = mouth pressure,

P_{comp} = compensator pressure.

Both pressures are measured relative to atmospheric pressure (P_A).

$$\text{pressure required} = P_{mouth} - P_{sternal} \qquad \textbf{Equation 2}$$

where:

$P_{sternal}$ = pressure at sternal notch depth.

We know $P_{sternal}$ and P_{comp}. Thus, combining equations 1 and 2:

$$P_{mouth} - P_{sternal} = (P_{mouth} - P_{comp}) + (P_{comp} - P_{sternal}) \qquad \textbf{Equation 3}$$

Since $P_{mouth}-P_{comp}$ is the actual pressure measured during each trial, equation 3 simplifies to:

$$\text{Pressure relative to sternum} = P_{measured} + (P_{comp} - P_{sternal}) \qquad \textbf{Equation 4}$$

Inspired volumes were corrected for absolute ambient pressure (*i.e.* atmospheric, pressure at the regulator, alveolar pressure and pressure at the anatomical reference point), and for apparatus dead space effects using the Boyle–Mariotte Law.

$$P_1 V_1 = \text{constant} \qquad \textbf{Equation 5}$$

If V_1 represents integrated volume, and V_2 represents the actual lung volume at alveolar pressure.

Then:

$$V_2 = P_1 \cdot V_1 / P_2 \qquad \textbf{Equation 6}$$

where:

P_1 =atmospheric + regulator pressure,

P_2 = atmospheric + Pmouth.

Alveolar air is compressed and decompressed by changes in ambient and alveolar pressure. Apparatus dead space (V_D) stores compressed gas and returns gas to the lungs during decompression, thus V_D must be considered a component of V_2. RV is also a component of V_2 since it is always present in the lungs. Substitution of terms into equation 6 gives:

$$V_2 = \frac{(P_A + Preg)(V_1 + RV + V_D)}{(P_A + Pmouth)} - RV - V_D \qquad \textbf{Equation 7}$$

Integrated volumes were then corrected to BTPS (equation 8). Pressure corrections included allowance for absolute pressure at anatomical reference sites, and alveolar pressure during static holds. This was essential since immersion compressed gas volumes, and alveolar pressure acted either to decompress (if negative) or further compress (if positive) lung volumes measured by inspired flow integration.

$$V_1\ BTPS = \frac{(P_A + Pmouth + Psternal - P_{H_2O})\ V_2 \cdot 310}{(P_A + Pmouth - 47.1)(273 + T_A)} \qquad \textbf{Equation 8}$$

Before analysis of compliance curves was undertaken, data points were rejected as errors according to the following criteria: (a) false positive/negative pressures (*e.g.* negative pressure measured at lung volumes $>V_R$), (b) points when glottal closure or swallowing occurred during trials, as recorded by the experimeter or detected from compliance curve shape (*e.g.* pressure fluctuations >0.49kPa ($5cmH_2O$) from mean curve), and (c) points of poor relaxation (as determined by curve analysis). Care was taken to avoid unnecessary exclusion of pressure–volume data point since Henry (1949) showed measurement

error to be inversely proportional to the square root of the measurement numbers used to derive a score. Data averaged from many observations possesses smaller error. Random error dictates that points will disperse about the *true* value. Inclusion of this dispersion algebraically reduces measurement error while providing a more acute appreciation of the *true* value, which always possesses some uncertainty (Beers 1957).

Compliance curves were examined using polynomial regression analysis (Dixon 1983) of the general form:

$$y = \beta_0 + \beta_1 x + \beta_2 x^2 + ... + \beta_j x^j + e \qquad \textbf{Equation 9}$$

Conventionally, static pressure–volume curves are plotted with pressure on the abcissa. Statistical convention dictates that independent variables appear on the abcissa. Static pressure is dependent upon lung volume, thus analysis was performed with volume as the independent, and pressure as the dependent variable.

Curves from upright and prone postures during control and immersion trials were separately analysed using orthogonal polynomials to derive the least squares curve fit of independent to dependent variables. Coefficients of each regression equation were entered into three analysis programmes written for this investigation. The first enabled numeric solution of the regression equation to derive lung volume at zero pressure[14] (*i.e.* V_R). The second differentiated the equation to provide respiratory elastance[15], compliance, and static volume and pressure at increments of 1% of vital capacity (VC), over the volume range from residual volume (RV) to total lung capacity (TLC). The third programme integrated pressure with respect to volume, thus providing respiratory elastic work[16] over a given volume range.

---

[14] Transcendental equations cannot be resolved analytically since x cannot be isolated. A value of x (volume) may only be attained numerically. Solution accuracy was set at ±1ml, which was beyond the capacity of the measurement system, but was deemed appropriate (Kuo 1972).

[15] Differentiation provides a measure of curve slope (*i.e.* dy/dx or dP/dV) which is respiratory elastance. When drawn according to physiological convention (pressure on abcissa), curve slope equals respiratory compliance.

[16] Area below the pressure–volume curve has the dimension of work (Chapter One).

Lung *centroid* pressure was determined from isovolume (V_R) pressure–volume curve displacements, measured between control and immersed curves, along the pressure axis. Pressure data were differential pressures between P_{alv} and local pressure at the sternal notch. If pressure is expressed in imperial units (cmH_2O), one may represent lung *centroid* as a vertical displacement below the sternal notch (upright immersion), or above the sternal plane (prone immersion), since these distances equate with the pressure head produced by an equivalent column of water.

In accordance with Jarrett (1965) curve shifts were expressed relative to upright control V_R. This method was chosen since the supine V_R may have been located on the lower bend of the curve, at which point curve shifts may not have been parallel. Use of mid–curve isovolumes ensured curve shift analysis over the approximately linear part of the curve. Over this region the curve shifts were expected to be parallel, thus providing a P_{LC} that was relevant to a greater range of lung volumes. Expressing P_{LC} relative to the upright control V_R provides a measure of the pressure required from the breathing apparatus to return respiratory mechanics to operating dimensions and pressures experienced under normal ambient conditions at sea level.

No previous work has investigated the prone lung *centroid* pressure. To facilitate comparability with the literature, prone P_{alv} was also expressed relative to the spinous process depth opposite a mid–sternal point. This was accomplished *post hoc* using anterior–posterior chest dimensions. These data were taken as approximating the supine observations of others (Paton and Sand 1947, Hong *et al.* 1960, Jarrett 1965, Craig and Dvorak 1975).

2.1.6 Analysis

Analysis was based on a repeated measures experimental design, with subjects providing control and immersion data. Statistical analyses took the form of paired or correlated t–tests with *a priori* levels of significance set at a probability of 0.05. Where significant changes were not observed, computations of statistical power were performed. Statistical power (ø) is defined as the probability of correctly accepting an experimental hypothesis (Keppel 1973, Gehring 1978).

2.2 RESULTS

2.2.1 Characteristics of subjects.

Physical characteristics of subjects are detailed in Table 2.2. Twenty subjects were trained to provide reproducible relaxation pressure–volume curves in air, however, three could not provide satisfactory pressure–volume data during immersion, in either the upright or prone postures. These subjects have been excluded from all data reported. [17]

2.2.2 Pressure- volume curves obtained in air.

Pressure–volume curves in air were constructed from an average of 28 (±8) data points, over the volume range from residual volume (RV) to total lung capacity (TLC). Typical curves are illustrated in Figure 2.5 [18]. Data points on these curves reflect relaxation pressure–volume curve reproducibility.

Total respiratory compliance for each subject, in upright and supine postures (when measured relative to upright relaxation volume) were within expected normal ranges[19] (Table 2.3). Adopting a supine posture translocated compliance curves rightward along the pressure axis (Table 2.3), producing a mean pressure increase of 0.67kPa (SEM=0.08kPa[20]) at a volume equal to the upright relaxation volume[21] (Figure 2.6).

When computed over one litre above the upright V_R, compliance systematically increased from 1.84 $l.kPa^{-1}$ (SEM=0.17) in the upright position, to 2.12 $l.kPa^{-1}$ (SEM=0.18) while supine (Table 2.3, $p<0.05$). Such an increment serves to verify technique validity, since Agostoni and Mead (1964) have shown supine postures to reduce gravitational effects upon the chest wall, facilitating easier expansion. It was concluded that upright and control data were reproducible and valid measures of total respiratory compliance.

---

[17] Subject numbers allocated in Table 2.2 remain consistent throughout this chapter.

[18] Pressure–volume or compliance curves are reported according to physiological convention, with pressure on the abcissa. The upper asymptote was not always observed, since data collection was primarily directed at the pressure–volume points between RV and about 2 litres above the V_R.

[19] Normal total respiratory compliances fall in the range 0.9 – 4.0 $l.kPa^{-1}$ (Cotes 1979).

[20] $\bar{x}$ = mean, SEM = standard error of the mean.

[21] This volume corresponds to FRC, and is defined as the volume achieved during total relaxation with the glottis and airway open.

Table 2.2: **Physical characteristics of subjects.**

SUBJECT	Age (yr)	Height (cm)	Stem height (cm)	Mass (kg)	Vital capacity (l.BTPS)	Chest circum. (cm)	Sternal length (cm)	A-P chest (cm)	Diver
1	26	179.4	96.7	72.2	6.09	91.5	21.5	17.7	Y
2	24	189.7	99.7	84.4	6.32	97.0	24.3	20.4	Y
3	26	181.5	97.2	79.3	6.80	100.5	24.5	21.5	Y
4	26	177.7	97.1	73.9	6.00	90.0	21.0	21.7	Y
5	22	185.3	99.4	74.8	6.38	98.9	22.3	19.1	Y
6	26	171.4	87.1	68.6	5.36	94.5	18.0	19.6	Y
7	39	180.4	93.7	95.0	5.24	108.6	22.7	24.4	N
8	26	179.1	94.8	83.1	6.63	94.0	21.0	20.6	Y
9	25	180.2	96.5	75.1	6.10	95.0	21.4	20.1	Y
10	32	184.5	96.1	77.4	6.58	101.8	23.5	21.1	Y
11	32	186.2	95.8	81.1	5.87	99.4	24.7	21.7	N
12	31	181.8	96.6	73.8	5.66	92.0	19.8	18.2	Y
13	28	184.5	92.6	89.0	6.00	102,5	23.4	24.1	Y
14	24	176.0	96.2	78.6	6.22	95.0	23.0	22.3	N
15	25	171.0	90.5	66.2	5.80	98.2	16.2	18.8	N
16	43	182.5	96.2	76.6	6.36	93.0	20.6	21.2	Y
17	23	181.8	95.2	82.5	6.21	101.1	18.1	21.4	Y
$\bar{x}$	28.1	180.8	95.4	78.3	6.10	97.3	21.5	20.8	
SEM	1.4	1.2	0.7	1.7	0.10	1.2	0.6	0.4	

Abbreviations: A-P = anterio-posterior, circum. = circumference, Y = yes, N = no.

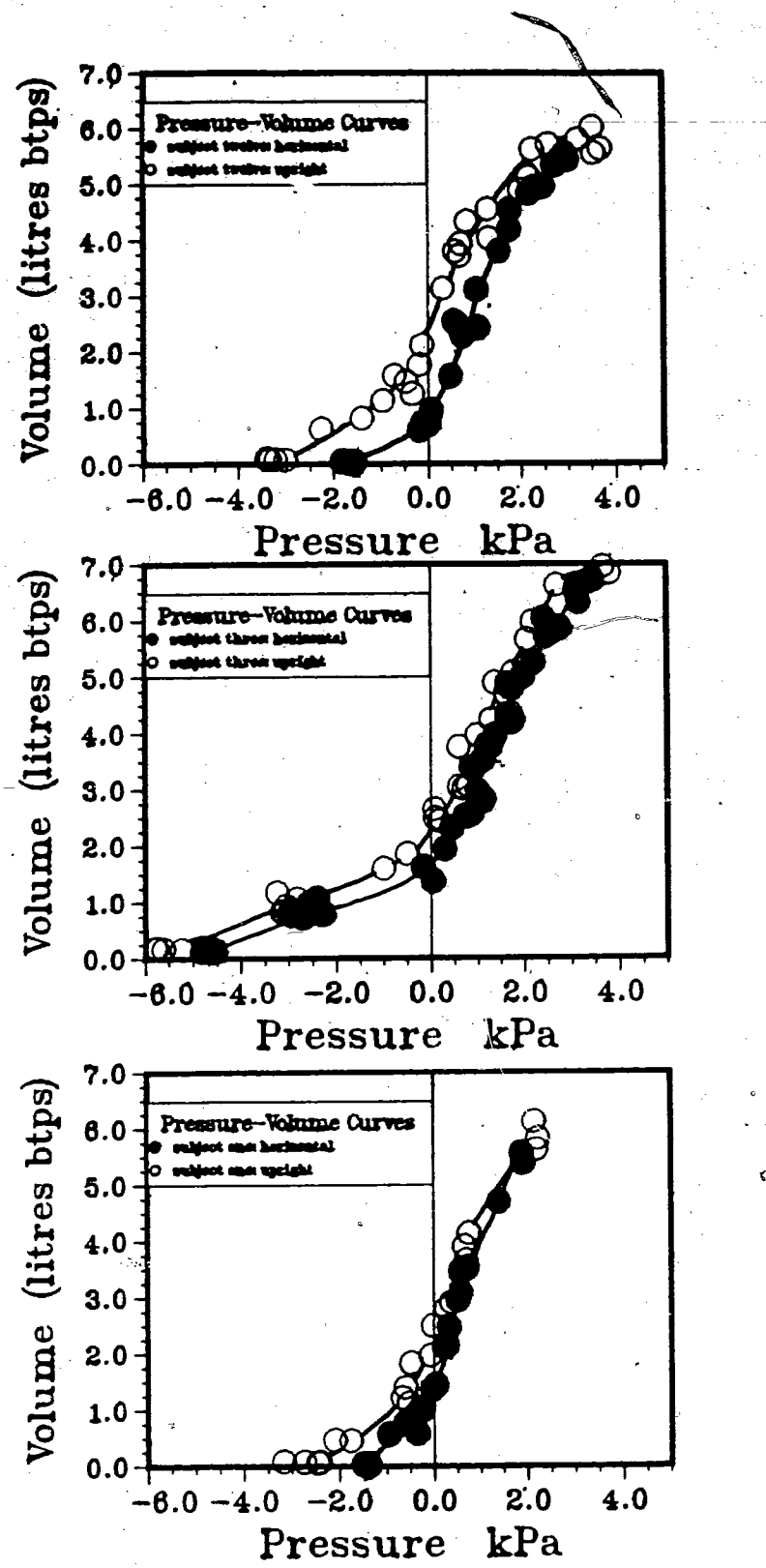


Figure 2.5: **Representative total respiratory pressure–volume curves obtained while upright and supine in air.**

Table 2.3: **Total respiratory compliance for subjects upright and supine in air.**

SUBJECT	Upright compliance	Supine compliance B	Supine compliance A	Curve shift
1	1.95	1.93	2.91	0.30
2	2.03	1.18	2.32	0.67
3	1.16	0.72	1.45	0.61
4	2.74	2.17	2.53	0.24
5	1.10	0.44	1.24	1.34
6	2.23	0.67	1.20	0.67
7	1.14	0.42	2.00	1.11
8	2.50	1.21	2.67	0.42
9	2.02	0.73	2.26	0.83
10	1.19	0.68	1.29	0.34
11	2.55	1.24	2.84	0.59
12	2.11	0.91	2.34	0.72
13	3.30	1.21	3.81	0.46
14	1.15	0.75	1.50	0.57
15	0.96	0.48	1.07	1.34
16	1.41	0.34	2.49	0.78
17	1.72	0.78	2.08	0.44
$\bar{x}$	1.84	0.93	2.12	0.67
SEM	0.17	0.12	0.18	0.08

Compliance units = kPa^{-1}.

Upright and supine compliance A (column 4) were obtained throughout a range of 1 litre from upright relaxation volumes. Supine compliance B (column 3) was derived from a mean of 5 points (representing 5% of VC) above supine relaxation volume. Curve shifts (kPa) represent positive, isovolume displacement of the supine curve relative to the upright curve along pressure axis (measured at upright control relaxation volume) due to postural change.

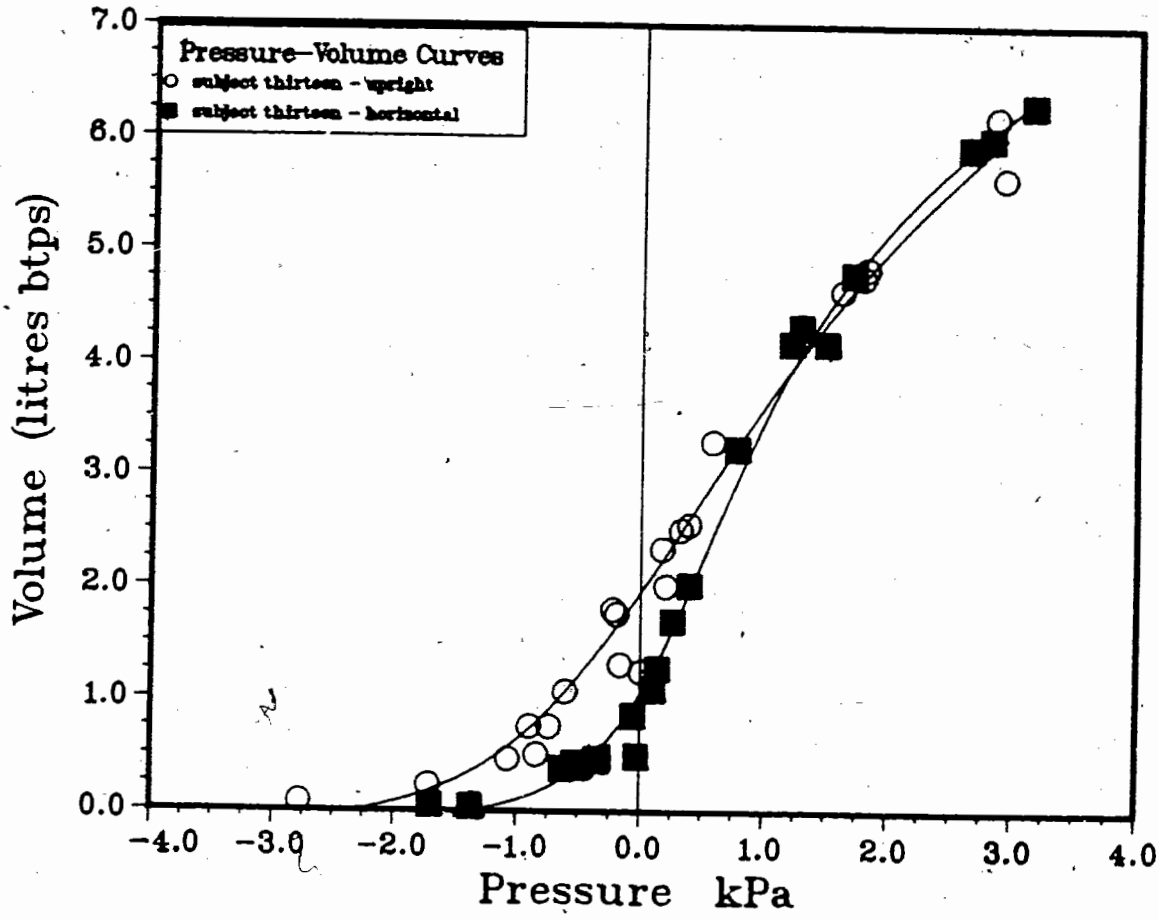


Figure 2.6: **Displacement of the static pressure–volume curve in air due to postural change.** [Curves were obtained from subject thirteen.]

2.2.3 Centroid determination without facial counter-pressure.

Nine subjects took part in trials breathing from a mouthpiece and wearing only a diving hood to support the cheeks against bulging during static holds at high lung volumes. Jarrett (1965) reported bulging prevented relaxation due to muscular effort required to hold the mouthpiece. Unlike Jarrett's data, where pressures remained constant with increased lung volume, subjects in the present study demonstrated highly reproducible pressure-volume data (Figure 2.7), though they reported a tendency to eject the mouthpiece. Holding the rigid breathing tube (Figure 2.3) on either side of the mouthpiece prevented expulsion, while tubing buoyancy aided retention during prone trials.

Pressure-volume curves were constructed from an average of 32 (SD = 7) data points from RV to TLC. Compliance curves proved highly reproducible (Figure 2.7), indicating subjectively consistent relaxation. Respiratory compliance (computed over one litre from control, upright V_R) systematically increased while upright, and decreased while prone (Table 2.4). Differences were non-significant (upright: 1.98 l.kPa^{-1} (SEM = 0.26) in air versus 2.12 l.kPa^{-1} (SEM = 0.24) immersed; and horizontal: 2.15 l.kPa^{-1} (SEM = 0.33) in air versus 1.83 l.kPa^{-1} (SEM = 0.28) immersed; $p>0.05$, $\phi = 0.56$ and 0.55 respectively[*]). Figure 2.8 shows increased upright compliance, which may be due to inspiratory muscle tone, preventing attainment of an *appropriate* positive relaxation pressure in the upright posture. Prone relaxation pressures would tend to become more negative underwater (relative to sternal plane), thus expiratory muscle tonus may have prevented total muscle-relaxation at RV. These observations support Jarrett's (1965) comments concerning use of facial counter-pressure to aid relaxation (*e.g.* the use of an oronasal mask).

Lung *centroid*, determined from group mean isovolume compliance curve displacements (control upright V_R) during immersion, was found to be 9.5cm (SEM = 1.59) below the sternal notch for upright subjects, and 8.7cm (SEM = 1.88) dorsal to the sternal plane in prone subjects. These represent a P_{LC} of 0.93 and -0.85kPa respectively. It is suggested that poor relaxation spuriously elevated the experimental V_R, and depressed positive relaxation pressures while upright (Figure 2.8).

---

[*] ϕ = statistical power, probability of correctly accepting experimental hypothesis (*i.e.* $1-\beta$) (Keppel 1973, Gehring 1978).

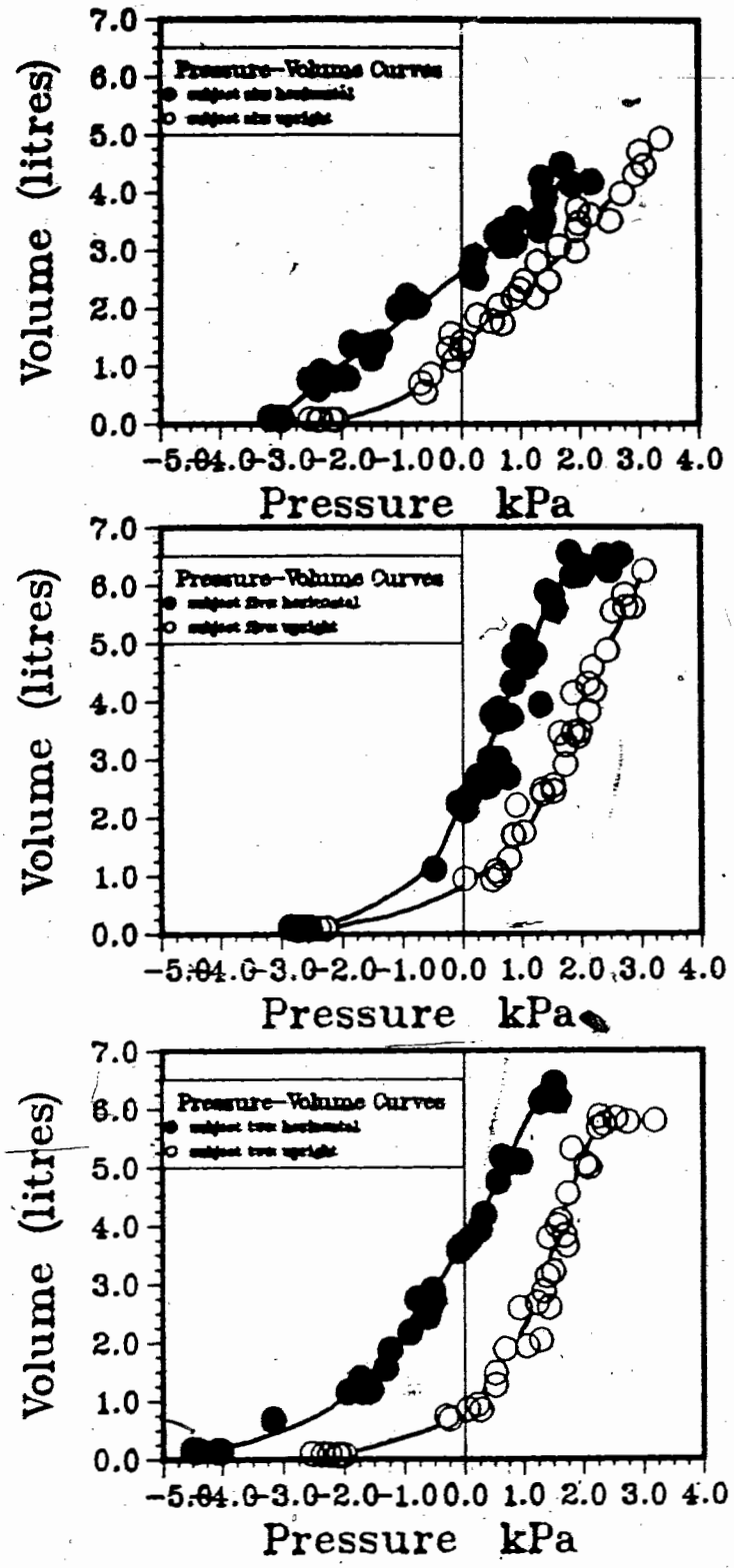


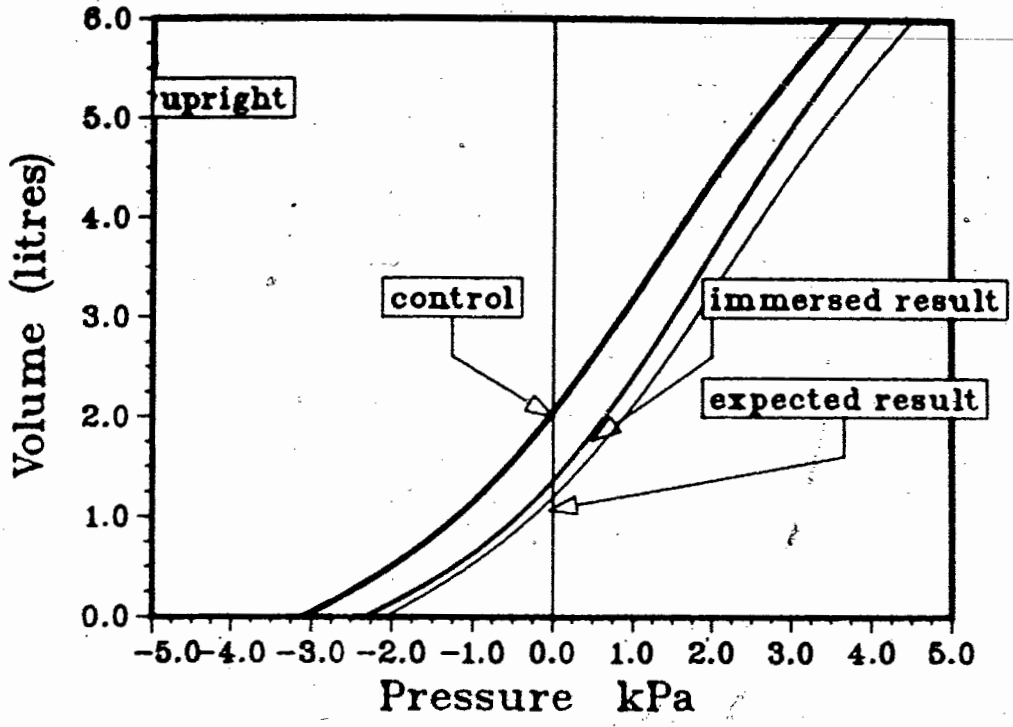
Figure 2.7: Pressure–volume curves obtained during immersion, without facial counter–pressure.

Table 2.4: **Total respiratory compliance for subjects immersed upright and prone without facial counter-pressure.**

SUBJECT	Upright compliance	Curve shift	Prone compliance	Curve shift
1	2.38	3.87	2.06	-4.67
2	2.22	10.94	1.65	-10.87
3	1.59	7.78	1.56	-13.89
5	2.38	11.42	1.90	-0.72
6	1.10	4.39	1.26	-11.30
10	1.41	15.29	1.06	-5.88
11	2.58	17.44	—	—
13	3.53	5.21	3.30	-13.33
15	1.91	9.51	—	—
$\bar{x}$	2.12	9.54	1.83	-8.67
SEM	0.24	1.59	0.28	1.88

Compliance units = $l.kPa^{-1}$.

Upright and prone compliances were computed throughout a range of 1 litre from upright control relaxation volumes. Curve shifts represent positive and negative isovolume pressure displacements (cm.), relative to the control upright relaxation volume, referenced to the sternal notch, and expressed in cmH_2O.



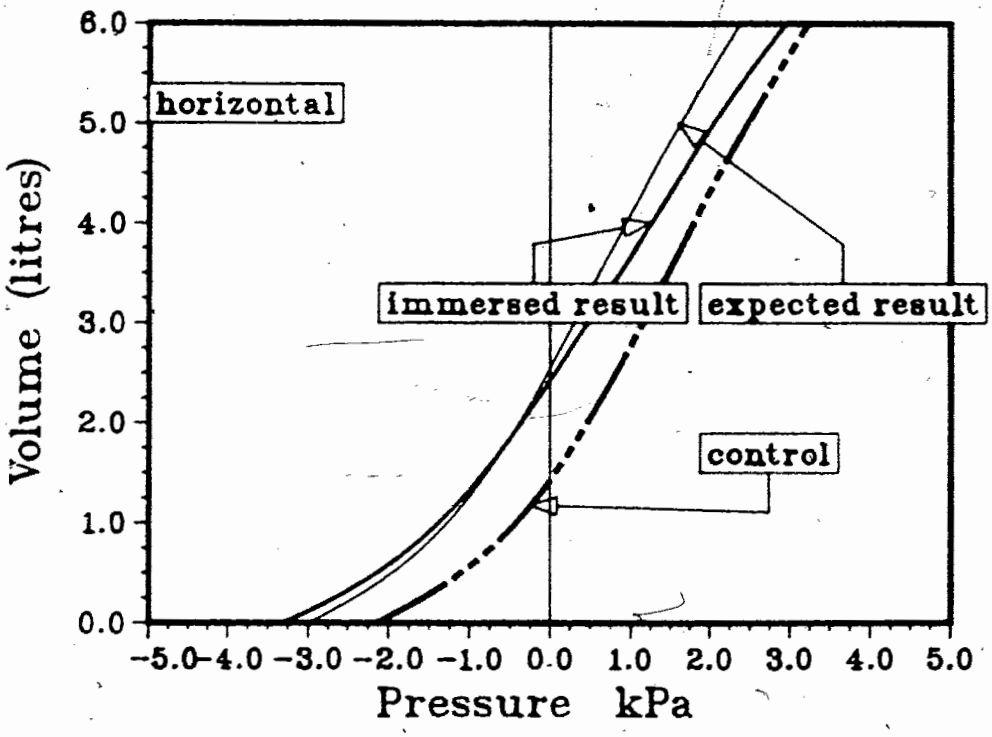


Figure 2.8: **Comparison of total respiratory compliance obtained in air and during immersion without facial counter-pressure.** [Plots are schematic representations of observed data trends. See text for discussion.]

One subject (thirteen) was tested without counter-pressure in two upright positions (head-out with water level at sternal notch, and head under) to gain perspective in comparability of current procedures to those of earlier researchers, who invariably used head-out immersions. Upright lung *centroids* were almost identical: 5.60cm and 5.21cm respectively. Compliance from the former trial was marginally greater (3.89 versus 3.53 $l.kPa^{-1}$). The subject, upon trial completion, reported difficulty retaining the mouthpiece.

One expects lung *centroid* , on the basis of anatomical and previous data (Table 2.1, corrected to a sternal notch reference), to occur deeper. Thus incomplete relaxation was suspected. To test this, a diving mask, capable of providing facial counter-pressure and firmly holding the mouthpiece in position, was adapted, and the experiments were repeated (Figure 2.4).

2.2.4 Centroid determination with facial counter- pressure.

Twenty subjects took part in immersion trials using facial counter-pressure. Only 17 provided satisfactory upright data, and only 13 could produce suitable prone data. In the upright cases, rejected subjects provided less reproducible control data. Two were non-divers, and apprehension and task novelty possibly confounded relaxation. Prone rejections were attributed to sensation reversal. Previous upright immersion displaced relaxation pressure positively. Prone immersion produced a negative shift, resulting in negative mouth pressures until relatively high lung volumes. It is postulated that oropharangeal pressures, disproportionate with inspired volume created false impressions of relaxation failure, or inadequate glottal opening, even though subjects were informed about the new sensations prior to the trials. Trial failure took one of two patterns: (1) identical relaxation pressure with lung volume increments, as reported by Jarrett (1965), and (2) relaxation pressure *hunting* , possibly due to subjects trying to produce preceptibly appropriate pressures (Figure 2.9).

To subjectively evaluate the band mask seven subjects, who took part in trials with and without facial counter-pressure, were asked to rate the mask. Five found facial counter-pressure more comfortable than without counter-pressure, two reported the reverse; four believed facial counter-pressure aided relaxation (two undecided; one negative); and one said additional noise (due to pressure leaks around seals) hindered concentration (six were unaffected).

Pressure-volume curves were constructed from an average of 37 (SD =11) data points from RV to TLC (Figure 2.10). Total respiratory compliance (computed over one litre volume range from control,

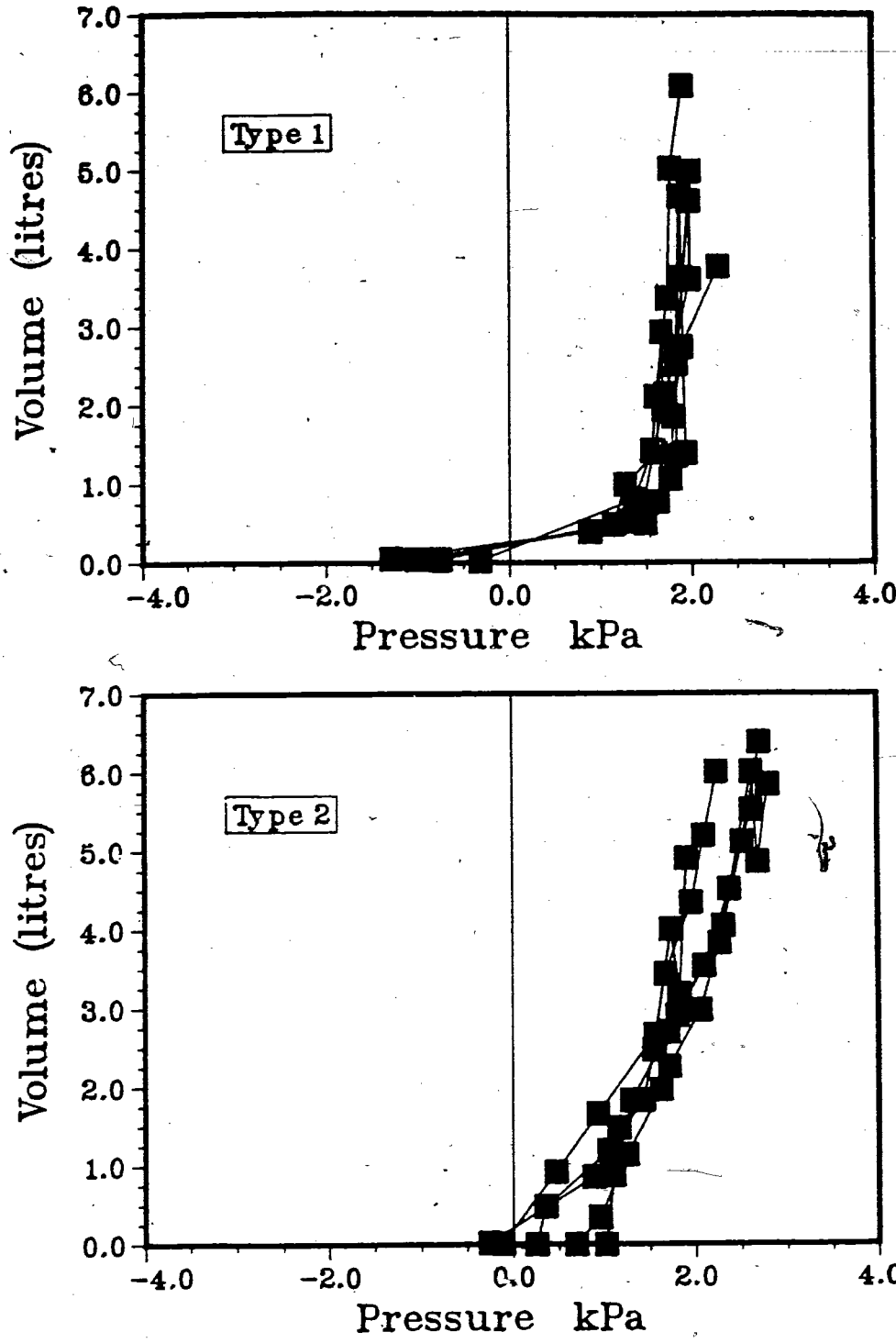


Figure 2.9: **Example of trial failure patterns during immersed pressure–volume measurement.** [See text for discussion.]

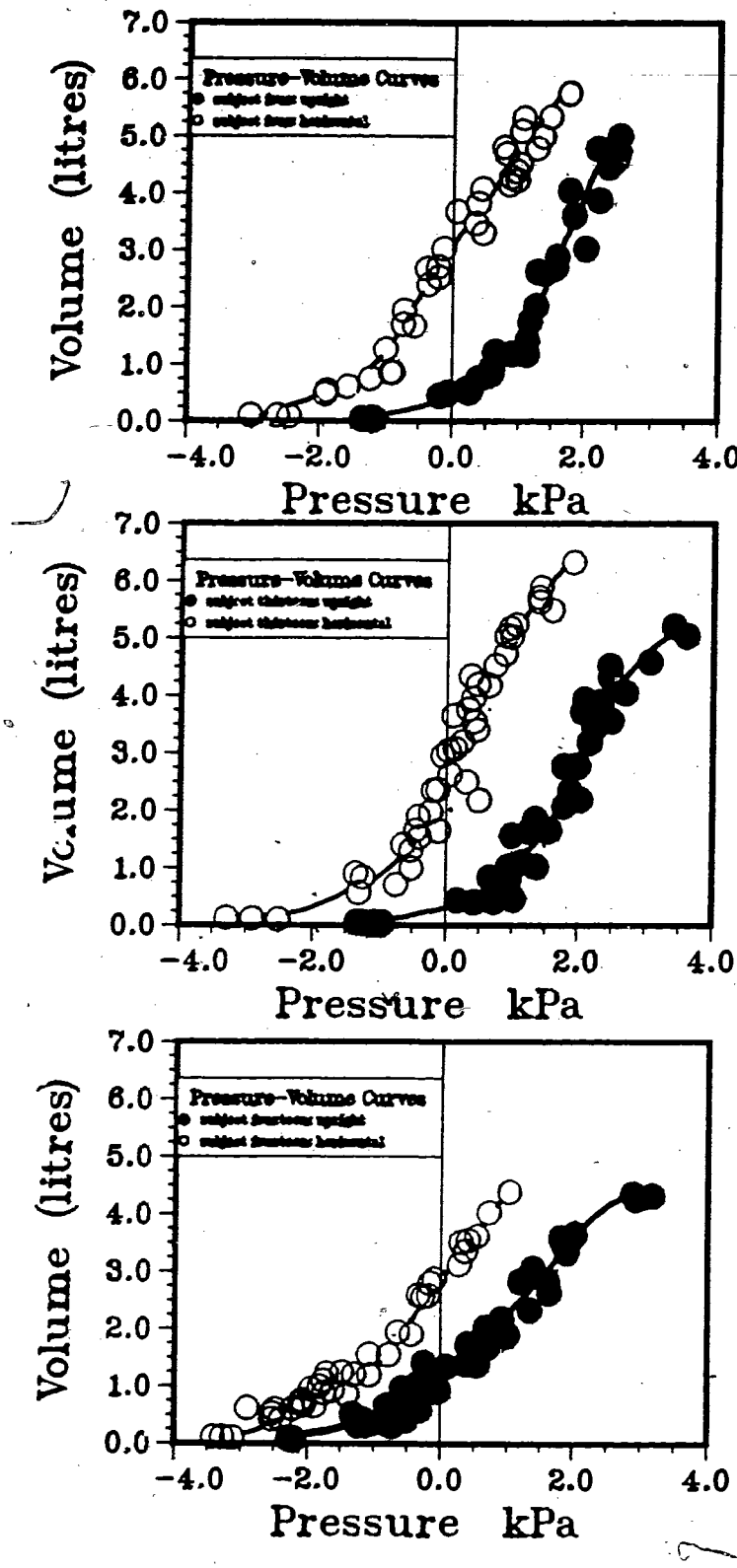


Figure 2.10: Pressure–volume curves obtained during immersion, when using facial counter–pressure.

upright V_R) was not significantly altered by immersion (upright:control = 1.84 l.kPa^{-1}(SEM = 0.17); immersed = 1.87 l.kPa^{-1} (SEM = 0.17); $p > 0.05$, $\phi = 0.55$), though prone immersion produced a reduced compliance (supine control = 2.12 l.kPa^{-1} (SEM = 0.18); prone immersed = 1.87 l.kPa^{-1} (SEM = 0.17); $p > 0.05$, $\phi = 0.56$), however this was created by changes in only 46% of subjects (Table 2.5). Upright and prone compliances were not significantly different during immersion ($p > 0.05$, $\phi = 0.55$).

Lung *centroid* , determined from isovolume compliance curve displacements during immersion, was found to be 13.6cm (SEM = 1.09) inferior to the sternal notch for upright subjects, and 7cm (SEM = 1.27) dorsal to the sternal plane in prone subjects. These represent a P_{LC} of +1.33 and −0.69kPa respectively.

Lung *centroid* determinations obtained with and without facial counter-pressure, on nine subjects who took part in both experiments were compared. It was found that counter-pressure produced a greater P_{LC} in upright subjects (1.33 versus 0.94kPa; $p > 0.05 < 0.1$, $\phi = 0.74$), and in prone subjects (−0.69 versus −0.85kPa; $p > 0.05$, $\phi = 0.55$).

2.2.5 Elastic work increments and compliance curve displacement.

Rahn *et al.* (1946) approximated elastic work, on the basis of geometric shape alone, for upright subjects in air to be ~0.18J. Reanalysing their original data using current curve fitting and analysis programmes provided a value of 0.118J for 0.5 litres [23]. In the current study upright control elastic work averaged 0.103J (SEM = 0.01) over 0.5 litres, which is in close agreement with Rahn's data[24]. Present results revealed that adopting a supine position displaced the pressure-volume curve positively about 0.67kPa (SEM=0.0.08). The change in V_R moved tidal volume (V_T) excursions down the compliance curve (Rahn *et al.* 1946, Figure 2.5), forcing subjects to breathe at volumes with greater elastance (Figure 2.11, Table 2.6). Thus respiratory elastance doubled by adopting a supine posture (from 0.62kPa.l^{-1} (SEM = 0.06) when upright, to 1.38 (SEM = 0.17) when supine; $p < 0.05$).

Upright immersion, with facial counter-pressure, displaced relaxation curves +1.33kPa, producing a significant (46.8%) reduction in V_R (Figure 2.11, $p < 0.05$). Subjects now respired over a region where

[23] Differences may be attributed to precision, and inaccuracy of the 'straight line' assumption inherent in fitting a triangle to a physiological curve.

[24] Elastic work was calculated over 500ml from V_R in each condition. A half litre was chosen since this value is commonly allocated to tidal volume.

Table 2.5: **Total respiratory compliance, and curve shifts, for subjects immersed upright and prone with facial counter-pressure.**

SUBJECT	Upright compliance	Curve shift	Prone compliance	Curve shift
1	3.24	9.73	3.12	-2.47
2	1.64	8.82	—	—
3	1.46	18.29	1.89	-7.07
4	2.84	11.10	2.01	-6.92
5	1.41	15.48	1.56	-4.76
6	1.35	10.71	1.06	-9.52
7	1.70	22.35	1.30	-7.16
8	1.50	15.38	1.15	-0.46
9	2.30	9.62	2.08	-1.22
10	2.29	18.60	1.67	-12.84
11	0.95	11.52	—	—
12	1.75	18.84	2.14	-3.95
13	2.32	14.76	2.91	-14.76
14	1.39	7.75	1.69	-6.54
15	1.50	12.61	—	—
16	2.36	7.50	—	—
17	1.72	17.74	1.71	-13.14
$\bar{x}$	1.87	13.58	1.87	-6.99
SEM	0.15	1.09	0.17	1.27

Compliance units = $l.kPa^{-1}$.

Upright and prone compliances were computed throughout a range of 1 litre from upright control relaxation volumes. Curve shifts represent positive and negative isovolume pressure displacements (cm.), relative to control upright relaxation volume, referenced to sternal notch, and expressed in cmH_2O. Subjects 18-20 are not included due to failure to provide reproducible immersion data (see: Figure 2.9).

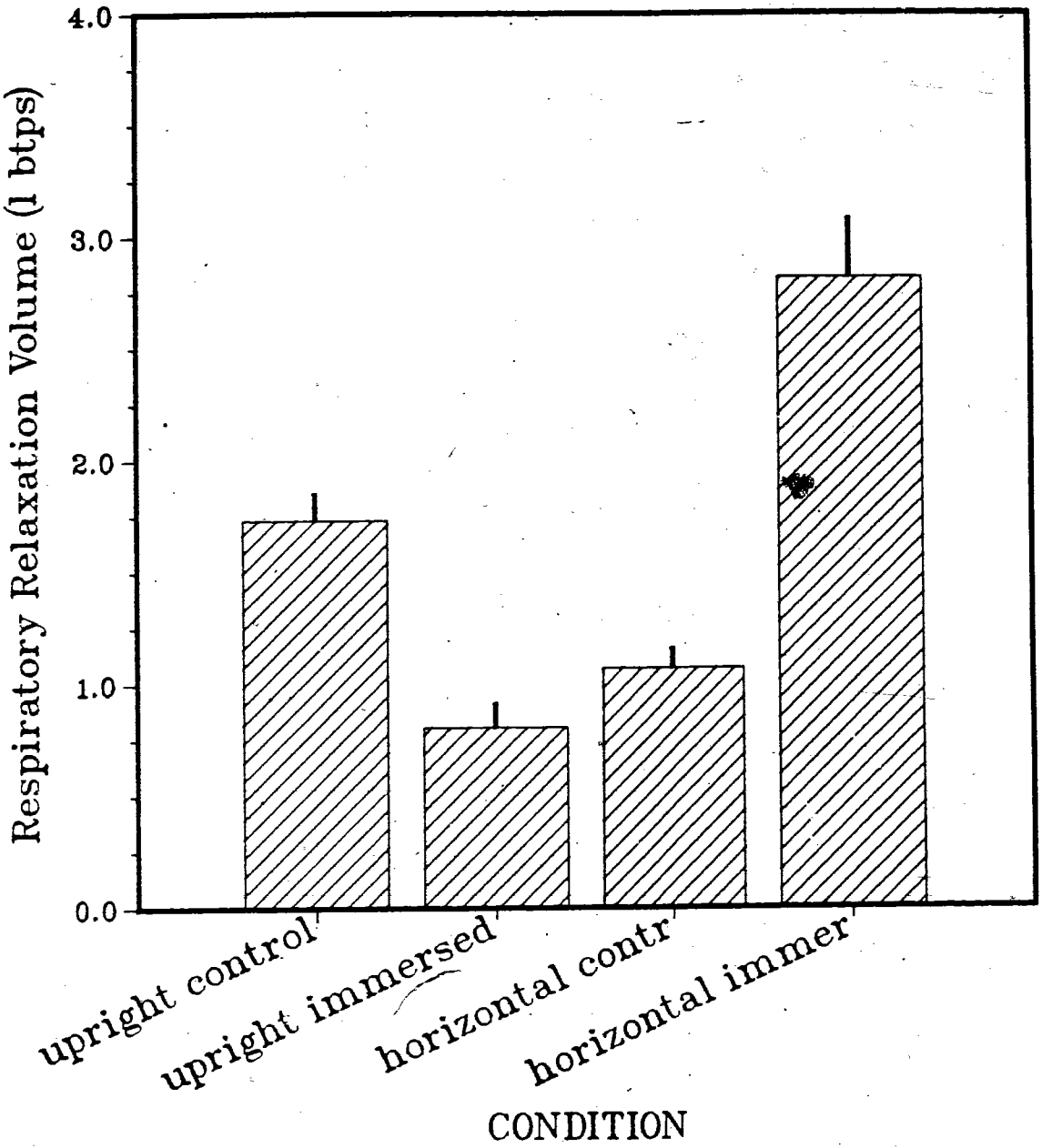


Figure 2.11: **Relaxation volumes during postural change and immersion with facial counter-pressure.** [For immersion trials, subjects were provided with air at the same hydrostatic pressure as the sternal notch. Data represent means and standard errors.]

Table 2.6: **Respiratory elastance in upright and horizontal subjects in air and during immersion with facial counter-pressure.**

SUBJECT	Control upright elastance	Control supine elastance	Immersed upright elastance	Immersed prone elastance
1	0.513	0.518	1.429	0.273
2	0.493	0.847	1.250	——
3	0.862	1.389	2.174	0.478
4	0.365	0.461	1.754	0.515
5	0.909	2.273	1.852	0.529
6	0.448	1.493	1.639	1.136
7	0.877	2.381	1.333	0.730
8	0.400	0.826	1.449	1.124
9	0.495	1.370	1.667	0.578
10	0.840	1.471	3.333	0.592
11	0.392	0.806	1.667	——
12	0.474	1.099	2.041	0.478
13	0.303	0.826	2.564	0.325
14	0.870	1.333	1.099	0.592
15	1.042	2.083	1.449	——
16	0.709	2.941	2.083	——
17	0.581	1.282	2.941	0.599
$\bar{x}$	0.622	1.376	1.866	0.611
SEM	0.056	0.168	0.147	0.072

Elastance units = $kPa.l^{-1}$.
Elastance (1/compliance) was computed over 0.5 litres for upright control and over 5% of VC for other conditions. Calculations commenced at relaxation volume for each condition.
Elastance measures respiratory stiffness.

elastance was three times greater than the upright control (1.87, SEM = 0.15; $p<0.05$; Table 2.6)[25], resulting in a two-fold elastic work increment[26] (Figure 2.12, $p<0.05$).

Prone immersion tripled the V_R when compared with supine control (Figure 2.11, $p<0.05$). Upright control elastance was returned during prone immersion (0.61, SEM=0.07; $p>0.05$, $\phi = 0.55$; Table 2.6), and elastic work similarly returned to control status (Figure 2.12, $p>0.05$, $\phi = 0.55$).

[25] Elastances cited in Table 2.6 (except upright control data) were computed over only 5% of VC. Normal V_T may range between 10–30% of VC. Throughout this volume range elastance undergoes a progressive decrease until the V_T excursion reaches the part of the curve which approximates linearity. Data are therefore higher than mean elastance encountered during V_T.

[26] The apparent discrepency between increments in elastance and elastic work is created by differences in volume range over which each was calculated. The former was computed over 5% of VC, while the latter was obtained over 500ml.

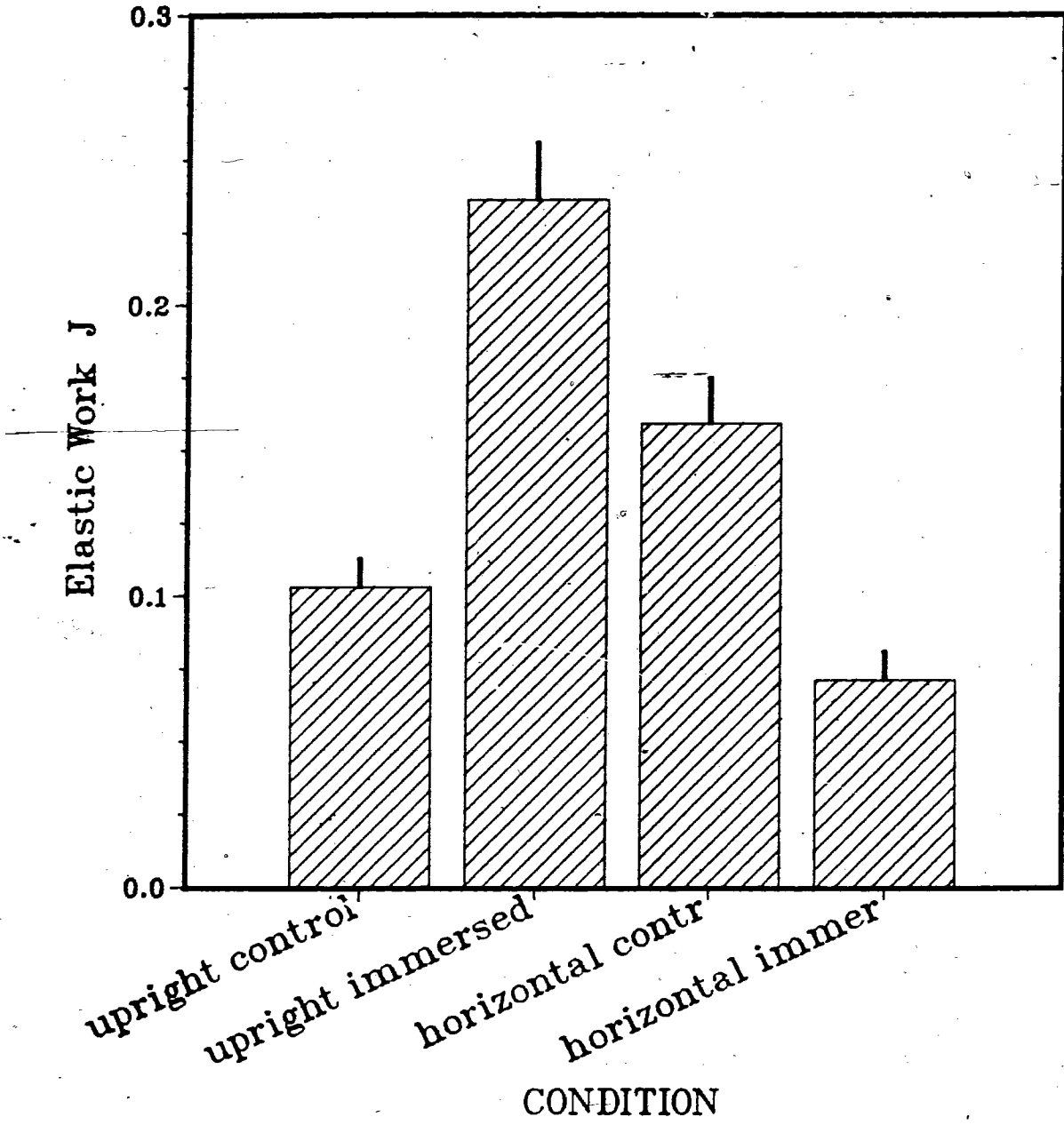


Figure 2.12: **Elastic work during postural change and immersion with facial counter-pressure.** [Data represent means and standard errors.]

2.3 DISCUSSION

2.3.1 Determination of the thoracic centre of pressure.

The major purpose of this investigation was the measurement of the thoracic centre of pressure, for which Paton and Sand (1947) introduced the term lung *centroid* pressure. During immersion P_{LC}, relative to the control V_R, was located at 13.6cm inferior to the sternal notch (seated upright), and at 7.0cm dorsal to the sternal plane (prone), as illustrated in Figure 2.13.

The similarity of control and immersed compliances reflected an anticipated parallel curve shift during immersion, and has been previously used as an index of relaxation reproducibility (Jarrett 1965, Agostoni *et al.* 1966). Furthermore, it reflected the effectiveness of facial counter-pressure in enhancing relaxation, since subjects were better able to attain expected relaxation pressures at both the high and low lung volumes.

Human lungs extend from 3-4cm above the first rib to the sixth costal cartilage (midclavicular line), eight rib (midaxillary line) and tenth rib (at the lateral border of erector spinal) (Gray 1977, McMinn and Hutchings 1977). Using these data one may approximate lung length to be ~30cm, however, the right lung is shorter by ~2.5cm due to a higher diaphragm positioning on this side to accomodate the liver (Gray 1977). Anterior-posterior dimensions vary with the plane of measurement. At the mid-sternal position, using mean anterio-posterior chest dimensions ($\bar{x}$ = 20.8±2.0cm) and thoracic vertebrae thickness of about 6cm (from skeletal measurements, as spinous process to ventral surface plus 1cm for tissue) one may approximate lung and sternal thickness to be about 14cm, though the lungs are conical in shape. Assuming uniform density and symmetry[27], one might expect to find the centre of mass about 15cm inferior to the sternal notch and about 8cm dorsal to the sternal plane.

Prone *centroid* measurements are in direct agreement with the data of Jarrett (1965), and are similar to the adjusted results of Hong *et al.* (1960), and Craig and Dvorak (1975) (Table 2.1), and the anatomical calculations. However, all previous reports of horizontal compliance curve shifts with immersion, dealt with supine subjects. Expressing P_{alv} relative to spinous process depth, and recording shifts relative to supine controls, facilitated a comparison of current data with earlier works. Horizontal lung *centroid* increased slightly to 7.6cm.

[27] Neither assumption would be accurate, however, they are used here simply to derive an approximation of the centre of mass.

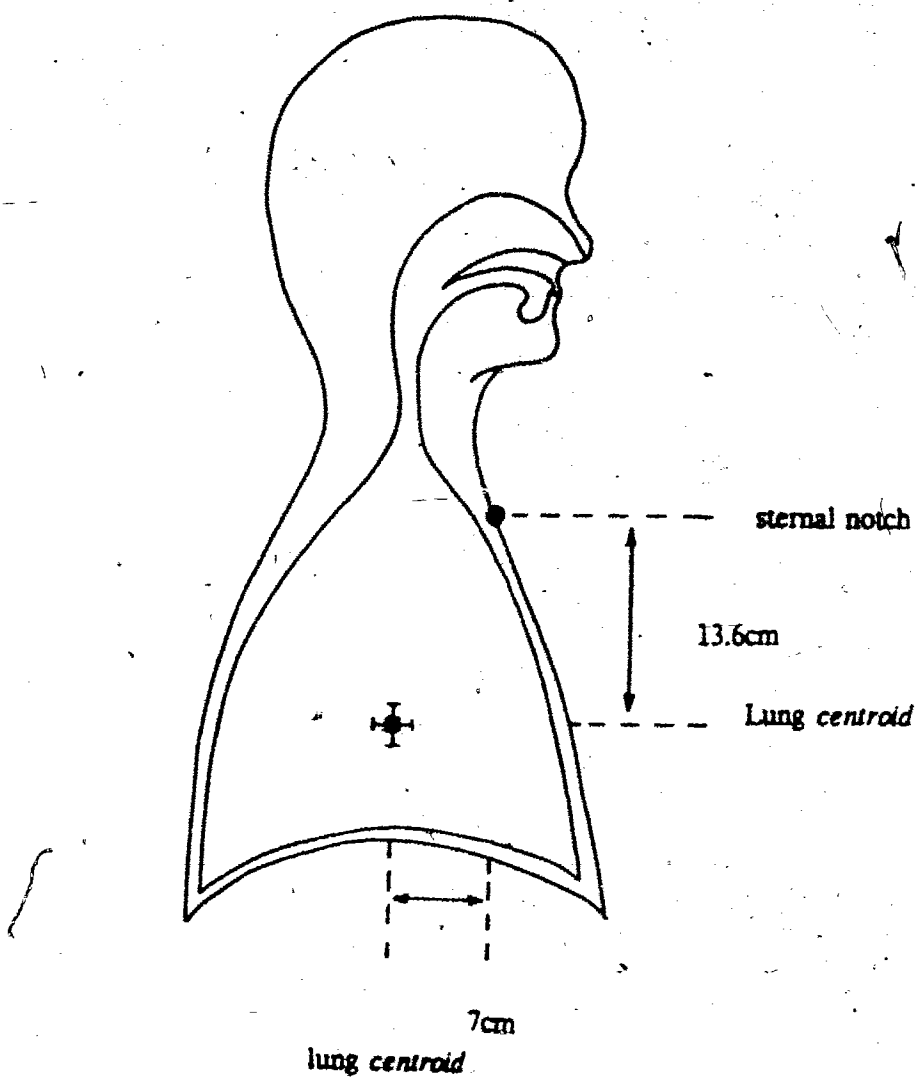


Figure 2.13: Schematic representation of the lung *centroid locus*. [*Centroid* is referenced to the sternal notch. Standard error bars are indicated.]

Present data for the upright lung *centroid* pressure agree with the values obtained when the results of Agostoni *et al.* (1966), Hong *et al.* (1969) and Flynn *et al.* (1975) were corrected to a common anatomical reference site (Table 2.1). Present data also approximated the centre of mass derived from calculations based upon average anatomical dimensions. However, the present results failed to support observations by Jarrett (1965), who placed P_{LC} at 19cm inferior to the sternal notch during upright immersion[28].

Jarrett (1965) obtained total respiratory pressure–volume curves using five standard lung volumes (up to 1.5 litres) above RV. Quadratic equations were fitted to the data[29]. Compliance curve displacements were derived during immersion to the sternal notch, and expressed relative to the control V_R. Fundamental to Jarrett's calculation was the determination of the control V_R. Two points are noted concerning its measurement. First, upright respiratory relaxation produced somewhat low values for the control V_R ($\bar{x}$ = 1.10 litres). Present observations produced a mean V_R of 1.72 litres (SEM = 0.12), in agreement with the literature (Table 2.7). Individual values measured from Jarrett's graphs gave V_Rs of 1.183, 1.142 and 0.980 litres. Residual volumes were normal ($\bar{x}$ = 1.25 litres). While it is not rare to find such a low V_R in normal subjects (Dahlbäck (1975) observed it in 1 out of 6; Bondi *et al.* (1976) found it in 1 in 10; and in the current study it was found in 3 out of 17 subjects), one would not anticipate a low V_R for all subjects[30]. It is suggested that the low relaxation volumes observed by Jarrett (1965), may be due to poor relaxation or glottal closure; both of which give incorrect relaxation pressures. Volume ranges used by Jarrett may have been too small for subjects to differentiate between pressure and volume change. Graphic data presented for two of Jarrett's subjects showed a considerable pressure overlap for each lung volume and for each volume increment, indicating difficulty in reproducing relaxation pressures with consistency (P_R). Some of the current subjects expressed uncertainty in judging total relaxation and glottal opening at volumes around V_R. It might be speculated that Jarrett's subjects had similar indecision, producing possibly a dubious V_R. It is believed that multiple and variable volume increments over the full VC range produce a better appreciation of the total respiratory V_R.

---

[28] Florio (Admiralty Research Establishment, Physiological Laboratories, U.K.) reported inability to replicate Jarrett's higher P_{LC} data in three subjects (unpublished observations).

[29] The data points presented are perhaps better described by a linear function ($r>0.99$), however a quadratic function better replicates physiological status at this point on the curve (Rahn *et al.* 1946).

[30] Assuming an 18% distribution of subjects with low V_R (based on the above observations), the probability of choosing all three with a low V_R would be ~0.005.

Table 2.7: **Respiratory relaxation volumes in air and during immersion** (data from previous investigators).

REFERENCE	Posture	Controls in air (l BTPS)	Immersed (l BTPS)
Jarrett 1965	seated	1.10	—
Beckman *et al.* 1961	seated	1.56	0.56
Agostoni *et al.* 1966	seated	1.86	0.56
Dahlbäck 1975	standing	1.90	0.60
Bondi *et al.* 1976	seated	1.89	0.56
Dahlbäck *et al.* 1978	standing	2.10	0.52
Present study	seated	1.72	0.81

The fact that measurements of lung V_R may be questioned, casts doubt on Jarrett's (1965) P_{LC} determination, since the latter was calculated from immersed compliance curve shifts relative to the lung V_R. One may raise the same questions regarding pressure–volume reproducibility during the immersion trials.

Second, the V_R of Jarrett's upright, immersed subjects does not appear to be at zero pressure. Jarrett's lowest measurement point was 0.5 litres above RV, which yielded a mean relaxation pressure of ~1.23kPa (~12.5cmH_2O). Subsequent investigations have found the mean immersion V_R to be above 0.5 litres (Table 2.7, Figure 2.11). Low V_Rs during immersion have been reported elsewhere (Dahlbäck (1975) found it in 1 out of 6 subjects; Bondi *et al.* (1976) found 1 in 10; Dahlbäck *et al.* (1978) 2 in 5; and in the current study it was observed in 3 out of 17 subjects), but its frequency is again low enough to further justify questioning its appearance in all three subjects[31]. It may have been that Jarrett's subjects failed to relax, and instead were exerting an expiratory pressure. This would account for both the discrepency between Jarrett and the current upright P_{LC}, and for an apparent failure of those subjects to attain the relaxation volume of the respiratory system.

Table 2.5 shows that five of the present 17 subjects (29%) support Jarrett's 19cm locus[32]. Table 2.5 also shows a large subject variability, in agreement with observations of Paton and Sand (1947), and Thompson and McCally (1967). If one accepts Jarrett's data, (for which no individual or dispersion values were reported) and accepts the existence of intersubject variability, then one must expect a study with a larger sample size to reveal this data variability, resulting in a qualitatively similar, but possibly a quantitatively different *centroid* locus. The present investigation has shown this, indicating the possibility that Jarrett's data falls at the upper limit of a lung *centroid* spectrum.

In respiratory pressure–volume diagrams, the abcissa represents a differential pressure. Pressure is classically measured as P_{alv} relative to body surface pressure (Rahn *et al.* 1946, Agostoni and Mead 1964). In air the latter equals P_A; during immersion, surface pressure increases vertically with depth (Figure 2.1). Since one does not know the mean surface pressure at the thorax, hydrostatic pressure is measured at a suitable anatomical location. Studies attempting verification of Jarrett's 19cm *centroid*, must express P_{alv} relative to the anatomical reference point used by Jarrett (*i.e.* sternal notch). However,

---

[31] If low control V_Rs were correct, one would expect low experimental V_R.

[32] Using present occurrence frequencies, the probability of selecting all three subjects with a large P_{LC} is ~0.018.

all previous studies (Table 2.1) have referenced P_{alv} to P_A, without acknowledging that comparison of such data is inappropriate unless a constant immersion depth is used. Data referenced to P_A, is referenced only to the anatomical site at the water surface. Data can be shifted to alter the reference pressure (and hence the anatomical reference) providing its depth is known. In this manner a family of compliance curves may be obtained from one set of data, simply by changing the anatomical reference point. Five research groups (Table 2.1) supported Jarrett's upright P_{LC} data. However, since they lack a common anatomical reference, they have only produced members of this family of curves, with curve shifts representing different reference pressures (Figure 2.14A). Comparison of these curves with Jarrett's data is not valid.

Data from previous studies were reanalysed in order to reference P_{alv} to pressure at the sternal notch[33] , and to normalise curve displacement to the control V_R, as used in this study and by Jarrett (1965). Agostoni *et al.* (1966) calculated curve displacement using the immersed V_R, calculating its shift from the same volume on the control curve. Since curves are not parallel at this level, due to immersion effects on lung, chest and total compliance curves, calculations are not comparable. Using upright control V_R, which coincides with parallel curve sections, displacement was recalculated for all previous investigations. Results of this analysis provided mean values of lung *centroid* between 11 and 17.8cm (Table 2.1). A weighted average of the 13 subjects contained in these five reports provided a lung *centroid* of 12.7cm (Figure 2.14B), which is in agreement with current observations.

Some degree of caution must be exercised concerning data obtained without facial counter-pressure, or some means of preventing voluntary effort being required to retain the mouthpiece during immersion. None of the studies cited (Figure 2.14B, Table 2.1), except Jarrett (1965), reported using such mechanisms. Present results show differences in P_{LC} with and without counter-pressure. It is not possible to evaluate the significance of this observation to previous research, since indices of reproducibility were not reported.

Horizontal lung *centroid* agrees well with Jarrett's data, both when expressed relative to the sternum and the mid-sternal spinous process (7cm, and 7.0±4.6cm respectively).[34] Work by Hong *et al.*

[33] Calculations were performed on mean data since individual and deviation values were absent in all reports. Anatomical distances were considered constant: 'shoulder' to sternal notch = 5cm; 'C7' to sternal notch = 7.2cm (from Craig and Dvorak 1975).

[34] The current study referenced P_{alv} to the sternal plane, while Jarrett (1965) used the sternal notch reference. These are essentially at equivalent immersion depths, unless the chest rises excessively. However, this would only occur at high lung volumes, which were

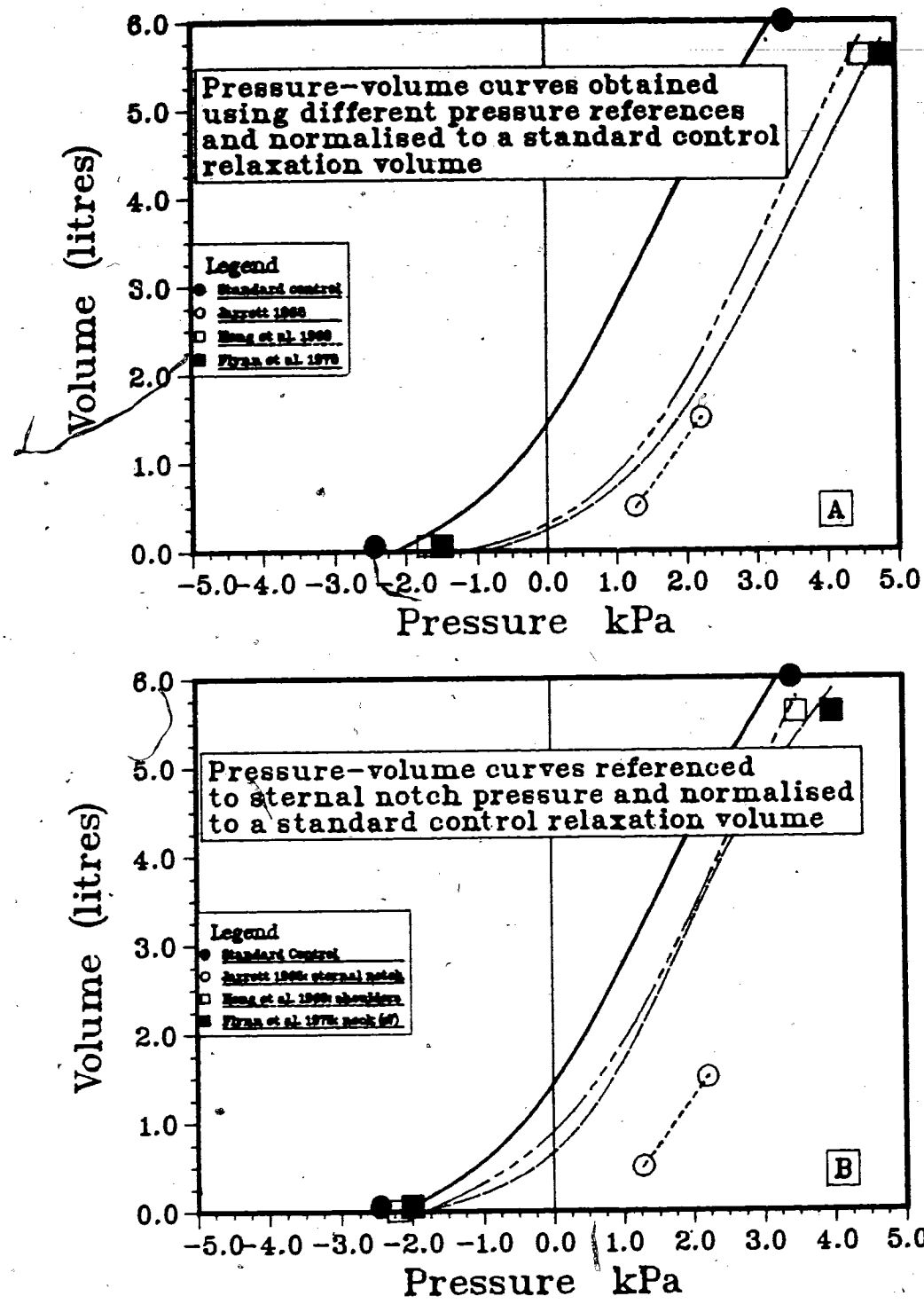


Figure 2.14: **Referencing static alveolar pressure to ambient pressure (A) and to a common anatomical point (B).** [Curves drawn from compliance curves of seated, immersed subjects. Immersion data are expressed relative to a common control relaxation volume. See text for discussion.]

(1960) and Craig and Dvorak (1975) provided similar values, though both reported curve shifts relative to the supine V_R. Corresponding experimental V_R data may fall along the lower bend of the compliance curve. Displacements at such volumes may not be parallel, rendering P_{LC} pertinent to that volume only.

It is concluded that seated and prone P_{LC} displays intersubject variability about respective means of 13.6cm (1.33kPa) inferior to, and 7.0cm (−0.69kPa) dorsal to the sternal notch. However it remains unclear as to the cause of this intersubject variability. Clearly the thorax lacks uniform distensibility, as apical alveoli have been shown to be expanded more than basal alveoli in control states (Glazier *et al.* 1967, Hogg and Nepszy 1969). Thus basal alveoli operate over more compliant portions of their pressure–volume curves (Rahn *et al.* 1946, Agostoni and Mead 1964), experiencing greater volume change during control tidal breathing (West 1965, Milic-Emili *et al.* 1966). Arborelius *et al.*' (1972a) demonstrated that during immersion apical ventilation is further increased while basal alveoli experience varying degrees of closure (Dahlbäck and Lundgren 1972, Dahlbäck 1975, Bondi *et al.* 1976, Prefaut *et al.* 1979).

During immersion thoraco–abdominal hydrostatic compression is uneven, causing elevation of the diaphragm (Agostoni *et al.* 1966, Hong *et al.* 1969) and widening of the rib cage base (Minh *et al.* 1977). Shoulder girdle weight is removed (Agostoni *et al.* 1966), as is the thoracic gravitational loading. Diaphragmatic elevation combines with alveolar buoyancy to displace lung tissue cranially (Arborelius *et al.* 1972a). However, it is unlikely that intersubject differences in lung distension or thoracic compression may fully account for the observed lung *centroid* variance.

When one considers anthropometric variability in combination with respiratory changes during immersion, one cannot readily explain the observed intersubject variability. No correlations existed between lung *centroid* and anthropometric dimensions, and control or experimental lung relaxation volumes or respiratory compliances. It is hypothesised that lung *centroid* was regulated by some, as yet unresolved, combination of respiratory mechanics and perceived relaxation. Subjects were well trained, providing highly reproducible control and immersion data. No trend existed between data reproducibility and the *centroid* magnitude, as subjects producing highly consistent relaxation pressures were located at either end of the P_{LC} spectrum. The role of perceived relaxation could be evaluated using respiratory muscle electromyographic analysis during pressure–volume manoeuvres [35]. It may be possible that

[34](cont'd) presumably above the upright control V_R. It is thus considered that both studies used a common anatomical reference at this isovolume.

[35] A computer controlled system for quasi-static manoeuvres is being developed at Chalmers

relaxation levels are pertinent to the *centroid* locus. High reproducibility may indicate maintainence of constant respiratory tonus during immersion, even after considerable practice and numerous experiments.

Evidence implicating a possible increase in diaphragmatic muscle tone during immersion was provided by Bishop (1963). Anaesthetised cats were exposed to negative breathing pressures during periods of spontaneous respiration. This is at least physically analogous to immersion. As the magnitude of the negative pressure increased, the electromyographic activity of the diaphragm was elevated. Eventually a negative pressure was reached where the diaphragm was active throughout the entire respiratory cycle.

Minh *et al.* (1977,1979) observed greater inspiratory pressures during isovolume respiratory occlusion, attending diaphragmatic stimulation in immersed dogs. Changes were attributed to an elevation of the abdomen, and stretching of the diaphragmatic fibres (Agostoni *et al.* 1966, Minh *et al.* 1977, 1979, Banzett *et al.* 1985, Reid *et al.* 1985), thus shifting the diaphragm towards a more favourable length-tension relationship (Evans and Hill 1914). Reid *et al.* (1985) and Banzett *et al.* (1985) recorded reduced intercostal and diaphragmatic electrical activity attending respiration during immersion. The latter reported substantial intra- and intersubject variance. Both studies concluded that reflex central, inspiratory compensation for diaphragm length increments, reduced innervation to maintain a constant tidal volume.

It is possible, that in well trained, relaxed subjects, variability in inspiratory/expiratory muscle tone may partially account for intersubject variance in the lung *centroid* locus.

2.3.2 Physiological mechanisms producing compliance curve shifts.

Transposition of pressure-volume curves occurs when environmental or physiological perturbations produce changes in P_{alv} at constant occluded lung volumes. Normal respiration occurs up and down the compliance curve. In upright man V_R falls on the linear portion of the curve, so respiratory excursions occur at constant compliance. A positive curve shift lowers the V_R and may lower compliance. Tidal volumes may now occur over a region of elevated elastance (Table 2.6).

Elastic work is performed to expand lung tissue and the chest wall [36], and is represented by the

[35](cont'd) University of Technology, Sweden; Warkander and Dahlbäck (personal communication).

[36] The chest wall includes the rib cage and abdomen, which act in parallel (Agostoni and

area between the pressure axis and the compliance curve over the V_T range (Figure 2.15A). A positive curve shift, for a constant V_T, increases elastic work due to elevated respiratory elastance. Since elastic work is returned during expiration, which is essentially braked or passive at rest, elastic work increments represent heightened inspiratory effort.

In the present study, when breathing air supplied at the hydrostatic pressure of the sternal notch, positive curve shifts (Table 2.5) halved the V_R (Figure 2.11), and caused a two-fold rise in elastic work to 0.24J (Figure 2.12). During immersion to the neck (C7), Flynn *et al.* (1975) found elastic work rose to 0.27J (V_T = 0.5 litre, n = 1), while Hong *et al.* (1969), reported mean elastic work at 0.79J (range = 0.69 to 0.88J) during immersion to shoulders (V_T = 1.04 litres, n = 4).

Elastic work is mainly dependent upon two factors: elastance and V_T (Figure 2.15B). Since neither Hong *et al.* (1969) nor Flynn *et al.* (1975) reported elastance, direct comparisons are not possible. Given that Flynn *et al.* (1975) expressed P_{alv} relative to a point above the sternal notch, which would produce a greater curve shift and elevate elastance more than in the current study, their elastic work is very similar to the present data. Hong *et al.* (1969) used a 1.0 litre mean V_T. Doubling V_T requires about four times (Figure 2.15B) the elastic work. Allowing for this difference their data also agree with present observations.

Accepting that curve shifts result from hydrostatic pressure gradients acting on the body surfaces, one may suggest three factors that are responsible for this curve displacement: (1) thoracic compression, (2) abdominal compression, resulting in cranial movement of the diaphragm, and (3) elevated thoracic blood volume due to a greater hydrostatic pressure acting on the legs.

Hartshorne (1847) reported that immersion elevated thoracic blood volume; though its quantification is relatively recent. Stigler (1911) reported right side cardiac dilation during immersion. This was verified directly by Echt *et al.* (1974), Lange *et al.* (1974) and Risch *et al.* (1978a, 1978b). Cardiac output increments generally attend dilation, Kroetz and Wachter (1933), Arborelius *et al.* (1972b), Löllgen *et al.* (1980), Hajduczok *et al.* (1984) and Krasney *et al.* (1984), though some have reported no rise or even a decline (Hood *et al.* 1968, Rennie *et al.* 1971). Hajduczok *et al.* (1984) reported cardiac output to rise immediately upon immersion to four times the steady state immersion value.

---

[16](cont'd) Mead 1964).

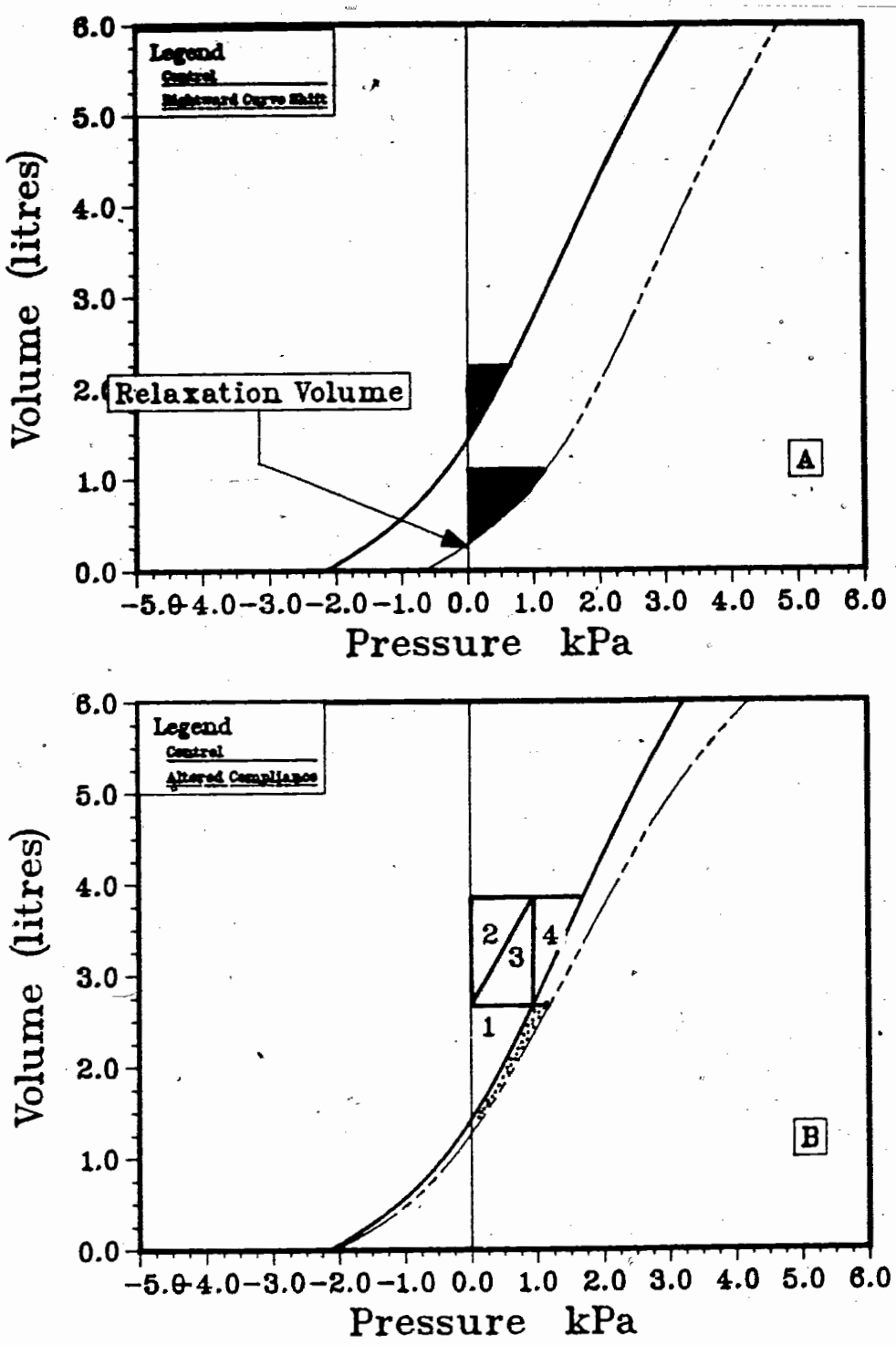


Figure 2.15: **Elastic work of breathing in air and during immersion.** [2.15A: Elastic work changes associated with a positive shift of the total respiratory compliance curve. 2.15B: (1) a four-fold increase in elastic work would be produced by doubling the tidal volume; (2) altered elastic work attending reduced compliance.]

Agostoni *et al.* (1966) postulated that hydrostatic compression and a shift of peripheral blood to the thorax mediated a displacement of the chest wall compliance curve. Recalculation using their data curves and the current isovolume methods, revealed immersion to the xyphoid displaced the chest wall curve 0.41kPa (4.2cmH_2O), while subsequent immersion to the neck/thorax junction produced a further 1.01kPa (10.3cmH_2O) shift. The first immersion would increase thoracic blood volume and elevate the diaphragm, while the second may increase blood volume further, but would primarily provide thoracic compression. Hong *et al.* (1969) observed similar changes in the transrespiratory curve following identical immersions.

Dahlbäck (1978), studied thorax only and head-out immersion, using a modified body plethysmograph, and found thorax immersion produced a significantly greater reduction in expiratory reserve volume. Though total respiratory compliance was not reported, one might speculate, since compliance curve shifts determine resting ERV, that hydrostatic compression was mainly responsible for the present compliance curve displacements [37].

It is concluded that rightward movement of the compliance curve (relative to sternal notch pressure) observed in this project, is attributable primarily to compressive forces on the thorax. Partitioning the curve shift between compression (thoracic or abdominal) and blood volume changes is not possible from the current techniques or from the data available in the literature.

[37] Thorax immersion alone would not be associated with pulmonary vascular engorgement.

CHAPTER 3

CHARACTERISATION OF STATIC PULMONARY MECHANICAL ATTRIBUTES DURING WHOLE-BODY, UPRIGHT IMMERSION BREATHING AIR AT VARIOUS STATIC RESPIRATORY LOADS.

3.1 RESPIRATORY ELASTICITY.

James Carson (1820) reported the earliest recorded investigations of lung elasticity. He described the elastic recoil of the lung measured using a water manometer connected to the airways of dead animals. Pneumothorax caused a positive airway pressure, which was used as an index of lung recoil. Hutchinson (1849–1852) repeated these measurements on fresh cadavers, observing a linear pressure–volume (compliance) relationship. Donders (1853) subsequently postulated that there was an inherent tendency of the lung to recoil towards zero volume.

Although Rahn and his co–workers are recognised as producing the first respiratory pressure–volume diagram (Rahn *et al.* 1946), similar curves were previously, and independently recorded by Romanoff (1910–1911) and Rohrer (1916) from passive, static relaxation against an occluded airway. Jaquet (1908), Bernoulli (1911) and Cloetta (1913) also derived similar curves from lung volume changes accompanying altered breathing gas pressures.

These methods recorded lung elastic recoil under static conditions. Buytendijk (1949) developed the quasi–static technique, where measurements were taken as subjects breathed slowly between total lung capacity (TLC) and residual volume (RV). A third technique permitted an approximate measure of lung elasticity from end–tidal compliance measurement during spontaneous breathing (Neergaard and Wirz 1927, Bayliss *et al.* 1939, Mead and Whittenberger 1953): the so called functional or dynamic compliance.

Components contributing to thoracic elasticity include the parallel elasticity of the rib cage and abdomen–diaphragm and the series elasticity of chest wall and lung tissue (Agostoni and Rahn 1960, Agostoni and Mead 1964). The classical sigmoidal pressure–volume (compliance) curve of the total respiratory apparatus (Romanoff 1910–1911, Rohrer 1916, Rahn *et al.* 1946) results from summation of the chest wall and lung tissue curves (Figure 3.1).

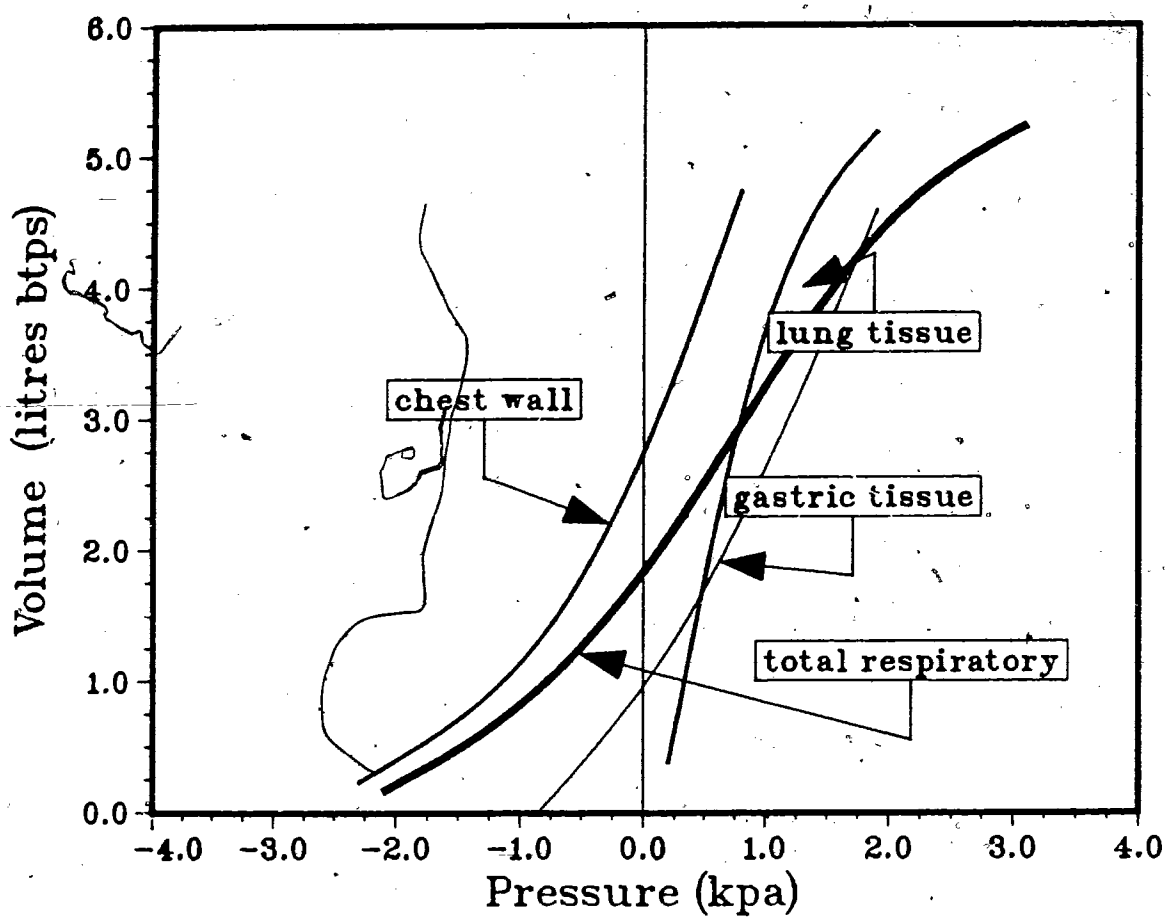


Figure 3.1: Static relaxation pressure–volume curves of total respiratory apparatus and its components. [Redrawn from Agostoni and Rahn 1960 – data for one subject from dual measurement at each of eight lung volumes.]

Lung tissue elasticity is assessed from transpulmonary pressure change ($P_{tp} = P_{alv} - P_{pl}$). Pleural pressure ($P_{pl}$) was first measured by Ludwig (1847) using a balloon-tipped intrapleural canula. Luciani (1878) approximated P_{pl} using an oesophageal catheter (P_{oes}), a technique which, through development by Buytendijk (1949), has become the criterion experimental measure of P_{pl}.

Chest wall elasticity is evaluated from transthoracic pressure change ($P_{tth} = P_{pl} - P_{bs}$). Since lung tissue and the chest wall act in series, it can be shown that transrespiratory pressure (P_{trs}) is obtainable from the difference between P_{alv} and body surface pressure (P_{bs}):

$$P_{tp} = P_{alv} - P_{pl} \qquad \textbf{Equation 1}$$

$$P_{tth} = P_{pl} - P_{bs} \qquad \textbf{Equation 2}$$

If: $$P_{trs} = P_{tp} + P_{tth} \qquad \textbf{Equation 3}$$

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

Thus: $$P_{trs} = P_{alv} - P_{bs} \qquad \textbf{Equation 4}$$

These relationships are illustrated in Figure 3.2.

Each of the pressures obtained is a function of lung volume (Carson 1820, Hutchinson 1849–52, Donders 1853, Rohrer 1916, Rahn *et al.* 1946). Hutchinson described transrespiratory pressure to be a linear function of lung volume, though subsequent studies have shown the relation to be sigmoidal (Romanoff 1910–1911, Rohrer 1916, Rahn *et al.* 1946, Lawton and King 1949). In 1913 Cloetta found pressure–volume linearity over physiological volume ranges.

Curve shapes are attributable to mechanical changes associated with volume change. King and Lawson (1950) proposed these changes to be analogous to those displayed by other extensible materials, when stretched. However, physiological mechanisms must also play a role.

At low lung volumes alveoli undergo serial collapse (Velasquez and Farhi 1964, Labadie and Van Eenige 1969), which along with the chest wall stiffness, produces the lower curvature of the total respiratory curve. A similar bend has been observed in the lung tissue compliance curve (P_{tp}) at volumes

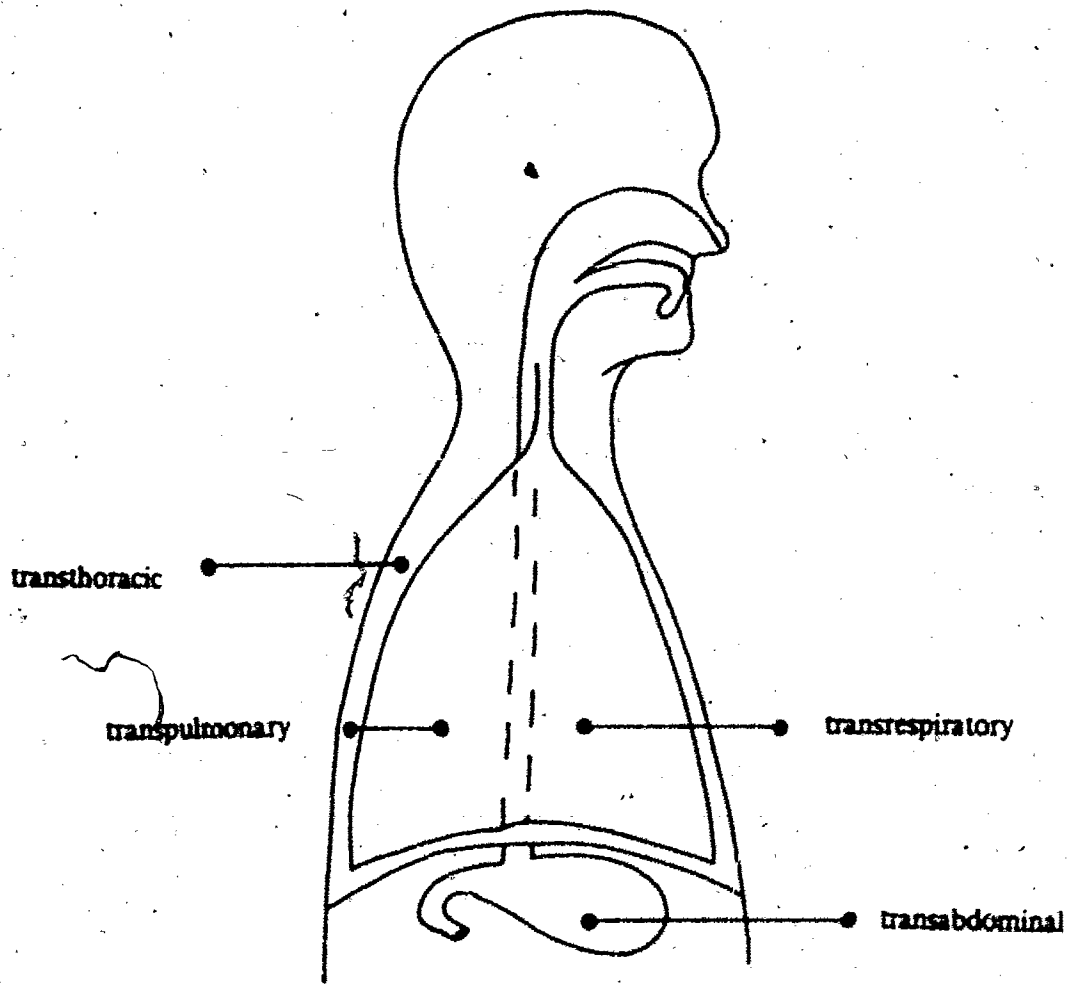


Figure 3.2: Schematic of interrelationships between static respiratory pressure measurements. [Terms are defined in Chapter One.]

<20% VC. Originally, this curve was considered to be artifactual (Knowles *et al.* 1960, Permutt and Martin 1960), however, it has been hypothesised that air trapping may produce such a curve inflection. Trapping would produce zones where local P_{pl} exceeds P_{alv} measured at the mouth, during lung collapse (Butler and Smith 1957, Glaister *et al.* 1973, Ingram *et al.* 1974, Pedersen *et al.* 1984). Gibson and Pride (1976), Milic-Emili (1984) and Pedersen *et al.* (1984) maintain P_{tp} can only reflect lung elasticity if free communication is present between alveoli and the mouth. This would invalidate the use of negative P_{tp} during calculations of lung compliance and elastic work.

At mid-lung volume range of the P_{tp}, P_{trs}, P_{tth} curves approximate linearity. However, since the overall curve shape is non-linear, elastance [1] is uniquely related to lung volume. At any point along these curves, elastance is the mean mechanical status of the series and parallel tissue elements (Labadie and Van Eenige 1969), each at a volume point on its own unique compliance curve. As lung volume approaches TLC more elements achieve their maximal expansion, producing the upper asymptotes of the total respiratory and lung tissue curves (Figure 3.1).

Recoil properties of lungs are primarily governed by surfactant surface tension (Neergard 1929, Radford 1964), while rib cage and abdominal rigidity regulate chest wall elasticity. Other contributing factors are described in Table 3.1.

Evaluating lung and chest wall elasticity requires measurement of P_{pl}. It has been demonstrated that P_{oes} provides an index of P_{pl} [2] (Fry *et al.* 1952, Stead *et al.* 1952, Cherniack *et al.* 1955, Attinger *et al.* 1956a, Butler *et al.* 1957, Ehrner 1960, Milic-Emili 1984), however, P_{oes} is more positive, and its dynamic respiratory changes (crest to trough) are smaller than P_{pl}. Elasticity measurements based on P_{oes} are valid only if P_{oes} faithfully reflects pleural surface pressure.

The oesophagus is a flaccid musculomembranous tube (25cm long, 2cm diameter), connected to the mediastinum by areolar tissue and muscle, and possessing superior and inferior sphincters (Gray 1977). Oesophageal pressures have two origins: intrinsic (local spasms, peristaltic waves), and extrinsic (intrathoracic pressures: P_{pl}, cardiogenic, tracheal movement). One must assume ones technique permits

---

[1] A measure of resistance to distension (elastance = 1/compliance). Elastance is used here instead of compliance, since the former is more readily associated with elastic work, which describes work necessary to expand lung tissue.

[2] There are two P_{pl}s: pleural surface pressure and pleural liquid pressure. The latter is more negative, while the former affects mechanical status of thoracic structures. P_{oes} approximates pleural surface pressure.

Table 3.1: **Factors contributing to chest wall and lung tissue static elasticity.**

CLASSIFICATION	COMPONENTS	CONTRIBUTION
Tissue network	Tissue cells (epithelium, fibroblasts, hystocytes)	Minor in adults
	Smooth muscle tonus (bronchial, parenchyma, interlobular septa)	Minor unless extensive
	Tissue fibres (elastic, collagen, reticulin)	Significant
	Visceral pleura	Major in disease
Surface tension	Surfactant	Major
Pulmonary blood	Acute engorgement	Unresolved
	Chronic engorgement	Significant
Lung volume	Interaction of above components with chest wall	Major

Table constructed from Radford (1964) Pp.437-445, and Cotes (1979) Pp.69-71.

distinction between P_{oes} of different origins, and that P_{oes} is not altered by measurement interventions (*i.e.* reactive error).

When the body is upright in air at normal ambient pressures, elastance has been shown to range between 0.25–1.11kPa.l^{-1} (total system), 0.26–1.00kPa.l^{-1} (lung tissue) and 0.28–1.00kPa.l^{-1} (chest wall) [3] (Christie and McIntosh 1934, Stead *et al.* 1952, Butler *et al.* 1957, Marshall 1957, Donleben 1959, Ehrner 1960, LeRoy and Guerrant 1965, Agostoni *et al.* 1966, Turner *et al.* 1968, Jonson 1970, Cotes 1979). During upright immersion Jarrett (1965) observed rightward, but parallel movement of the total respiratory compliance curve, producing greater elastance at the immersed relaxation volume (V_R), but not at the upright control V_R. Subsequent investigations have revealed a qualitatively similar shifting of the total respiratory compliance curve with immersion (Hong *et al.* 1969, McKenna *et al.* 1973, Flynn *et al.* 1975, Minh *et al.* 1979, Chapter Two).

Bondurant *et al.* (1957) reported reduced lung tissue, dynamic compliance[4] following g-suit inflation, and immersion. They attributed these changes to pulmonary vascular engorgement. Subsequent replication of this study revealed the compliance change was mediated by mechanical compression of the oesophagus, rather than altered pulmonary mechanics (Bondurant *et al.* 1960b). Basch (1887) first measured the effects of pulmonary congestion on lung mechanics, demonstrating a reduced V_T during fixed pressure ventilation, and an elevated end-expiratory volume. Subsequently, Frank (1959), Caro *et al.* (1960), Gianelli *et al.* (1967), Wood *et al.* (1971) and Prefaut *et al.* (1976) supported this apparent stiffening, however, they reported pulmonary compliance to be volume-dependent. At low lung volumes, lung recoil was diminished, while at high volumes recoil was increased. Frank (1959) found no change in compliance with engorgement when lung volumes approximated control FRC [5]. Prefaut *et al.* (1976) observed that changes in static P_{tp} during immersion were dependent upon the lung volume at which the measurements were taken. At high lung volumes, immersion increased transpulmonary recoil pressure, while the opposite was observed at low lung volumes.

---

[3] These range values represent a compilation from several sources. The normal interrelationship of the three elastances is not apparent, since the sources cited each reported the elastance of only one or two components, and not all three.

[4] Elevated elastance.

[5] Frank (1959) recalculated data of Basch (1887), finding that his observations were made at volumes below FRC. See also Borst *et al.* (1957).

Sterk (1970) found immersion of fully equipped divers reduced dynamic lung tissue compliance. From calculations by the current author, on the single compliance curve provided, it was found that compliance decreased 13%. The dynamic compliance may have been reduced due to the mechanical characteristics of the breathing apparatus. In 1973 Sterk reported a 23% reduction in quasi-static compliance during similar immersions (*i.e.* reduced compliance), though individual and statistical data were not reported. One cannot totally attribute these changes to immersion alone, since restrictive diving equipment would alter pulmonary mechanics by restricting lung volume excursion (Caro *et al.* 1960, Manco and Hyatt 1975).

Burki (1976) found $C_{q-s(l)}$ was not significantly altered immersion. Subsequent trials were performed with venous occlusion cuffs around the thighs to prevent large increments in pulmonary congestion, yet subjects still experienced reduced pulmonary compliance ($p>0.05$). If pulmonary congestion had mediated compliance changes, one would anticipate a return towards control, and not similar reductions.

Dahlbäck *et al.* (1978) reported immersion to decrease mid-lung volume, pulmonary quasi-static compliance 30% in five subjects. However, some reservations are held by the present author. First, control data were based on 18 observations, of which 39% were outside the upper limit for normal lung compliance[4], which may imply poor reproducibility. Second, compliance was obtained from visually fitted tangents, drawn along the pressure-volume curves. This introduces error[5]. Third, the magnitude of compliance change is primarily attributable to a 273% rise in one subject, whose control data exceeded normal limits on every trial. Recalculating from tabulated data, and excluding abnormal controls and matched experimental data, produced a 9% compliance fall with immersion, as opposed to the original 30% reported ($p<0.05$, t test for correlated samples). Finally, compliance was computed using slope portions which contained P_{tp} data on either side of the zero pressure intercept[6], at least in the example

---

[4] One might speculate age related elastance changes produced abnormal data (Mittman *et al.* 1965), as subjects are readily identified. However, abnormal data occur inconsistently in all but one subject.

[5] For example, assume a P_{tp} at control V_R of 0.6kPa, and at V_R plus 1 litre of 1.1kPa. Compliance = 2.0 $l.kPa^{-1}$. If the experimenter judged the latter pressure to be 1.2kPa, the compliance would be reduced to 1.67 $l.kPa^{-1}$, representing a 16.5% error.

[6] Dahlbäck employed quasi-static manoeuvres in which P_{tp} was usually negative, except below 20% VC. This is different from static P_{tp} technique, which usually produces positive P_{tp}.

provided[9]. Positive P_{tp} was interpreted as an index of air trapping. Such points occur when P_{oes} turns positive and exceeds mouth pressure (P_{ao}) (since the quasi-static manoeuvre is expiratory, and P_{ao} is always positive), and have been considered artifactual (Knowles *et al.* 1960, Agostoni and Mead 1964, Milic-Emili *et al.* 1964a, Sutherland *et al.* 1968, Green *et al.* 1974), or indices of air trapping (Butler and Smith 1957, Glaister *et al.* 1973, Ingram *et al.* 1974, Dahlbäck 1978, Milic-Emili 1984, Pedersen *et al.* 1984, Bake[10]). With alveolar closure, a greater change in P_{tp} is needed to produce a given volume change (*i.e.* compliance is reduced) (Milic-Emili 1984). Macklem[11] suggested that air trapping, if present, will be accompanied by elevated P_{pl} and local P_{alv}, as lung volume decreases. Local P_{alv} will exceed P_{ao} due to absence of free communication, producing positive P_{tp} (negative with static manoeuvre), if P_{oes} faithfully reflects P_{pl}. However, it remains unknown whether changes reflect true P_{tp} or measurement artifact, particularly during immersion. The extent to which one can rely on P_{oes} measurement at lung volumes below 20% VC is still debatable (Milic-Emili[12]). In either case data cannot be used to study intrinsic lung recoil, since recoil analysis assumes free communication between the mouth and alveoli.

Only one group has investigated chest wall elasticity during immersion (Agostoni *et al.* 1966). The authors reported only mean curves (no dispersion indices) without elastance values. P_{tth} was referenced to P_A. Submersion to the neck moved the chest wall compliance curve positively and increased its slope (reduced elastance)[13], possibly indicating a removal of shoulder girdle weight from the thorax. No studies have been located within the literature in which all three elastic components were studied.

---

[9] This subject experienced a 273% compliance decrease from control data, which themselves were outside normal upper compliance limits.

[10] Personal communication; Department of Clinical Physiology, Faculty of Medicine, University of Göteborg, Sweden.

[11] Personal communication: Professor of Medicine, Meakins-Christie Laboratories, McGill University, Montreal.

[12] Personal communication: Professor of Physiology, Meakins-Christie Laboratories, McGill University, Montreal.

[13] Slope change was derived by digitising graphic data.

3.2 LUNG VOLUMES

Borelli (1680) was the first to quantify inspired gas volumes, drawing attention to air which, even with the most forceful expiration, could not be expelled. Davy (1800) named this the 'residual volume'. Hutchinson (1846) subsequently named and quantified other volume subdivisions and capacities on over 2000 subjects of varying stature and health.

Lung volumes are determined by the interaction of several forces: (1) inspiratory/expiratory muscle tonus, (2) transrespiratory pressure ($P_{trs} = P_{alv} - P_{bs}$), (3) elasticity of chest wall and lung tissue, and (4) external forces (*e.g.* gravity, thoracic compression caused by body surface pressure increments). Immersion interacts with these forces and is universally reported to produce lung volume changes (Table 3.2).

3.2.1 Purpose of the chapter.

The objective of this investigation was to examine static respiratory mechanical properties during upright, whole-body immersion. Of particular interest was the evaluation of respiratory compliance and its components of lung tissue and chest wall compliance, and the evaluation of the elastic work of breathing during immersion with, and without hydrostatic compensation of breathing gas pressure. It was hypothesised that changes in breathing gas pressure would alter lung volume subdivisions, in particular the relaxation volume (V_R), and change the elastic portion of respiratory work attributable to tissue expansion.

Table 3.2: **Lung volume compartment dimensions, in air and following immersion, reported in literature.**

SOURCE	Condition	TLC	RV	VC	IC	FRC	ERV	IRV	V_T
Hamilton	D			4.86					
& Mayo 1944	W			4.55					
Brozek	D		1.59						
et al. 1949	W		1.57						
Hong *et al.*	D			4.50	3.42		1.09		0.71
1960 (sup)	W			4.49	3.37		1.12		0.79
Beckman	D			3.99			1.56	2.12	0.64
et al. 1961	W			3.86			0.56	2.79	0.51
Boren *et al.*	D	6.36	1.61	4.71	3.66	2.66	1.05		
1966									
Agostoni	D		1.67			3.53	1.86		
et al. 1966	W		1.39			1.90	0.56		
Craig &	D			5.05					
Dvorak 1975	W			4.60					
Flynn *et al.*	D	6.49	1.12	5.41	3.23*	3.26	2.13		
1975	W	6.30	1.04	5.26	4.25*	2.05	1.01		
Bondi *et al.*	D	6.85	1.55	5.30	3.52*	3.33	1.89		0.63
1976	W	6.19	1.40	4.78	4.23*	1.96	0.56		0.60
Dahlbäck	D			5.33			2.10		
et al. 1978	W			5.05			0.52		
Robertson	D**	6.82	1.54	5.28	3.42*	3.40	1.87	2.62	0.79
et al. 1978	W**	6.45	1.48	4.97	4.38*	2.07	0.59	3.41	0.97
Buono 1983	D	7.15	1.40	5.75					
	W	6.77	1.36	5.41					

Condition: D = dry, W = immersed, * = by subtraction, ** = plethysmographic measurement, (sup) = supine with water at sternum.

3.3 METHODS

3.3.1 Subjects

Ten male non-smokers (including 8 divers), screened by questionnaire for normal lung function history, participated as paid subjects. Medical questionnaires were screened by a physician prior to experimentation. All received information packages and signed informed consent releases. At least two investigators were in the laboratory, and a physician on call, with emergency resuscitation equipment (within three minutes), throughout all experiments. To avoid age related compliance changes, subjects participating in elastic work trials were all less than 40 years of age (Mittman *et al.* 1965).

Subjects performed a series of respiratory, static pressure-volume relaxation manoeuvres over the lung volume range from residual volume (RV) to total lung capacity (TLC). Manoeuvres were inspiratory commencing at RV. Using oesophageal balloons, transrespiratory, transpulmonary and transthoracic pressures were measured at each relaxation pause. Trials were performed upright in air, and during total immersion using positive facial counter-pressure. To facilitate reproducibility of static pressure-volume data, subjects were chosen from the best subjects used in the determination of the lung *centroid* .

Subjects also performed spirometric manoeuvres to enable measurement of vital capacity, inspiratory and expiratory reserve volumes, and inspiratory capacity.

3.3.2 Apparatus

Lung volumes were measured using a pneumotachograph (Fleisch #4) coupled with a differential pressure transducer (Validyne DP103± 0.25kPa (±2.5cmH_2O)). Alveolar pressure was measured at the mouth, with the airway occluded and glottis open, using a differential pressure transducer (S.E. 1150±6.2). Oesophageal pressure was measured using an oesophageal balloon (10cm long, wall thickness 8.45 x 10^{-3}mm) and catheter [14] (teflon: internal diameter 1.35mm, wall thickness 0.3mm) connected to a differential pressure transducer (Validyne MP45 ±3.92kPa). Oesophageal pressure was determined using P_A as reference, while transpulmonary pressure (P_{tp}) was obtained from the difference between mouth pressure and P_{oes}.

[14] Appendix Two details procedures for making oesophageal pressure probes.

According to Milic-Emili *et al.* (1964) oesophageal balloon volumes should be minimal to adequately gauge pleural pressure. To determine this volume, the pressure-volume characteristics of individual balloon/catheter systems were analysed. Catheters were attached to Tygon® tubing (internal diameter 3.18mm, wall thickness 1.59mm), which, in turn, was connected to the differential pressure transducer. Trials during immersion required longer tubing connections than control trials (2.30 and 3.78 metres respectively), due to the greater distance between the transducer and the subject in the former. Characteristics of both systems required evaluation.

Balloon/catheter systems were evacuated in air to a pressure of -2.94kPa (-30cmH_2O). Air was then added in 0.2ml increments, with system pressure being monitored at each step. Data plots identified catheter volumes at which an extended pressure-volume plateau was located. At these volumes the balloon/catheter compliance would not confound oesophageal pressure measurement (Figure 3.3). To the minimal volume was added an additional 0.1ml to prevent high positive pressures from compressing balloon volume, driving volume down to the steep part of the compliance curve. In this manner catheter volumes were kept minimal to avoid pressurising the oesophagus (Milic-Emili 1964). System volumes in air were 0.95±0.3ml, and for immersion studies were 2.40±0.2ml. The former values were slightly greater than recommended volumes (Gibson and Pride 1976, Milic-Emili 1984) of 0.5ml, and were attributed to larger probe volumes in the present study. Volumes were much larger than 0.2ml suggested by Milic-Emili *et al.* (1964). This is attributed to wall thickness differences between 'hand-dipped' and manufactured balloons (Lemen *et al.* 1974, Milic-Emili 1984). Volumes for probes used during immersion matched those of Sterk (1973), who also studied immersed subjects[15].

Output from mouth and oesophageal transducers was amplified (S.E. 423/1E amplifier demodulator). Flow signals were amplified (Daytronic LVDT, model 9130) and low pass filtered at 5Hz (Rockland model 432 dual high/low filter). All output passed via an analog/digital converter (Tecmar Labpac) to an IBM(PC) for 50Hz sampling and storage.

System linearity was evaluated[16] using a U-tube water manometer in parallel with each transducer. A step series of known pressures (±0.2kPa) was applied to each system over the range -5.9kPa to

---

[15] Milic-Emili *et al.* (1964) have shown that larger balloon volumes displace the P_{tp} curve, without altering its slope. Thus lung compliance is not influenced by balloon volume.

[16] Since it was crucial to know the system response (i.e. from transducer to computer), and since system components act in series (causing system amplitude response to be the product of component responses - Fry 1960), it was decided to evaluate the complete system, rather than linearity of separate components.

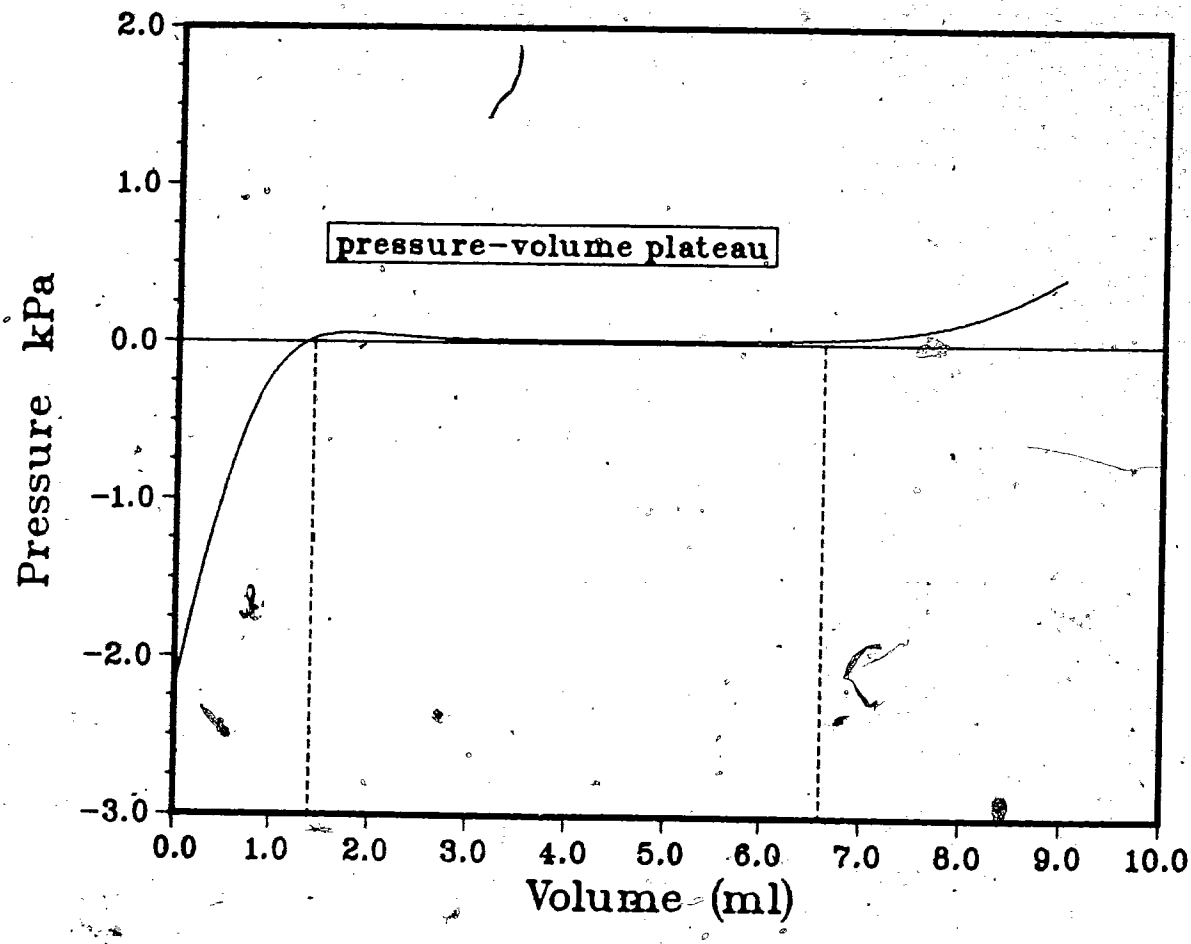


Figure 3.3: Pressure–volume characteristics of intraoesophageal pressure probes.

+4.9kPa. This range was deemed to include mouth pressures anticipated during static pressure–volume relaxation manoeuvres. Both pressure systems were linear ($r > 0.999$), and remained so throughout the investigation.

Volumes were calculated by integration of flow signals with respect to time. System validation included all components between the pneumotachograph and integration routines. Integrated volumes displayed a non–systematic variance with flow rate ($\bar{x} = 1.004 \pm 0.042$ litres) when a one litre standard was passed through the pneumotachograph at flow rates between 20–40 $l.min^{-1}$.

Accuracy of volume and pressure measurement was confirmed by repeatedly applying known volumes and pressures to each system. The standard error for dry volumes was 0.007 litres (using 0.961 litre injecter as a standard), and for immersed volumes it was 0.016 litres (using 5.921 litre syringe standard). Five pressure standards were used (−1.96, −0.98, 0, +0.98, +1.96kPa (−20, −10, 0, +10, +20cmH_2O)). Pressures were applied with an accuracy of ±0.02kPa (±0.2cmH_2O). Standard error for measuring the five pressure standards using the mouth and oesophageal systems were 0.009 and 0.018kPa respectively.

Immersion trials were performed in the wet chamber of a hypo–hyperbaric chamber complex able to simulate depths of 300m with temperature and humidity control. Water temperature was regulated to 34.8±0.5° C. This necessitated the use of long pressure probes to connect the pneumotachograph to its differential pressure transducer. To evaluate length effects on volume measurement, trials were performed using known syringe volume standards (0.961 litres) and different probe lengths (0.415, 1.600 and 3.350 metres[17]). Increasing probe length resulted in a systematic but non–significant increment in volume measurement (0.97±0.03, 0.97±0.01, and 0.98±0.01 litres respectively), representing 0.52%, 1.04% and 1.77% above syringe volume, which itself had been determined by repeated water filling.

During immersion trials the pneumotachograph was kept above the water. Since both subjects and air supply were immersed, lengths of low resistance tubing were used to link each with the pneumotachograph (Figure 2.3). At either end of this tubing was a Mares (MR 12 III) regulator connected to compressed air storage tanks. Both regulators were kept in the same horizontal plane. Subjects were positioned in this circuit, close to the second regulator. Two, two–way taps permitted the experimenter to control airflow through the pneumotachograph, and from which regulator the subject

[17] Probe lengths corresponded with: (a) shortest possible probe length, (b) probe length for dry trials, (c) probe length for immersion trials.

breathed. Subjects were shown how to operate the taps to obtain air on demand.

The pneumotachograph was heated to prevent condensation in the resistor tubes during expiratory manoeuvres. Laboratory trials using copper/constantan thermocouples (0.125mm diameter) and a cold junction compensator (Omega Engineering Inc., Connecticut) measured inspired gas temperatures of 39.3° C (SEM = 0.002) and expirate temperatures of 39.7° C (SEM = 0.35) [11]. This difference was negligible in any BTPS volume conversion, and was ignored. Gas temperatures were deemed to be 39.3° C.

During elastic work trials regulator 1 was positioned at the sternal notch depth. Due to its large dead space, this airway was used only during trials.

The second regulator served three purposes:

(1) it provided an easily accessible emergency air supply to subjects,

(2) it permitted subjects to breathe comfortably below the water between trials, increasing their ability to reproduce relaxation pressures, and

(3) it reduced apparatus dead space.

Trials performed in air used a small length of low resistance tubing (volume = 0.365 litres) to connect subjects directly to the pneumotachograph. All other tubing was disconnected but tap 1 remained. In each set-up there existed a volume of air between the subject and tap 1 (1.59 litre and 0.95 litres - immersion and control respectively). Lung volumes were corrected for gas expansion during negative pressure holds, and for compression during positive pressure holds.

Immersion spirometry and elastic work trials were performed with positive facial counter-pressure, using a modified Kirby-Morgan band mask (U.S. Divers - Figure 2.4). A small pressure line connected regulator 2 directly with the air space inside the hood and pressurised the latter to a pressure equal to the depth of the regulator. Since the pressure line by-passed tap 2, constant mask pressure was maintained during pressure-volume manoeuvres. The hood provided strong rubber strappings to maintain its position during immersion. These straps facilitated mouthpiece stability, as it was impossible to physically eject the mouthpiece when strapped in position. Subjects wore nose clips during all trials to prevent air losses.

---

[11] Response time 0.1-0.5 seconds.

During all trials a U-tube water manometer was connected in parallel with the mouth pressure transducer, enabling the experimenter to determine when a stable relaxation pressure was attained.

Preliminary immersion trials produced relaxation mouth pressures beyond the transducer capacity. To bring pressures into range a pressure compensator was connected to the transducer reference side (which had been open to air during control trials). The compensator consisted of a perspex cylinder, and pressure probe, immersed to known water depths, and connected in parallel with both the manometer and transducer. The compensator consequently applied a positive reference pressure to the transducer, enabling measurement of greater relaxation pressures.

3.3.3 Calibration

Calibration was performed daily, or more frequently if the apparatus was shut down between subjects, or when large time gaps occurred between successive testing periods. Pressure calibration was performed at zero and +4.9kPa (+50cmH_2O) using a water manometer (±0.02kPa (±0.2cmH_2O)), utilising previously confirmed system linearity.

Flow calibration was carried out with apparatus *in situ*, preceding trials and with the pneumotachograph heater off. For control trials a series of six syringe volumes (0.96 litres) were passed through the pneumotachograph at various flow rates. Integrated flow was averaged and matched against the volume standard. Immersion trial calibration could not be performed in this manner. With regulator 1 submerged it was found that calibration by pumping volumes at regulator pressure produced erroneous volumes. Finucane *et al.* (1972) demonstrated that geometry change of tubing upstream of a pneumotachograph had a marked effect upon pressure differential across the resistive element. Thus flow calibration was performed with apparatus fully assembled by employing a single inspiratory calibration stroke with a large syringe (5.92 litres). Calibration was performed with the regulator 10-15cm below the surface. Testing revealed no effect of regulator depth on calibration data, and calibration data obtained with the apparatus immersed now agreed with that obtained during control calibrations.

3.3.4 Procedures: elastic work study

Subjects performed inspiratory, static pressure-volume manoeuvres over the volume range from RV to TLC in a replication of the protocol used in Chapter Two (Section 2.1.4), thereby providing a means of re-evaluating the upright lung *centroid*. Subjects were retrained for 15-20 minutes to again

provide reproducible data[19]. One subject was new and undertook identical training to that described in Chapter two. During immersion trials both regulators were positioned at the depth of the sternal notch.

Trials were preceded by 3–5 rapid breaths, at slightly larger than normal tidal volumes, to lower blood CO_2 tension and reduce respiratory drive. Three normal tidal volume breaths followed this tachypnea to remove volume history influences on compliance (Ferris and Pollard 1960). Expiration to RV followed. During immersion, transition to RV was determined by rapid air expulsion from regulator 2, and RV attainment by cessation of bubbles from this regulator. At RV subjects were isolated from atmospheric air by closing tap 1 (control) or from regulator 2 by closing tap 2 (immersion). On verbal command (head taps during immersion) subjects gently inspired when tap 1 was opened, and relaxed when it closed. To prevent large uncontrolled inspirations subjects were trained to resist inflow at low relaxation volumes. Inspiration was continued until subjects sensed tap 1 close, at which point they again relaxed. To prevent accidental gas loss, tap 1 was not opened till the experimenter observed commencement of inspiratory effort (change in manometer pressure). Using this sequence subjects were able to provide 5–10 different points at which the pressure–volume relation between RV and TLC could be evaluated. Trials were repeated 5–7 times. Unsatisfactory trials were repeated. An electronic trigger was activiated at each relaxation point and recorded on a separate computer channel to identify points of relaxation.

Subjects performed trials seated upright with their back vertical and hips flexed at ~90°. To reduce buoyancy during the immersion trials, subjects placed a weight belt over the thighs, and the experimenter held the head position constant.

Oesophageal balloons were inserted by the experimenter in the presence of a physician and/or a member of the supervisory committee trained in placement of oesophageal catheters. Nasal and pharyngeal mucosa were locally anaesthetised (xylocaine endotracheal aerosol[20]). Insertion procedures were demonstrated to the subjects and fully explained while the topical anaesthetic took effect. Subjects were seated in a padded chair with arm and back support. Balloons were smeared with a sterile, water–soluble lubricant (K–Y Jelly©, Johnson & Johnson). The more patent airway was determined by

---

[19] Five subjects had previously taken part in six groups of static pressure–volume trials, two were involved in five groups of trials, and two took part in four groups of trials.

[20] Subjects were questioned regarding allergy to local anaesthetic prior to use. All subjects responded negatively.

examination and the balloons were inserted via the nares, with the subject's head fully extended. The probe tip was located in the pharynx behind the uvula using a tongue depressor and flash light. At this point the balloon was withdrawn ~1cm, for subject comfort, and the subject sat erect and commenced strong drinking and swallowing of water through a straw. Further catheter insertion was timed to coincide with swallowing.

Preliminary catheter insertions were used to identify balloon positioning in the oesophagus. Balloons were located below the gastro-oesophageal junction (indicated by positive pressure with inspiration), then withdrawn slowly to identify the cardiac sphincter (positive pressure turning negative on inspiration). It was intended to locate balloons ~10cm above the gastro-oesophageal junction, however, preliminary work revealed considerable intrasubject variability in depth between repeated positionings. Ehrner (1960) similarly reported large variability (±5cm) for repeat placements. It was decided to reference balloon depth to the nares. All probes were inserted 45cm, then withdrawn and secured at 40cm from the nares to the balloon tip.

Air within oesophageal balloons collects at the point of least pressure (*i.e.* the upper junction of balloon and catheter), thus oesophageal pressure (P_{oes}) was measured approximately 30cm from the nares. Milic-Emili *et al.* (1964) had shown the pressure-volume characteristics of this position to be very similar to those obtained from the lower one third of the oesophagus. Gibson and Pride (1976) found mid-oesophageal positioning superior. Furthermore, the influence of basal alveoli collapse, during immersion (Pedersen *et al.* 1984), upon lower third P_{oes} was unknown.

During preliminary testing, P_{tp} was monitored during spontaneous breathing to evaluate cardiogenic influences. Christie and McIntosh (1934) reported P_{pl} fluctuations of 0.83kPa (8.5cmH_2O), Lemen *et al.* (1974) observed a 0.49kPa variance, and Gibson and Pride (1976) found ~0.89kPa change in P_{tp}, each due to cardiac oesophageal compression. Maximum peak to minimum trough P_{tp} differences in the current preliminary work were 0.145kPa (SEM = 0.033[21], sampled at 50Hz). Fluctuations did not show recognisable cardiogenic patterns, and were interpreted as being caused by oesophageal contraction or movement artifact. The pressure difference magnitude did not warrant recording at P wave onset to signal P_{tp} measurement times (Trop *et al.* 1970), since P_{tp} measures were to be averaged over 50 data points (*i.e.* one second collection) which would include at least one full cardiac cycle.

[21] $\bar{x}$ = mean, SEM = standard error of the mean.

It was concluded that P_{oes} should be measured in the lower portion of the middle third of the oesophagus (*i.e.* 30cm from nares). In this position end tidal P_{oes} averaged −0.42kPa (SEM = 0.04) before commencing experiments, and −0.41kPa (SEM = 0.05), before the catheter was removed at the end of the experiment. These data, in agreement with measurements of Ehrner and Nisell (1959), provided confidence that P_{oes} measurement was valid, and that probes had not moved significantly during the experiments.

After the oesophageal catheters were positioned, subjects performed a valsalva manoeuvre to evacuate air from the balloons. Each balloon was injected with 5ml of air to ensure even filling. A volume was then removed, to leave the predeterimined balloon/catheter volumes. Repeat valsalva manoeuvres enabled balloon volume confirmation between trials.

Trials did not commence until 20–30 minutes after the catheter was positioned, to ensure subsidence of peristaltic contraction. Balloons and catheters were washed, sterilised (Cidex® activated 2% glutaraldehyde solution), and refrigerated between trials, after dusting with talcum.

3.3.5 Procedures: static lung volume measurement.

Spirometric measurement of lung volumes[22] was performed using the apparatus and subject positioning described above (Figure 2.3). Measurements were made while subjects breathed through the heated pneumotachograph. Between submerged trials subjects breathed through regulator 2 to avoid CO_2 accumulation.

The depth of both demand regulators was identical for each submersion trial, and trials were randomly performed at four covert regulator depths: mouth level, 13.5cm below sternal notch (mean lung *centroid* position[23]), and 10cm above and below lung *centroid* . Before commencing each trial subjects breathed at the new delivery pressure for about one minute.

Tidal volumes (V_T) were recorded following several shallow, rapid breaths. Tachypnea was designed to reduce respiratory drive during trials; since apparatus dead space was large, rebreathing would progressively elevate V_T. Tachypnea was followed by several normal breaths. Subjects were asked to select and maintain a normal V_T and breathing frequency (f_b). Redirection of inspiratory

[22] Vital capacity, inspiratory and expiratory reserve volumes, inspiratory capacity and tidal volume.

[23] As determined from Chapter two.

airflow through the pneumotachograph was done covertly.

Vital capacities (VC) and inspiratory capacities (IC) were measured as averages from repeated trials rather than from peak values (Beers 1957). Both the VC and IC measurements were recorded from inspiratory volumes, the latter being performed following a normal V_T expiration. Inspiratory and expiratory reserve volumes (IRV, ERV) were determined by subtraction.

3.3.6 Calculations.

Elastic work study.

Pressure measurements represented differential pressures between P_{alv} and sternal notch pressure (transrespiratory pressure*[24]), and between P_{alv} and P_{oes} (transpulmonary pressure). Transrespiratory pressures (P_{trs}) were obtained using Equation 5:

$$\text{Pressure relative to sternum} = \text{Pmeasured} + (\text{Pcomp} - \text{Psternal}) \qquad \textbf{Equation 5}$$

where:

Psternal = pressure at sternal notch.

Transpulmonary pressures (P_{tp}) were derived by subtraction at the pressure transducer. Transthoracic pressures (P_{tth}) were calculated from isovolume subtraction (*i.e.* $P_{trs}^{*} - P_{tp} = P_{tth}$) during computer analysis.

Analysis programmes were written to perform calculations. Using electronic trigger points as relaxation pause markers, the programme integrated flow between consecutive markers, and determined transrespiratory, transpulmonary and transthoracic pressures over a one second period (50 samples), two seconds after each marker point.

Inspired volumes were corrected for absolute ambient pressures (*i.e.* atmospheric, pressure at the regulator, alveolar pressure and pressure at the anatomical reference point), and for apparatus dead space effects using the Boyle–Mariotte Law.

$$P_1V_1 = \text{constant} \qquad \textbf{Equation 6}$$

[24] Transrespiratory pressure is generally computed at P_{alv} minus mean body surface pressure. The latter, at this point, was unknown. This correction was made subsequent to measuring P_{LC}.

If V_1 represents integrated volume, and V_2 represents actual volume inspired.

Then:

$$V_2 = P_1 \cdot V_1 / P_2 \qquad \textbf{Equation 7}$$

where:

P_1 =atmospheric + regulator pressure,

P_2 = atmospheric + Pmouth.

Alveolar air is compressed and decompressed by changes in ambient and alveolar pressure. Apparatus dead space (V_D) stores compressed gas and returns gas to lungs during decompression, thus V_R must be considered a component of V_2. RV is also a V_R component. Substitution of terms into equation 7 produces:

$$V_2 = \frac{(P_A + Preg)(V_1 + RV + V_D)}{(P_A + Pmouth)} - RV - V_D \qquad \textbf{Equation 8}$$

Integrated volumes were then corrected to BTPS (equation 9). Pressure corrections included allowance for absolute pressure at the anatomical reference site, and alveolar pressure during static holds. This was essential since immersion compressed gas volumes, and alveolar pressure acted either to decompress (if negative) or further compress (if positive) lung volumes measured by integration of inspiratory flow.

$$V_3\ BTPS = \frac{(P_A + Pmouth + Psternal - P_{H_2O})\ V_2 \cdot 310}{(P_A + Pmouth - P_{H_2O})(272 + T_A)} \qquad \textbf{Equation 9}$$

Before analysis of compliance curves was undertaken, data points were rejected as errors according to the following criteria: (a) false positive/negative pressures (*e.g.* negative pressure achieved at lung volumes greater than the V_R), (b) points when glottal closure or swallowing occurred during trials, as recorded by the experimenter or detected from the compliance curve shape (*e.g.* pressure fluctuations

>0.49kPa (5cmH_2O) from mean curve), and (c) points of poor relaxation (as determined by curve analysis). Care was taken to avoid the unnecessary exclusion of pressure–volume data points (Henry 1949).

Compliance curves were examined using polynomial regression analysis (Dixon 1983) of the general form:

$$y = \beta_0 + \beta_1 x + \beta_2 x^2 + ... + \beta_j x^j + e \qquad \textbf{Equation 10}$$

Conventionally, static pressure–volume curves are plotted with pressure on the abcissa. Statistical convention dictates independent variables appear on the abcissa. Static pressure is dependent upon lung volume, thus analysis was performed with volume as the independent and pressure as the dependent variable.

Curves for the total respiratory, lung tissue and chest wall were separately analysed using orthogonal polynomials to derive the least squares curve fit of independent to dependent variables. Control and immersion data were analysed independently. Coefficients of each equation were entered into three analysis programmes written for this investigation. The first enabled numeric solution of the regression equations to derive lung volume at zero P_{trs} pressure[25] (*i.e.* V_R). The second differentiated the equations producing respiratory, lung tissue and chest wall elastance[26], compliance, and static volume and pressure at increments of 1% of the vital capacity (VC), over the volume range residual volume (RV) to total lung capacity (TLC). The third programme integrated pressure with respect to volume, providing respiratory, lung tissue and chest wall elastic work[27] over a given volume.

Lung *centroid* was determined from positive pressure, isovolume (V_R) displacement of the total respiratory pressure–volume curve. Pressure data were differential pressures between P_{alv} and pressure at the sternal notch. If pressure is expressed in imperial units (cmH_2O), one may represent lung *centroid*

[25] Transcendental equations cannot be resolved analytically since x cannot be isolated. A value of x (volume) may only be attained numerically. Solution accuracy was set at ±1ml, which was beyond the capacity of the measurement system, but was deemed appropriate (Kuo 1972).

[26] Differentiation provides a measure of curve slope (*i.e.* dy/dx or dP/dV) which is respiratory elastance. When drawn according to physiological convention (pressure on abcissa), curve slope equals respiratory compliance. Compliance = 1/elastance.

[27] Area below the pressure–volume curve has the dimension of work (Chapter One).

as a vertical displacement below the sternal notch, since distances equate with the pressure head produced by an equivalent column of water. In accordance with Jarrett (1965), curve shifts were expressed relative to the upright control V_R. Use of mid-curve isovolumes ensured curve shift analysis over an approximately linear part of the curve. Curve shifts over this range were expected to be parallel, and therefore to provide a P_{LC} relevant to a greater variety of lung volumes.

Using the P_{LC} for each subject separately, the total respiratory compliance curves from immersion trials were moved leftwards (negatively) so that the V_R matched the V_R measured in air. Using these pressure-volume coordinates, the P_{trs} and P_{tth} static pressure-volume data were reanalysed. P_{alv} and P_{oes} were now expressed relative to the body surface pressure at the external point in the same horizontal plane as the individual lung *centroids*. Pressure-volume characteristics were deemed to represent mechanical changes accompanying immersion with gas delivery at P_{LC}.

P_{trs}, P_{tth} and P_{tp} curves enabled computation of elastic work components (Figures 3.4A,B,C):

(1) total respiratory elastic work,

(2) lung tissue elastic work,

(3) chest wall elastic work,

(4) hydrostatic work performed by the regulator during immersion[21], and

(5) work performed by the respiratory muscles.

Analysis was performed over a 1 litre V_T since control spontaneous breathing produced a V_T of 1.12 litres (SEM = 0.10, n = 10). Analyses started at the control V_R. Additional analysis commenced at volumes corresponding with the regulator delivery pressures equivalent to the pressure at mouth depth, and at depths 10cm above and below the lung *centroid locus*. Since the results from the previous chapter had shown that immersion caused parallel shifts of the respiratory compliance curve, this method of analysis was appropriate.

Of prime concern in this investigation was the work performed by the inspiratory muscles to inflate the elastic respiratory system. To quantify this work one requires a knowledge of total respiratory elastic work, and the hydrostatic work performed on the respiratory system by the diving regulator (with its positive or negative pressure bias relative to P_{LC}). During respiratory volume inflation, inspiratory muscle work is given by the following general relationship:

[21] This work work is provided by the potential energy difference between the regulator pressure and P_{LC}.

Inspiratory muscle work = (total respiratory elastic work) – (regulator work) **Equation 11**

All work subdivisions were measured relative to the pressure at the control respiratory V_R.

In air, and during immersion with air delivery at P_{LC}, the regulator performs no respiratory work and inspiratory muscle work equals total respiratory elastic work, the latter being obtained by integration of the area under the P_{trs} curve (*i.e.* area V_1SV_2, Figure 3.4A[29]). Chest wall elastic work equals area V_1RCV_2, while lung tissue elastic work corresponds to area V_1LUV_2.

When gas delivery, during immersion, is at pressures less than P_{LC} (*e.g.* mouth pressure and pressure at the depth 10 cm above lung *centroid*), total respiratory work corresponds with the area V_1TSV_2 (Figure 3.4B). Hydrostatic work done on the respiratory system by the regulator has a negative sign, and equals area V_1TQV_2[30]. Inspiratory muscle work was obtained by subtraction of the regulator work from the total respiratory elastic work (area TSQ). Chest wall and lung tissue elastic work were obtained from areas below the respective curves (V_1RCV_2 and V_1LUV_2).

Delivery at gas pressures greater than P_{LC} (*e.g.* at depth 10cm below lung *centroid*) produces positive work by the regulator on the respiratory system (area V_1TQV_2). Inspiratory muscle work equals area TSQ (Figure 3.4C). Total respiratory, chest wall and lung tissue elastic work were again derived by integration (areas V_1TSV_2, V_1RCV_2 and V_1LUV_2 respectively).

Static volume measurement.

Spirometric volumes were determined by computer integration of flow with respect to time, from points of zero flow. Volumes were converted to BTPS units (equation 9).

3.3.7 Analysis.

Analysis was based on a repeated measures experimental design, with five levels of one within-subjects factor (*i.e.* air delivery pressure). Subjects provided data in all cells. *A priori* significance

[29] Zero volume corresponds with RV.

[30] The respiratory system must perform positive work to overcome the hydrostatic pressure difference between P_{LC} and regulator pressure.

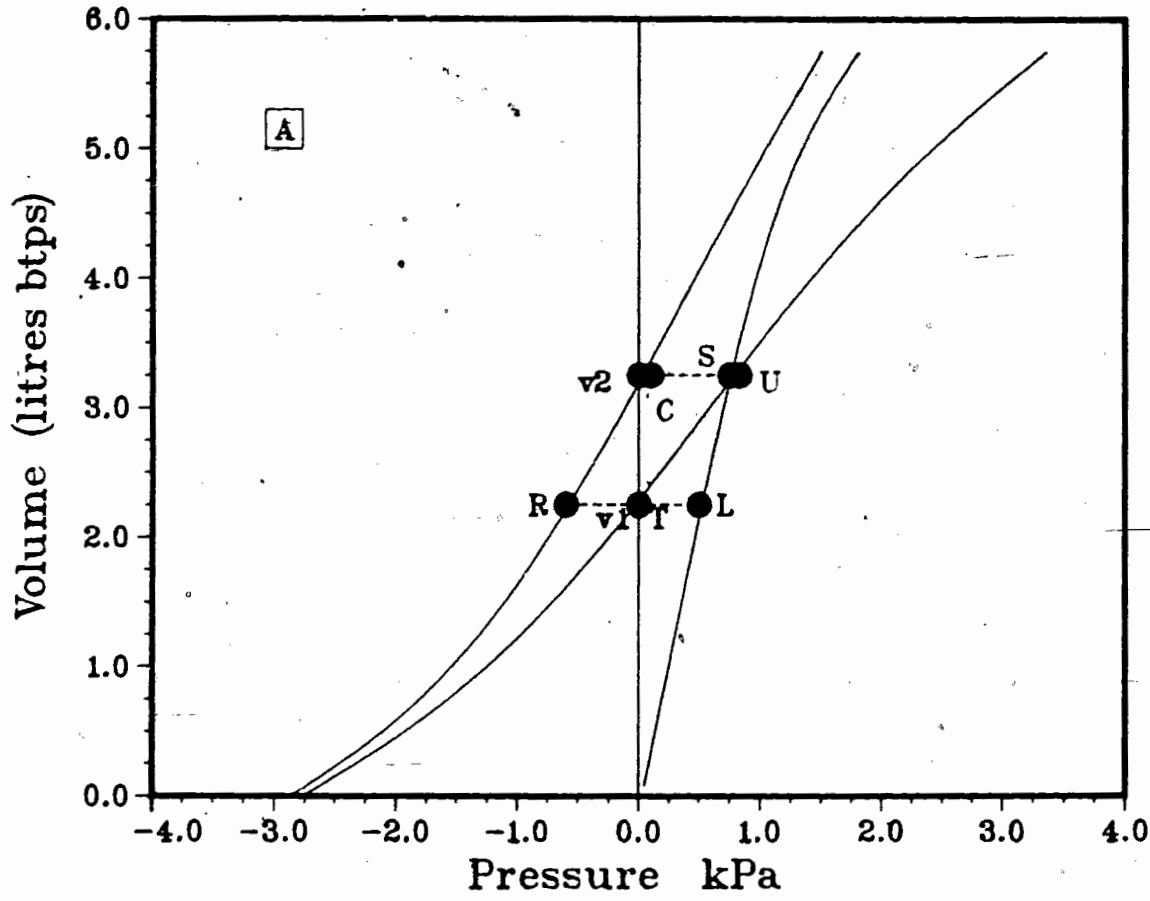


Figure 3.4A: **Partitioning elastic work of the respiratory system in air, and during immersion, breathing air at lung *centroid* pressure.**

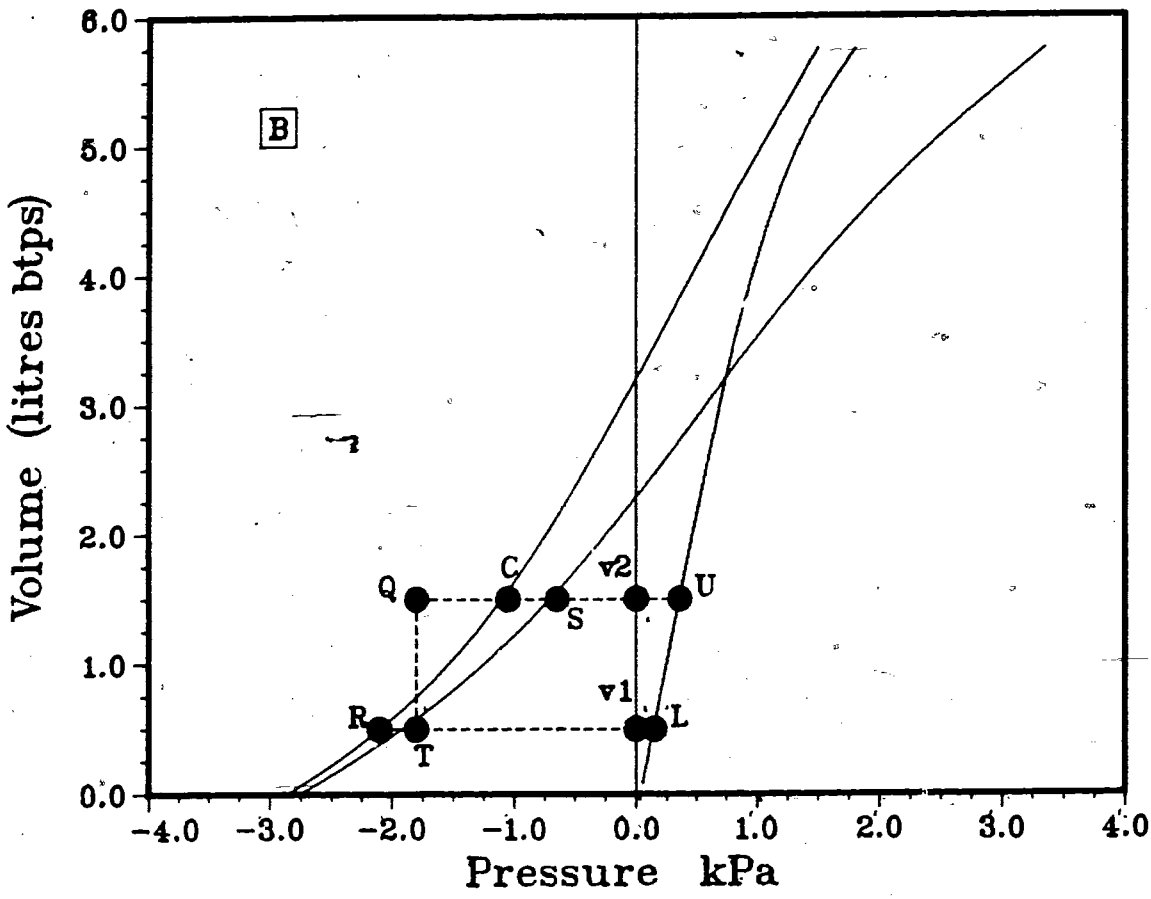


Figure 3.4B: Partitioning elastic work of the respiratory system in air, and during immersion, breathing air at pressures below lung *centroid* pressure.

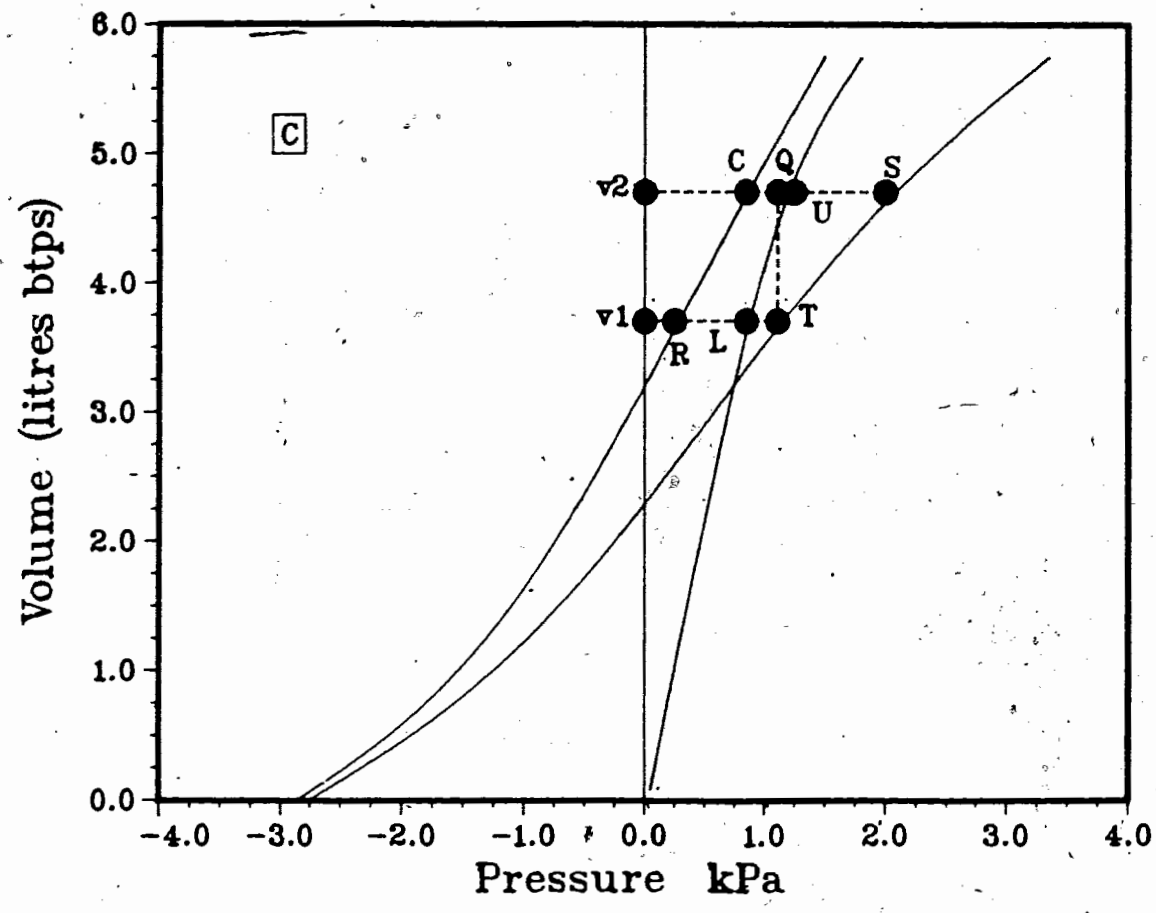


Figure 3.4C: **Partitioning elastic work of the respiratory system in air, and during immersion, breathing air at pressures greater than lung *centroid* pressure.**

was set at the 0.05 level. Statistical analyses took the form of: (1) correlated t-tests; (2) Hotelling's T-squared analysis for simultaneous comparisons between univariate factors for several variables; and (3) multivariate analysis of variance for repeated measures analysis over several levels of the treatment effect. Following significant *F* statistics, *post hoc* multiple comparisons (Tukey's HSD[31]) were used to isolate sources of significant variance (Keppel 1973, Huck *et al.* 1974). Where significant changes were not observed, computations of statistical power were performed. Statistical power (ø) is defined as the probability of correctly accepting an experimental hypothesis (Keppel 1973, Gehring 1978).

[31] Summary tables are contained within Appendix Three.

3.4 RESULTS

3.4.1 Characteristics of subjects.

Physical characteristics of subjects are detailed in Table 3.3. Subjects took part in both elastic work and lung volume experiments [32].

3.4.2 Elastic work partitions in air.

Pressure–volume curves were constructed from an average of 45 (±6, total respiratory system), 38 (±9, lung tissue) and 37 data points (±8, chest wall), over the volume range from RV to TLC. Typical curves are illustrated in Figure 3.5, with data points illustrating reproducibility. Control compliances were within expected normal limits (Table 3.4), with respective means at 1.53 (SEM = 0.14[33]), 3.24 (SEM = 0.21) and 3.32 $l.kPa^{-1}$ (SEM = 0.51). It was concluded that control data were reproducible and valid assessments of respiratory, lung tissue and chest wall compliance and elastic properties.

Elastic work[34] of the total respiratory system averaged 0.36kPa (SEM = 0.05) (Table 3.5), in agreement with Chapter Two and Rahn *et al.* (1946)[35]. Energy to perform this work was provided by the inspiratory muscles. Lung tissue elastic work averaged 0.64kPa (SEM = 0.04). That lung tissue exceeded total respiratory elastic work is explained by the outward chest recoil (work = −0.29kPa, SEM = 0.06), aiding lung expansion due to its tendency to move towards its relaxation volume at ~61% of VC (SEM = 5.75). [Fenn (1951) and Agostoni and Mead (1964) reported chest wall relaxation at ~55% VC.]

---

[32] Subject numbers allocated in Table 3.3 remain consistent throughout the chapter, but differ from those in Table 2.2.

[33] $\bar{x}$ = mean, SEM = standard error of the mean, ø = statistical power (probability of correctly accepting experimental hypotheses (Keppel 1973, Gehring 1978)).

[34] Elastic work data are reported over V_T of 1 litre, producing work per litre. Units are $J.l^{-1}$, which is equivalent to kPa.
Proof: Work = force · displacement; work per litre = force · displacement/volume = $J.l^{-1}$. Cancellation produces: force/area. But pressure = force/area, therefore correct units for elastic work over 1 litre V_T are pressure units, or kPa.

[35] Both studies expressed elastic work for a 0.5 litre V_T; doubling V_T elevates elastic work about fourfold (Figure 2.14).

Table 3.3: **Physical characteristics of subjects.**

SUBJECT	Age (yr)	Height (cm)	Stem height (cm)	Mass (kg)	Vital capacity (l.BTPS)	Chest circum. (cm)	Sternal length (cm)	A-P chest (cm)	Diver
1	26	179.4	96.7	72.2	6.21	91.5	21.5	17.7	Y
2	24	189.7	99.7	84.4	5.99	97.0	24.3	20.4	Y
3	31	181.8	96.6	73.8	5.39	92.0	19.8	18.2	Y
4	26	177.7	97.1	73.9	5.25	90.0	21.0	21.7	Y
5	22	185.3	99.4	74.8	6.55	98.9	22.3	19.1	Y
6	28	184.5	92.6	89.0	6.18	102.5	23.4	24.1	Y
7	24	176.0	96.2	78.6	5.89	95.0	23.0	22.3	N
8	23	181.8	95.2	82.5	5.77	101.1	18.1	21.4	Y
9	29	169.7	91.9	63.6	4.58	91.8	19.2	17.4	N
10	32	184.5	96.1	77.4	6.78	101.8	23.5	21.1	Y
$\bar{x}$	26.5	181.0	96.2	77.0	5.86	96.2	21.6	20.3	
SEM	1.1	1.8	0.8	2.3	0.21	1.5	0.7	0.7	

Abbreviations: A-P = antero-posterior, circum. = circumference, Y = yes, N = no.

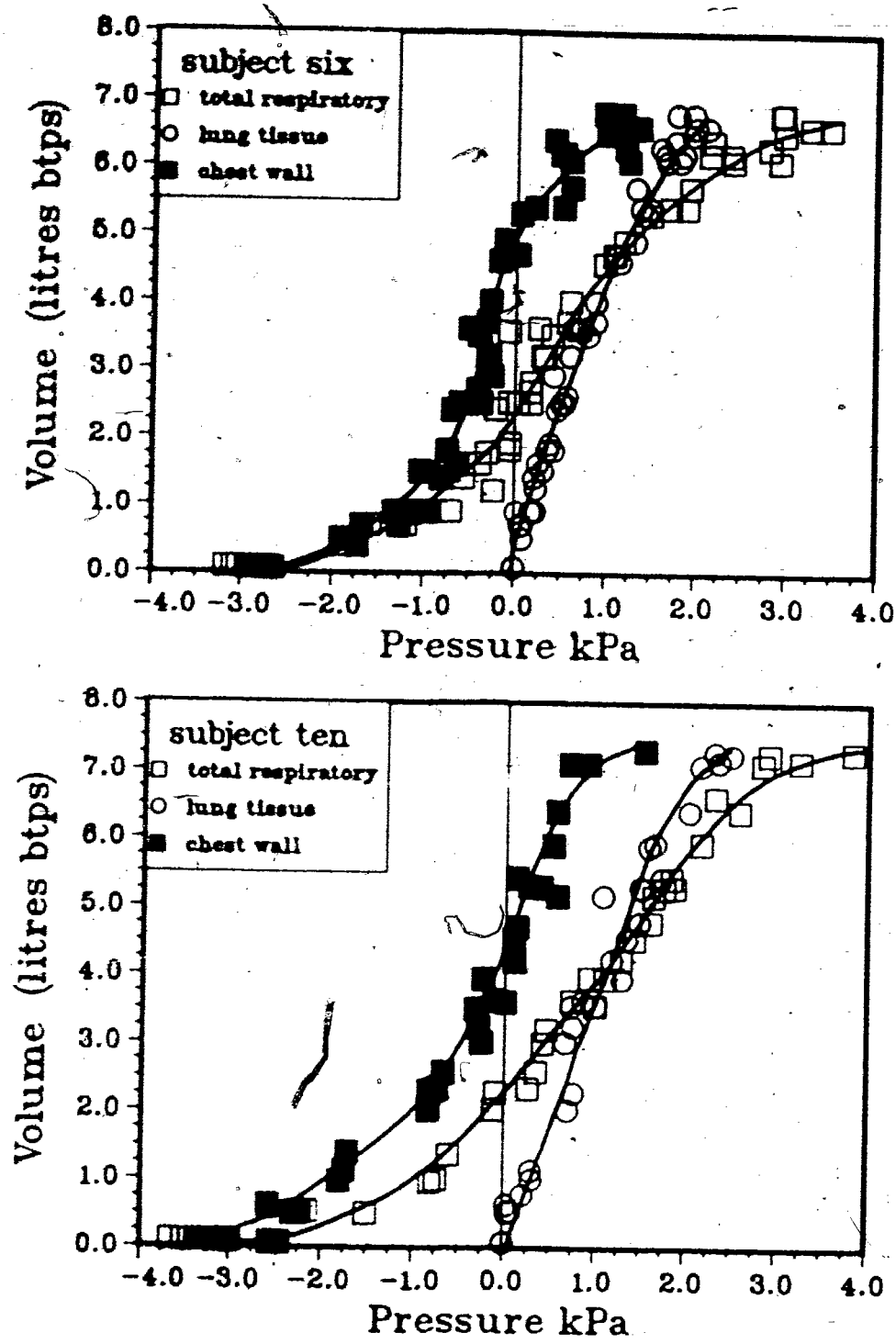


Figure 3.5: Representative pressure–volume curves for lung tissue, chest wall and total respiratory system, obtained while upright in air.

Table 3.4: **Respiratory, lung tissue and chest wall compliances for subjects upright in air.**

SUBJECT	Total respiratory compliance	Lung tissue compliance	Chest wall compliance
1	1.47	2.00	5.26
2	1.49	3.70	2.50
3	1.61	3.23	3.23
4	1.92	3.23	4.35
5	0.95	3.70	1.28
6	2.22	3.70	5.56
7	1.54	4.00	2.50
8	1.00	3.13	1.47
9	1.10	2.22	2.08
10	2.04	3.45	5.00
$\bar{x}$	1.53	3.24	3.32
SEM	0.14	0.21	0.51

Compliance units = $l.kPa^{-1}$.
Compliances computed over 1 litre commencing at the total respiratory relaxation volume.

Table 3.5: **Elastic work partitions of the thorax for subjects upright in air.**

SUBJECT	Total respiratory	Lung tissue	Chest wall	Work of respiratory muscles
1	0.340	0.613	−0.283	0.340
2	0.362	0.595	−0.269	0.362
3	0.341	0.653	−0.318	0.341
4	0.118	0.691	−0.581	0.118
5	0.552	0.830	−0.268	0.552
6	0.131	0.561	−0.443	0.131
7	0.364	0.482	−0.107	0.364
8	0.599	0.516	+0.082	0.599
9	0.524	0.792	−0.301	0.524
10	0.236	0.644	−0.415	0.236
$\bar{x}$	0.357	0.638	−0.290	0.357
SEM	0.052	0.035	0.057	0.052

Elastic work units = kPa (or $J.l^{-1}$).

The summation of lung tissue and chest wall elastic work should be equivalent to total respiratory elastic work. This is not precisely achieved since the three compliance curves were individually curve fitted, producing minor differences in the integrated areas.

3.4.3 Elastic work partitions during immersions.

The total respiratory compliance curves were first analysed for isovolume pressure displacements of the upright control V_R (as per Chapter Two). Curves were constructed from an average of 40 (s = 8) data points between RV and TLC, expressed at pressures relative to sternal notch pressure. Upright lung *centroid* was found to be 12.2cm (SEM = 1.6) below the sternal notch, thus P_{LC} was 1.19kPa. This point is in close agreement with the weighted average derived from data adjusted to a common anatomical reference (Table 2.1, 12.7cm), and the observations of Chapter Two (13.6cm). Differences were non-significant for paired (nine current subjects participated in Chapter Two) and independent t analyses (ø = 0.66) (Table 3.6).

Lung tissue and chest wall compliance curves were constructed from 40 (s = 9) and 28 (s = 10) data points between RV and TLC. Typical curves, illustrating reproducibility, are contained in Figure 3.6. Total respiratory and chest wall curves were obtained relative to pressures at the same depth as individual lung *centroids*. Component thoracic compliances are provided in Table 3.7.

Immersion elevated chest wall compliance (*i.e.* reduced its elastance) to 4.84 $l.kPa^{-1}$ (SEM = 0.89) from 3.32 $l.kPa^{-1}$ (SEM = 0.51) in the dry controls, possibly via removal of gravitational pull on the shoulder girdle (Craig 1960, Agostoni *et al.* 1966) ($p>0.05<0.1$, ø = 0.57). The chest, during immersion, recoiled towards a larger relaxation volume: 73%VC (SEM = 6.67). Total respiratory compliance (1.92 $l.kPa^{-1}$, SEM = 0.18) and lung tissue compliance (3.52 $l.kPa^{-1}$, SEM = 0.22) were not significantly altered (ø = 0.56). This latter observation opposes lung compliance reductions found by Sterk (1970, 1973) and Dahlbäck *et al.* (1979), using dynamic and quasi-static manoeuvres.

Elastic work partitions are summarised in Figures 3.7A,B and 3.8. With pressure-volume curves expressed relative to P_{LC}, elastic work calculations were performed over a 1 litre volume, commencing from the total respiratory relaxation volume. Subsequent calculations were performed at three lung volumes, each representing the volume point along the total respiratory pressure-volume curve which coincided with the hydrostatic pressure (relative to P_{LC}) at mouth depth, and at depths of 10cm above and below each subject's lung *centroid loci*. In all instances, elastic work changed sequentially with pressure change[36]. Air supply at P_{LC} during immersion, most closely reproduced elastic work subdivisions and inspiratory muscle work that existed in air. With gas delivery at 0.98kPa greater than P_{LC}, the regulator aided inspiration the greatest amount, producing the least inspiratory muscular work,

[36] Overall MANOVA F = 283.96 (4,6), $p<0.05$.

Table 3.6: **Upright isovolume compliance curve displacements during immersion with facial counter-pressure.**

SUBJECT	ISOVOLUME CURVE DISPLACEMENT(cm)
1	11.01
2	11.28
3	16.43
4	7.30
5	20.65
6	12.73
7	6.56
8	4.76
9	18.46
10	12.60
$\bar{x}$	12.18
SEM	1.64

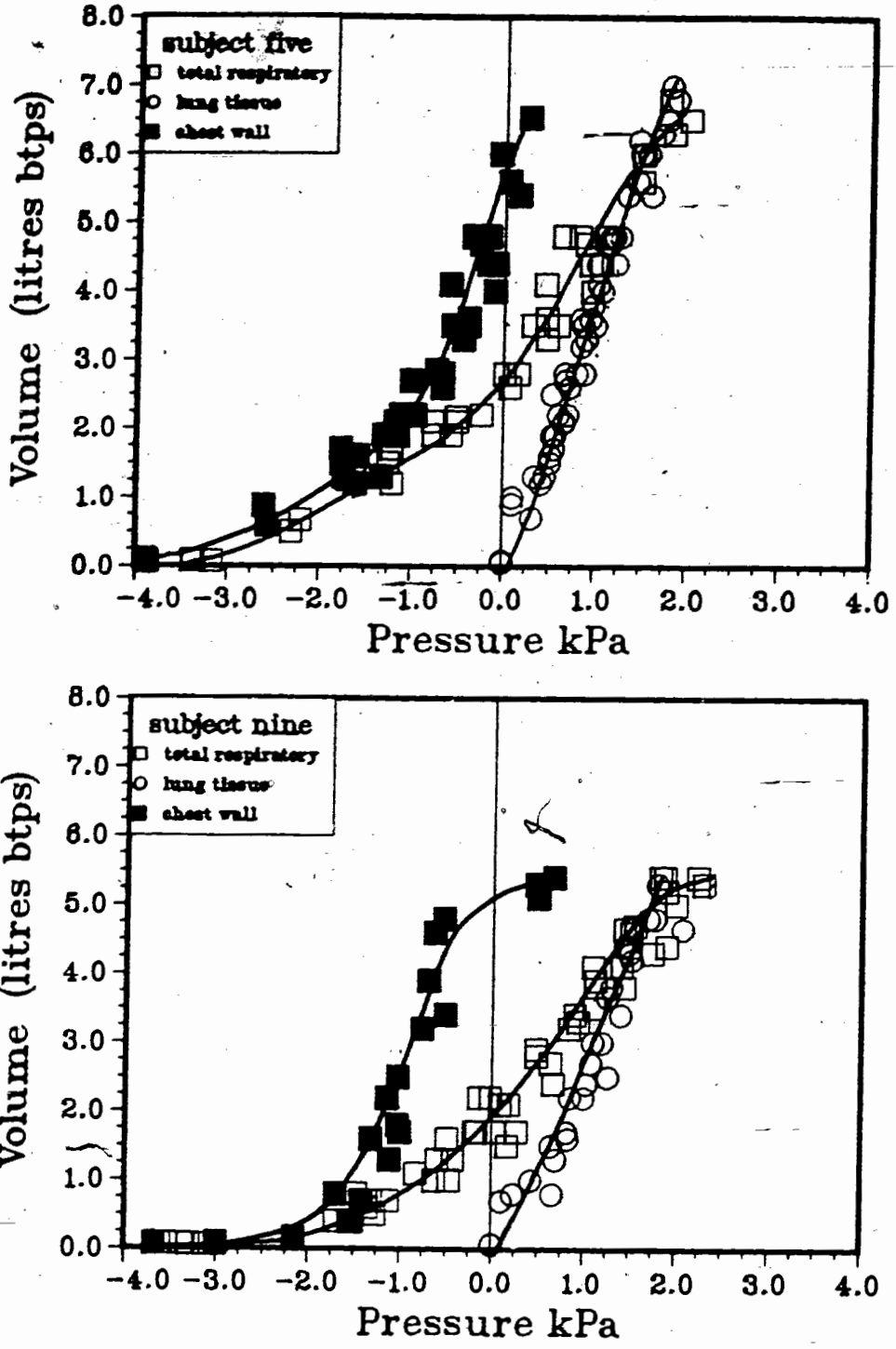


Figure 3.6: Representative pressure–volume curves for lung tissue, chest wall and the total respiratory system, obtained during immersion with facial counter-pressure.

Table 3.7: **Total respiratory, lung tissue and chest wall compliances for subjects immersed upright with facial counter-pressure.**

SUBJECT	Total respiratory compliance	Lung tissue compliance	Chest wall compliance
1	1.47	2.78	3.13
2	1.59	3.85	2.70
3	2.22	4.00	5.00
4	2.49	3.33	9.80
5	1.69	3.70	3.13
6	2.94	4.17	9.90
7	1.75	3.13	4.00
8	1.04	2.86	1.64
9	1.64	2.63	4.35
10	2.38	4.76	4.76
$\bar{x}$	1.92	3.52	4.84
SEM	0.18	0.22	0.89

Compliance units = l.kPa^{-1}.
Compliances computed over 1 litre commencing at the control total respiratory relaxation volume.

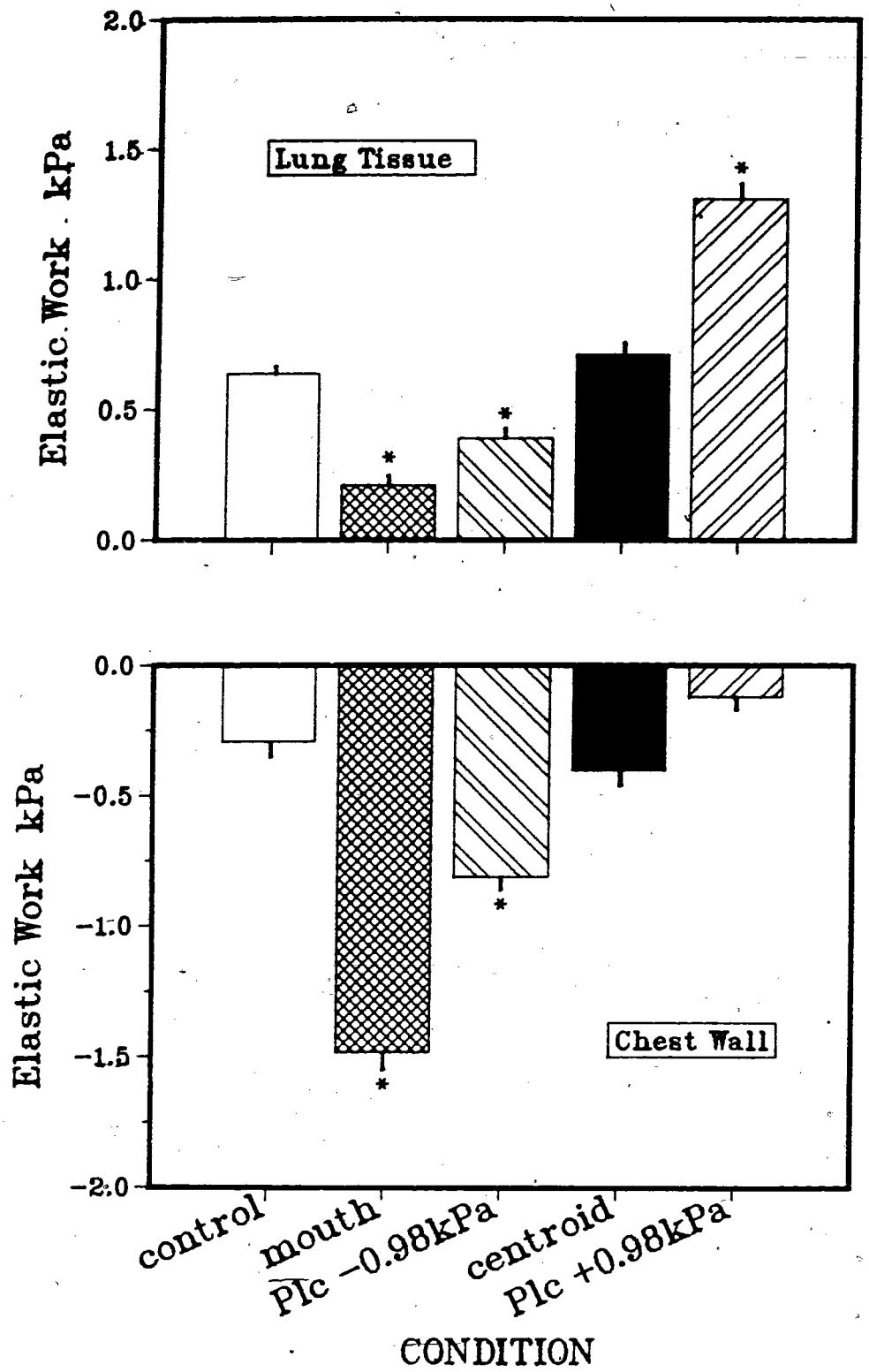


Figure 3.7A: **Lung tissue and chest wall elastic work, computed from pressure–volume curves adjusted to *centroid* pressure, using data obtained in air and during immersion at four breathing pressures.** [Work was calculated in $J.l^{-1}$ and is expressed in kPa units. Data represent means and standard errors. *=significantly different from control status ($p<0.05$).]

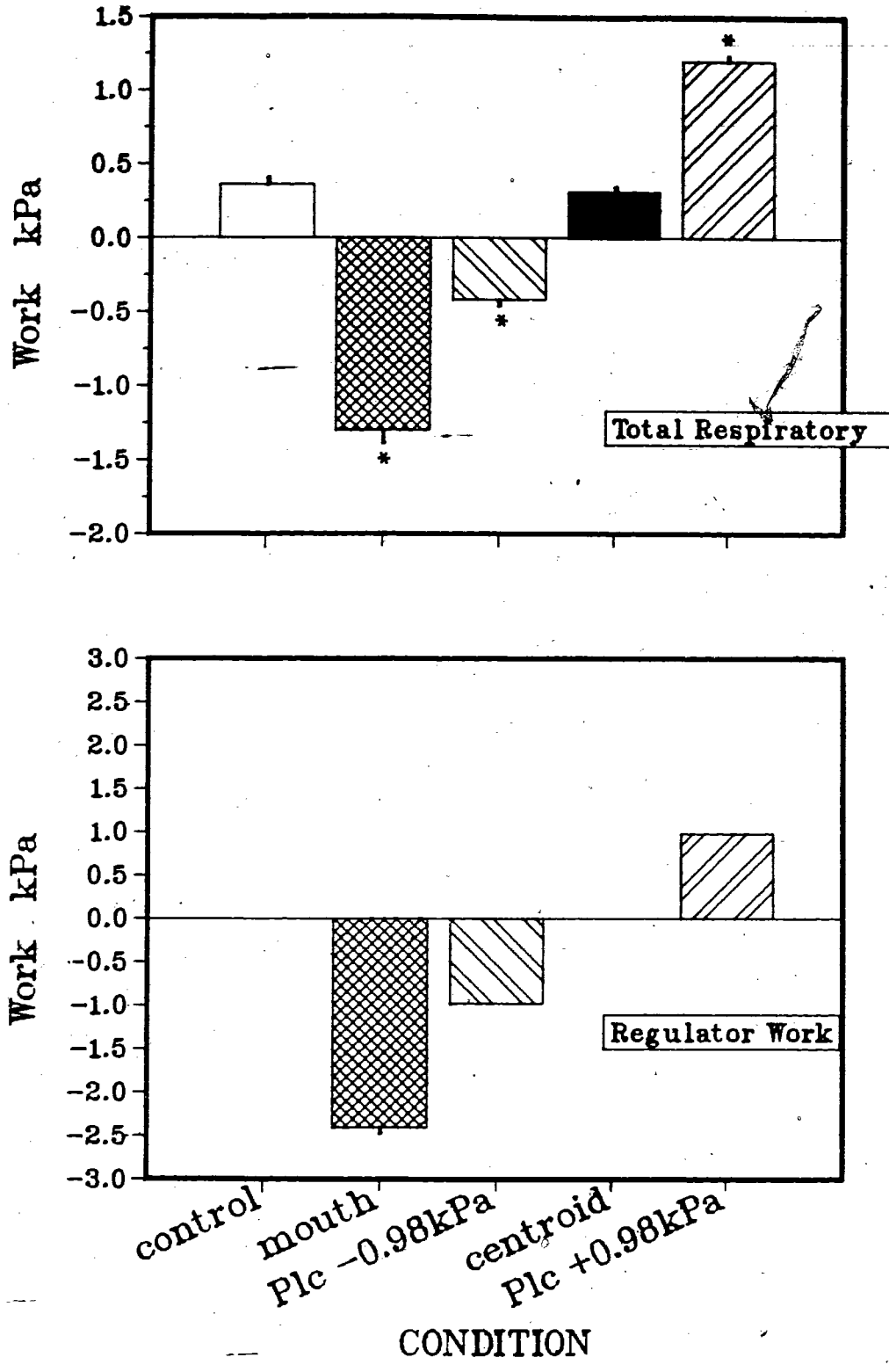


Figure 3.7B: **Total respiratory elastic work, and hydrostatic work performed by the demand regulator, computed from pressure–volume curves adjusted to lung *centroid* pressure, using data obtained in air and during immersion at four breathing pressures.** [Work was calculated in $J.l^{-1}$ and is expressed in kPa units. Data represent means and standard errors. *=significantly different from control status ($p<0.05$).]

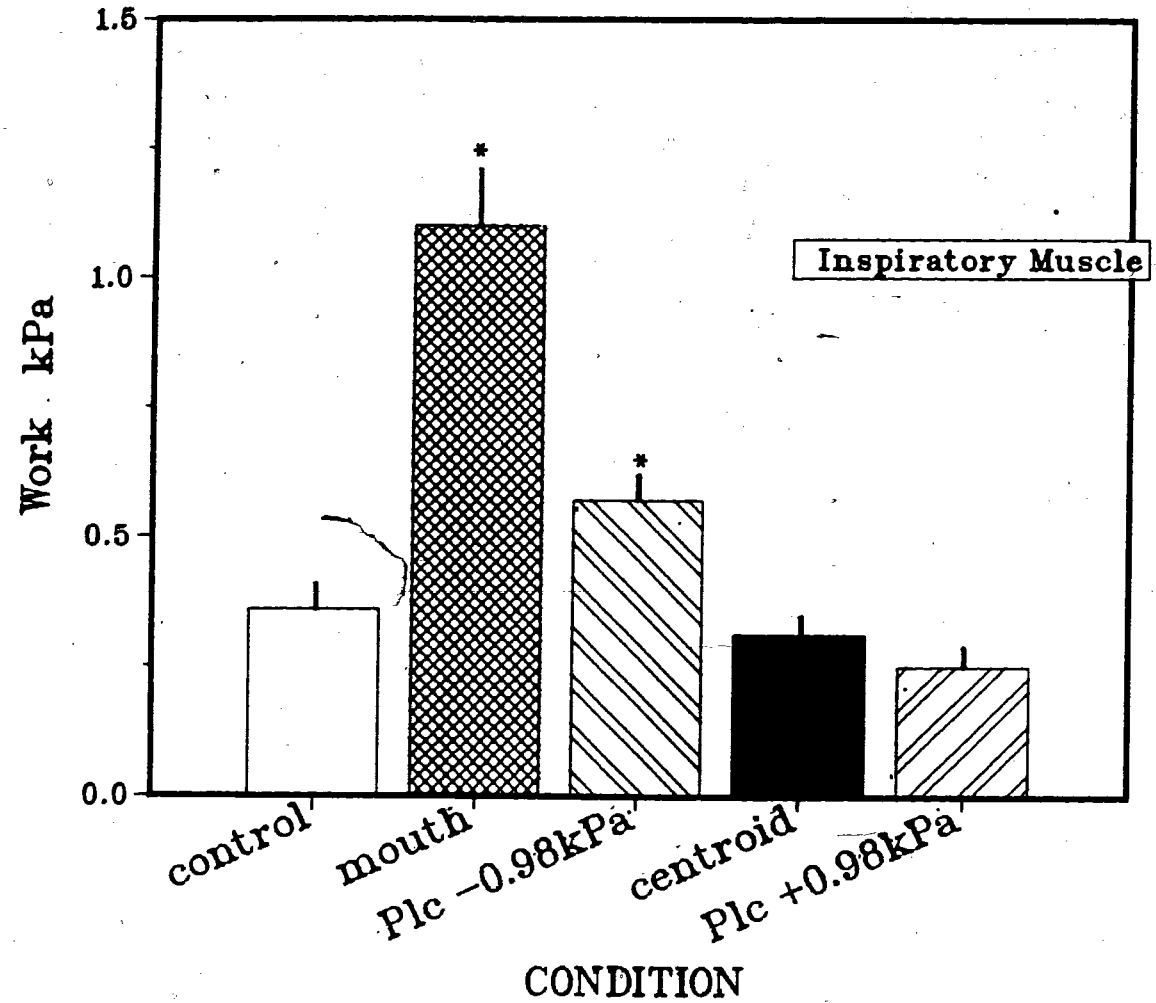


Figure 3.8: **Work performed by the inspiratory muscles, computed from pressure–volume curves adjusted to *centroid* pressure, using data obtained in air and during immersion at four breathing pressures.** [Work was calculated in $J.l^{-1}$ and is expressed in kPa units. Data represent means and standard errors. *=significantly different from control status ($p<0.05$).]

however, work was not significantly less than that recorded in air or during immersion at P_{LC} air delivery.

Differences in elastic work between control and P_{LC} air delivery during immersion, were non-significant for each elastic work component (Figures 3.7A,B, $p>0.05$, $\phi > 0.99$). Furthermore, with P_{LC} air supply total respiratory and lung tissue elastic work levels were significantly closer to control values ($p<0.05$) than that produced by each of the other breathing pressures. Chest wall elastic work at P_{LC} and P_{LC} plus 0.98kPa, differed significantly from levels measured at the other air pressures, but were not significantly divergent from control status ($p>0.05$, $\phi > 0.99$).

Mouth pressure air delivery produced elastic work levels which, in every case, differed significantly from values at control and the other three air supply pressures (Figures 3.7A,B, $p<0.05$).

P_{LC} plus 0.98kPa air supply produced significant changes from control and P_{LC} air delivery for total and lung tissue ($p<0.05$), but not for chest wall elastic work (Figures 3.7A,B, $p>0.05$, $\phi > 0.99$). With air provision at P_{LC} minus 0.98kPa, all elastic work components differed significantly from levels obtained under each of the other experimental and control states. [Summaries of *post hoc* comparisons are contained in Appendix Three].

Inspiratory muscle work quantified total elastic and hydrostatic work over a one litre inspiration. With mouth pressure air delivery, muscle work averaged 1.10kPa, and significantly exceeded control and all other experimental values (Figure 3.8, $p<0.05$). Application of positive compensatory pressures sequentially lowered inspiratory muscle work. At P_{LC} and P_{LC} plus 0.98kPa, average muscle work was no longer significantly different from control status (0.31 and 0.25kPa respectively; Figure 3.8; $p>0.05$, $\phi > 0.99$), but was significantly lower than than that produced with air delivery at P_{LC} minus 0.98kPa.

Altering the breathing pressure changed the total respiratory relaxation volume (V_R), moving the one litre V_T up and down the compliance curve (Figure 3.9), and producing concommitant total elastance alterations. The V_R increased systematically with breathing pressure, producing associated decrements in inspiratory muscle work.

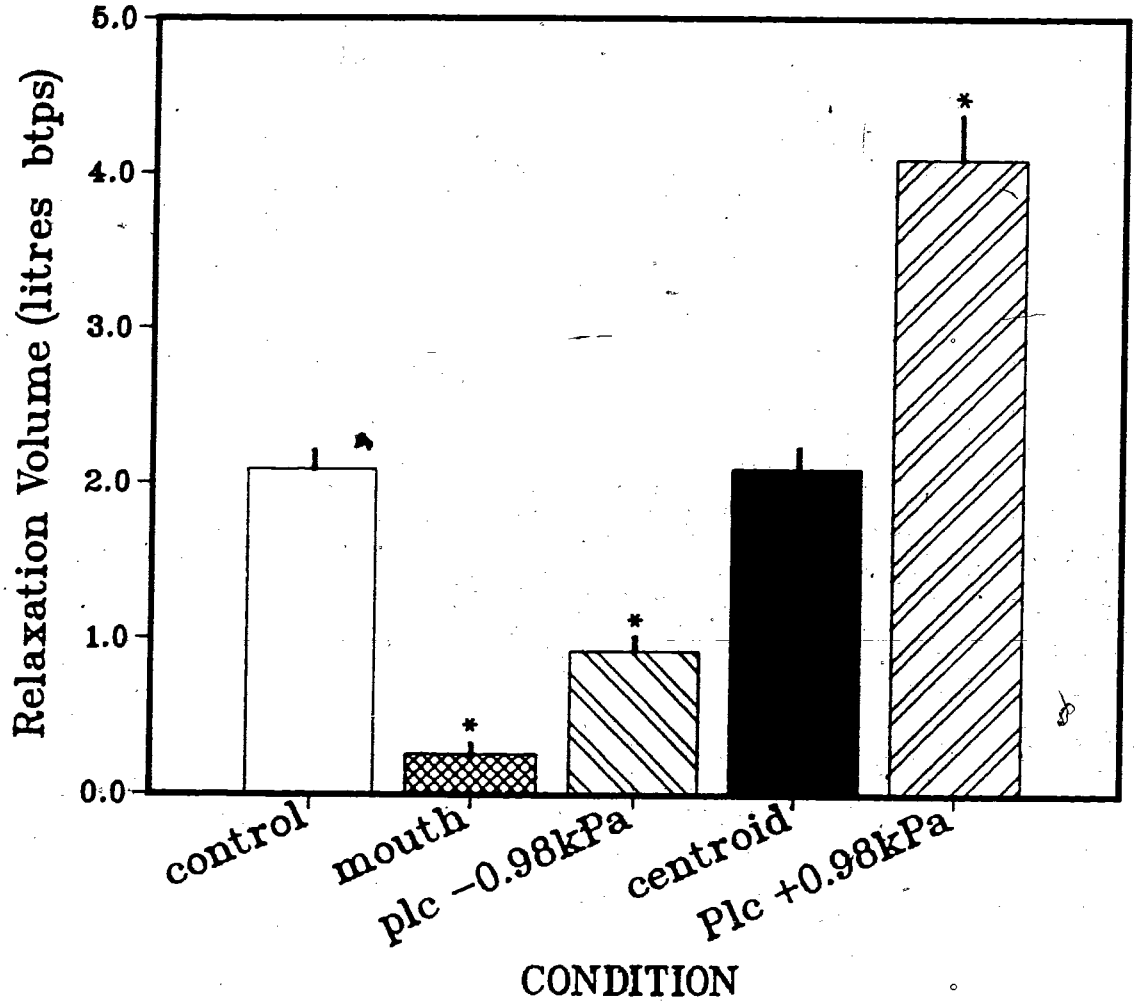


Figure 3.9: **Total respiratory system relaxation volumes accompanying breathing pressure changes.** [The equivalence of control and *centroid* pressure data during immersion, resulted from the isovolume repositioning of the immersed total respiratory compliance curve so that relaxation volumes of the two states were equivalent for each subject. Data represent means and standard errors. *=significantly different from control status ($p<0.05$).]

3.4.4 Lung volumes.

Spirometrically determined lung volumes were measured in air and during immersion. The air delivery pressure was varied randomly and covertly between trials. Control lung volumes agreed with levels previously reported (Table 3.2), except for V_T, which was generally larger (Figures 3.10 and 3.11).

Immersion with gas delivery at mouth pressure reduced VC from 5.86 to 5.57l BTPS, though vital capacities did not differ significantly from controls ($p>0.05$, $\phi > 0.99$) in any of the immersed conditions. However, with air provided at P_{LC} and P_{LC} plus 0.98kPa, the VC was significantly larger than at mouth pressure ($p<0.05$[37]). ERV was reduced from 2.06 to 0.86L BTPS with uncompensated immersion, producing reciprocal increments in inspiratory capacity (IC, 24.5%) and inspiratory reserve volume (IRV, 14.4%), in agreement with earlier works (Table 3.2).

During immersion ERV was sequentially elevated with breathing pressure increments (Figure 3.11), with the mean values at mouth pressure diverging significantly from those obtained under all the other conditions, except for air supply at P_{LC} minus 0.98kPa ($p<0.05$). Mean ERV at *centroid* and P_{LC} minus 0.98kPa were not significantly different ($p>0.05$, $\phi > 0.99$). IRV similarly varied with delivery pressure.

Increments in air delivery pressure systematically reduced IC. At P_{LC} plus 0.98kPa air supply, the IC was significantly lower than at each of the other conditions (Figure 3.10, $p<0.05$). Differences between mouth and *centroid* pressure air deliveries were also significant ($p<0.05$).

Tidal volume changes did not follow the changes in breathing pressure. All variances were non-significant ($p>0.05$, $\phi > 0.99$). VT at P_{LC} air delivery was however, closest to control volumes.

With the exception of VC, air supplied at P_{LC} during immersion was best able to return lung volumes to control status. These data conform with the working definition of lung *centroid* pressure, as being the air pressure required to return immersed thoracic relaxation volume towards values which existed in air.

[37] Overall MANOVA $F = 391.73$ (5,4), $p<0.05$.

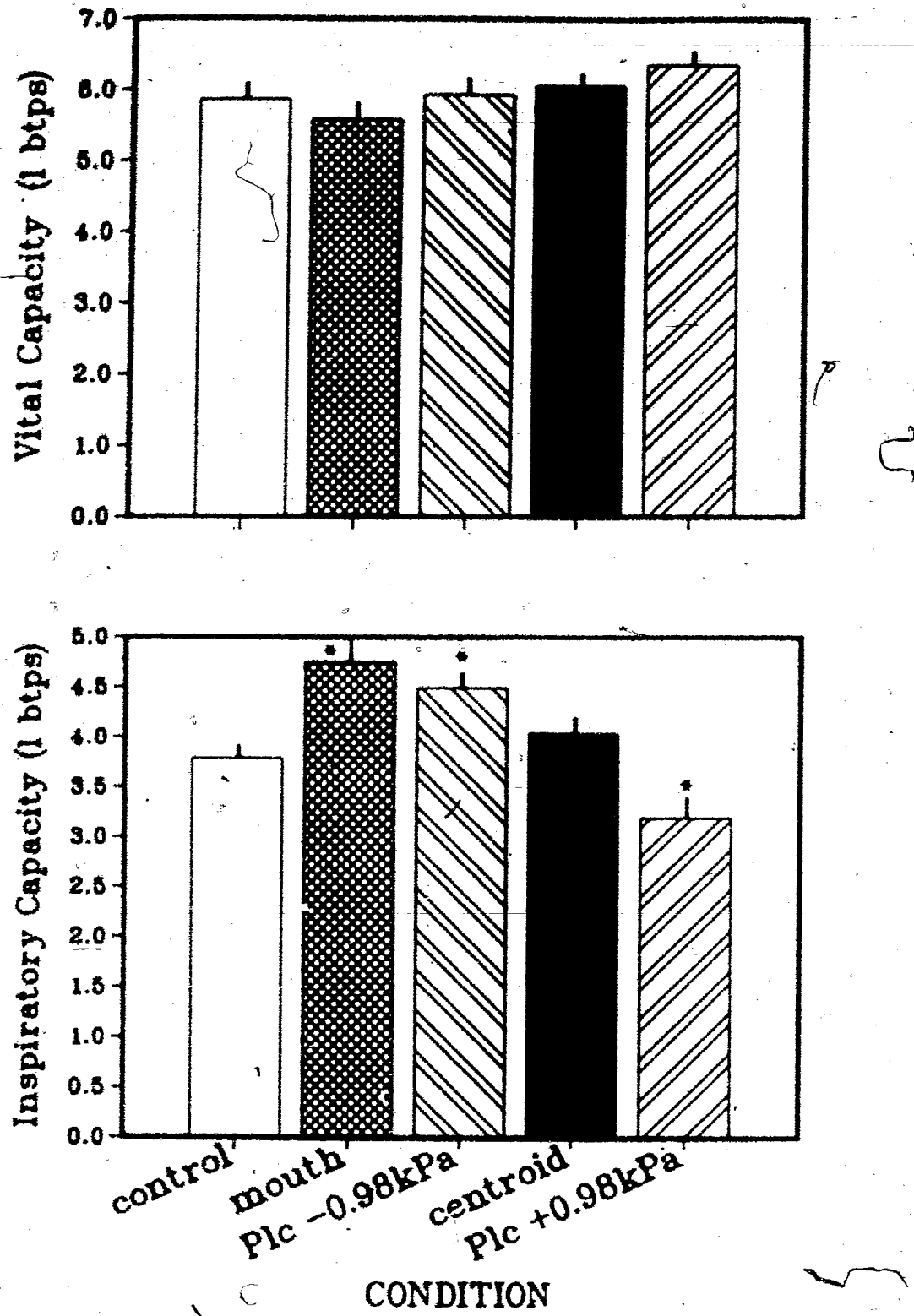


Figure 3.10: Vital capacities and inspiratory capacities obtained in air, and during immersion with and without breathing pressure compensation. [Data represent means and standard errors. *=significantly different from control status ($p<0.05$).]

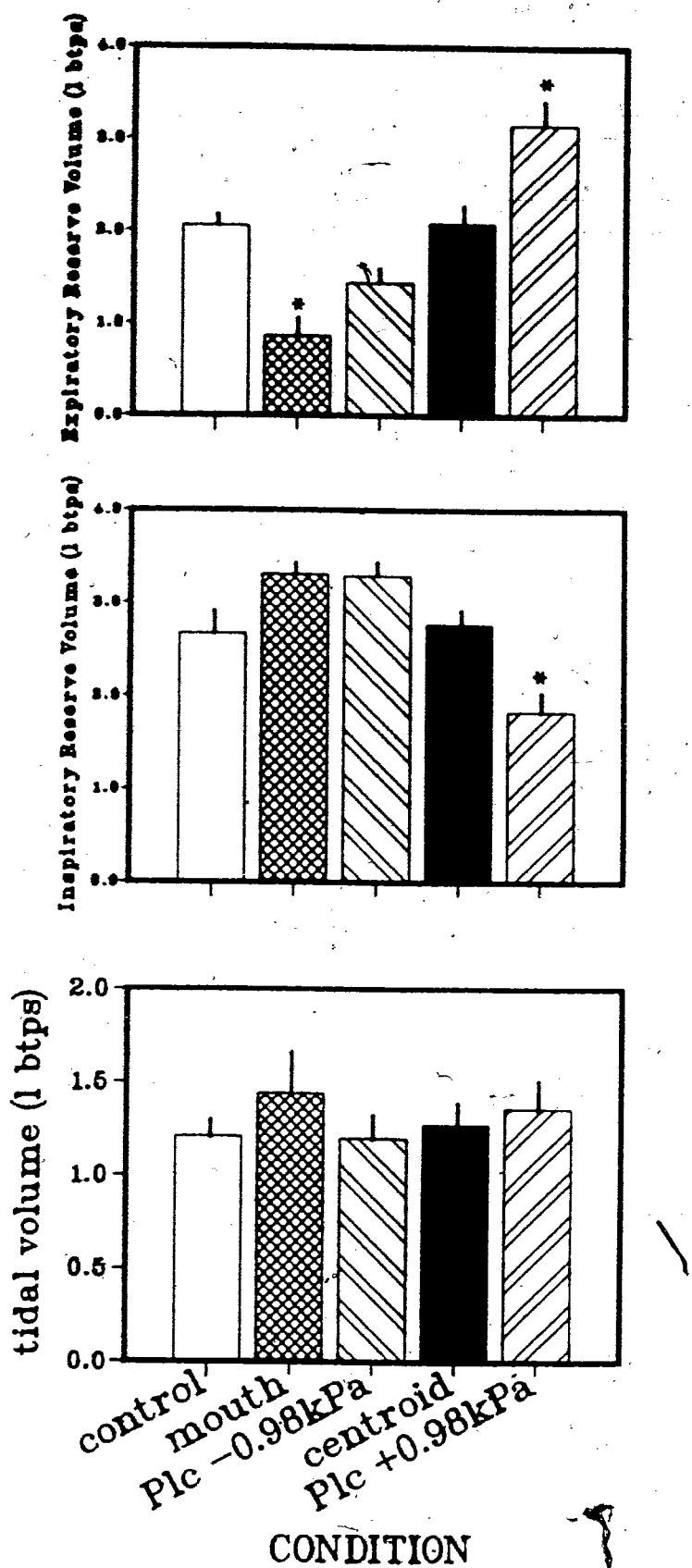


Figure 3.11: **Expiratory reserve volumes, inspiratory reserve volumes and tidal volumes obtained in air, and during immersion with and without breathing pressure compensation.** [Data represent means and standard errors. *=significantly different from control status ($p<0.05$).]

3.5 DISCUSSION

3.5.1 Verification of lung centroid

The present study positioned lung *centroid* 12.2cm inferior to the sternal notch (Table 3.6), confirming observations from Chapter Two. During immersion, with air provided at the group mean lung *centroid* pressure, the mean expiratory reserve volume was returned to within 50mls of the control value (Figure 3.11). Furthermore elastic and inspiratory muscle work were returned to control levels. It was concluded, in accordance with the definition of lung *centroid* used within this thesis, that the individual and group mean lung *centroids* were accurately and validly determined.

3.5.2 Validity of oesophageal balloon techniques.

Measurement of chest wall and lung tissue elasticity is dependent upon the fidelity with which P_{pl} can be approximated by P_{oes}. P_{pl} varies along the sagittal (Mead 1961, Milic-Emili *et al.* 1964b, 1966, D'Angelo *et al.* 1970, Agostoni *et al.* 1970) and transverse planes (Agostoni and D'Angelo 1969, Miserocchi *et al.* 1981). P_{oes} cannot be considered a measure of mean pleural surface pressure, but rather as an index of local average P_{pl}, which varies with the site measurement.

P_{oes} has been consistently shown to track P_{pl} (Fry *et al.* 1952, Stead *et al.* 1952, Cherniack *et al.* 1955, Attinger *et al.* 1956a, Butler *et al.* 1957, Ehrner 1960), while remaining slightly more positive and marginally failing to replicate peak-to-trough dimensions during cyclic changes. Cherniack *et al.* (1955) suggested that P_{oes} provided a better appreciation of non-elastic than elastic components, though subsequent technical advances largely negate their observation (Milic-Emili *et al.* 1964a, 1964b).

P_{oes} is susceptible to pressure artifact of intrinsic and extrinsic origin (*e.g.* cardiogenic pulses, peristaltic contractions, movement artifact) of which most are removed via technical or measurement allowances. Some artifacts are difficult to detect and remove. At volume extremes, non-peristaltic contractions and/or oesophageal compression have been reported (Fry *et al.* 1952, Ehrner 1960, Knowles *et al.* 1960, Permutt and Martin 1960, Milic-Emili *et al.* 1964a, 1964b, Gibson and Pride 1976, Clarysse and Demedts 1985). The latter are attributed to mediastinal compression due to posterio-lateral movement of the heart, as the diaphragm moves up at low volumes (Knowles *et al.* 1960). Bondurant *et al.* (1960b) and Wood *et al.* (1971) suggested mediastinal expansion may also cause compression artifacts. Donleben (1959) found oesophageal compliance changes adversely affected P_{oes} measurements, while Echt *et al.* (1974) and Johnson *et al.* (1975) reported that the P_{oes} gradient changed with

immersion.

Given the above limitations, and current procedures, it is believed that present techniques provided valid approximations of local P_{pl} changes. This view is supported by pre– and post–trial P_{oes}, which averaged –0.41kPa (–4.60cmH_2O) and –0.41kPa (–4.16cmH_2O) respectively, and by the normality of control pulmonary compliances.

3.5.3 Elastic work during immersion.

Since immersion is physically analogous to negative pressure breathing, one would anticipate that elastic and inspiratory muscle work would be increased. Hong *et al.* (1969), Flynn *et al.* (1975) and Chapter Two reported such increments in total respiratory elastic work, due to compliance curve pressure shifts. However, by definition, transrespiratory pressure (P_{trs}) is referenced to body surface pressure (Rahn *et al.* 1946, Agostoni and Mead 1964). The salient point concerns the selection of the appropriate P_{bs}. In the present chapter, P_{bs} was chosen as the hydrostatic pressure in the same horizontal plane as individual lung *centroid loci*, rather than P_A, or pressure at the sternal notch. Using this reference, the P_{trs} at the respiratory relaxation volume was equivalent to that obtained in control trials.

With data referenced to P_{LC}, subjects breathed up and down stationary compliance curves. Altering breathing pressures by moving the demand regulator vertically between the mouth and a point 10cm below lung *centroid* , caused subjects to change relaxation volume without altering the zero pressure intercept (Figure 3.9). Thus total respiratory elastic work, with negative pressure at the relaxation volume (*i.e.* regulator at mouth or 10cm above lung *centroid*), was negative, while inspiratory muscle work was positive (*i.e.* inspiratory muscle work = (total respiratory elastic work) – (work provided by regulator): Figure 3.4B). Conversely, a positive pressure at the relaxation volume (*i.e.* regulator at 10cm below lung *centroid*) required a positive total respiratory elastic work, which was provided partially by the air delivery pressure, and partly by inspiratory muscle work (Figure 3.7). Control and *centroid* derivations followed the same convention.

In all instances elastic work changed sequentially with breathing pressure (Figures 3.7A,B, 3.8). The demand regulator provided negative work when positioned above lung *centroid* depth, and was analogous to negative pressure breathing, aiding only expiration. Chest wall recoil aided inspiration as the chest attempted to expand to about 73% of VC. .

At positive pressure air provision (relative to *centroid*), inspiratory muscular work was minimal, as observed by Rahn *et al.* (1946) and Flynn *et al.* (1975). Chest wall elastic work was marginally negative, aiding inspiration. Lung tissue work was maximal, as V_T excursions were now over regions of greatest elastance, and the P_{tp} curve was further from the zero pressure axis. Positive regulator work (relative to P_{LC}) aided inspiration while impeding expiration.

Rahn *et al.* (1946) found a 'U-shaped' relation between total respiratory elastic work in air and breathing pressure (Figure 3.12). When breathing pressures were expressed relative to P_{LC}, current data revealed a similar relationship between the respiratory muscle work and breathing pressure, over the range from -2.43 to 0.98kPa (Figure 3.13). To reduce energy used to expand elastic tissues, the use of breathing pressures close to P_{LC} appears desirable during immersion. A greater delivery pressure produced a further, but non-significant muscle work reduction. Such breathing pressures may produce pharyngeal discomfort, unless appropriately countered with externally applied facial and throat pressures (Thompson and McCally 1966, 1967). It is concluded that during upright immersion, gas delivery at P_{LC} provides more favourable total respiratory, chest wall and lung tissue elastic mechanical status, since it best replicates upright status in air.

In 1947 Paton and Sand postulated hydrostatic compression during immersion would render the thorax more compliant. Agostoni and Rahn (1960b) observed chest wall compliance elevation in subjects immersed to the xyphoid process. In subsequent immersions to the neck (Agostoni *et al.* 1966), chest wall compliance curves similarly revealed an increased compliance (Figure 3.14), though individual and mean compliances were not reported. Current data confirmed these observations, with 80% of the present subjects experiencing chest wall compliance increases (Tables 3.4, 3.7) accompanying immersion, producing an elevation in the chest wall V_R from 61% to 73% of VC.

It is hypothesised that chest wall compliance elevations were mediated by buoyancy effects. Craig (1960) found shoulder girdle weight to influence both the total respiratory compliance and the ERV. During immersion the arms will become moderately buoyant, minimising the weight on the shoulder girdle and also lessening thoracic weight. Under such conditions less pressure is required to produce an equivalent volume expansion and, by definition (*i.e.* dV/dP), the chest wall compliance increases.

The observation that static pulmonary compliance remained unaltered during immersion requires further consideration. Given that immersion elevates thoracic blood volume (Arborelius *et al.* 1972b, Lollgen *et al.* 1980), pulmonary compliance changes accompanying immersion have been ascribed to

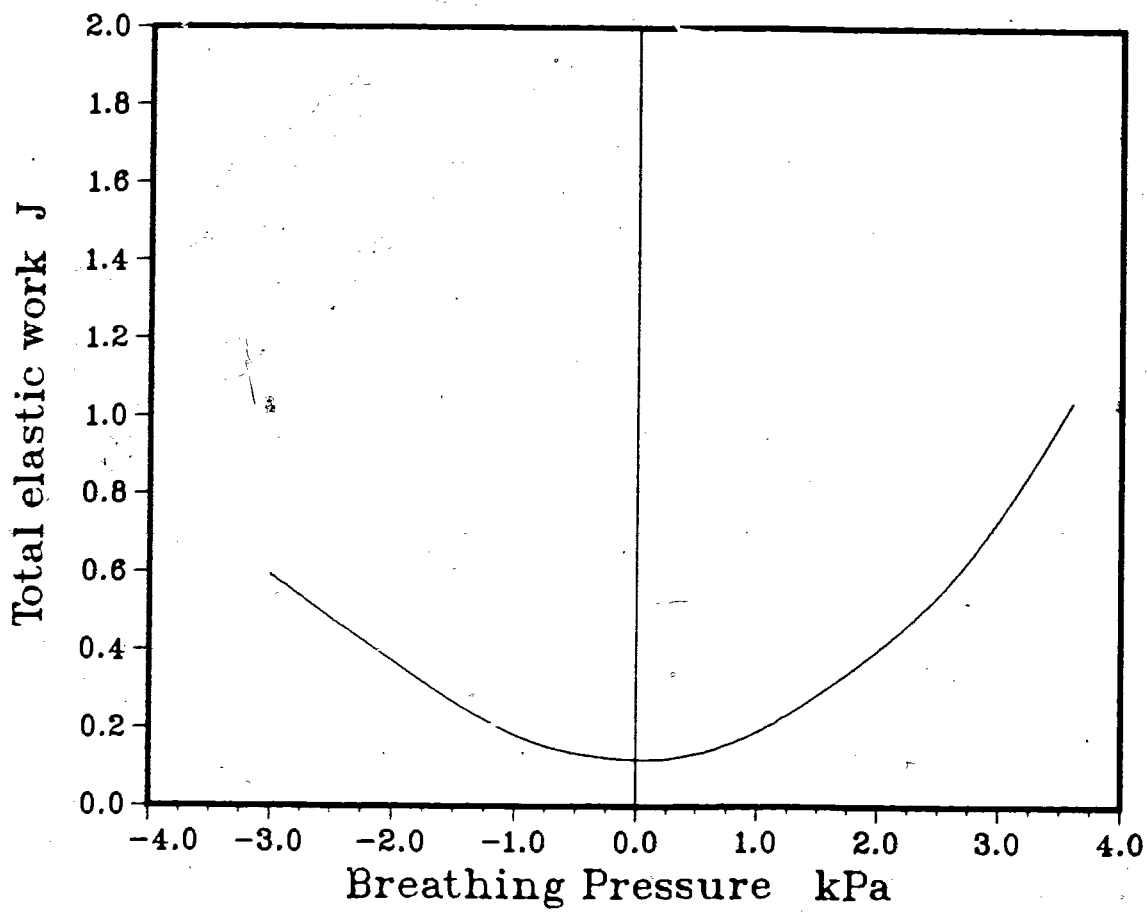


Figure 3.12: **Total respiratory elastic work and breathing pressure.** [Modified from Rahn *et al.* (1946). Originally the negative slope represented inspiratory work, while the positive slope represented expiratory work. However, these were combined as total elastic work over regions where the two curves overlapped.]

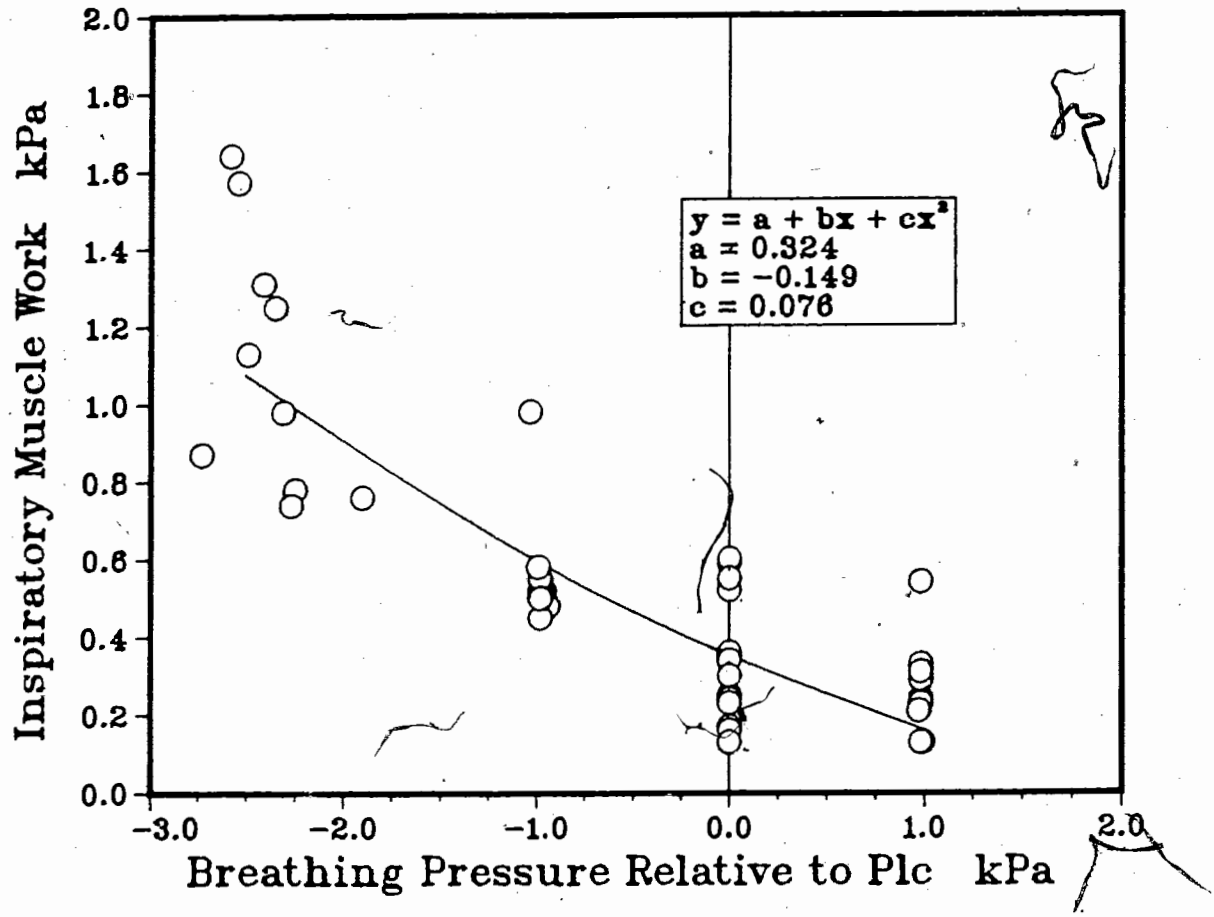


Figure 3.13: **Inspiratory muscle work as a function of breathing pressure (relative to *centroid* pressure during upright immersion).** [Data points represent current experimental observations, while the curve describes the exponential nature of the relation between air pressure delivery and inspiratory muscle work.]

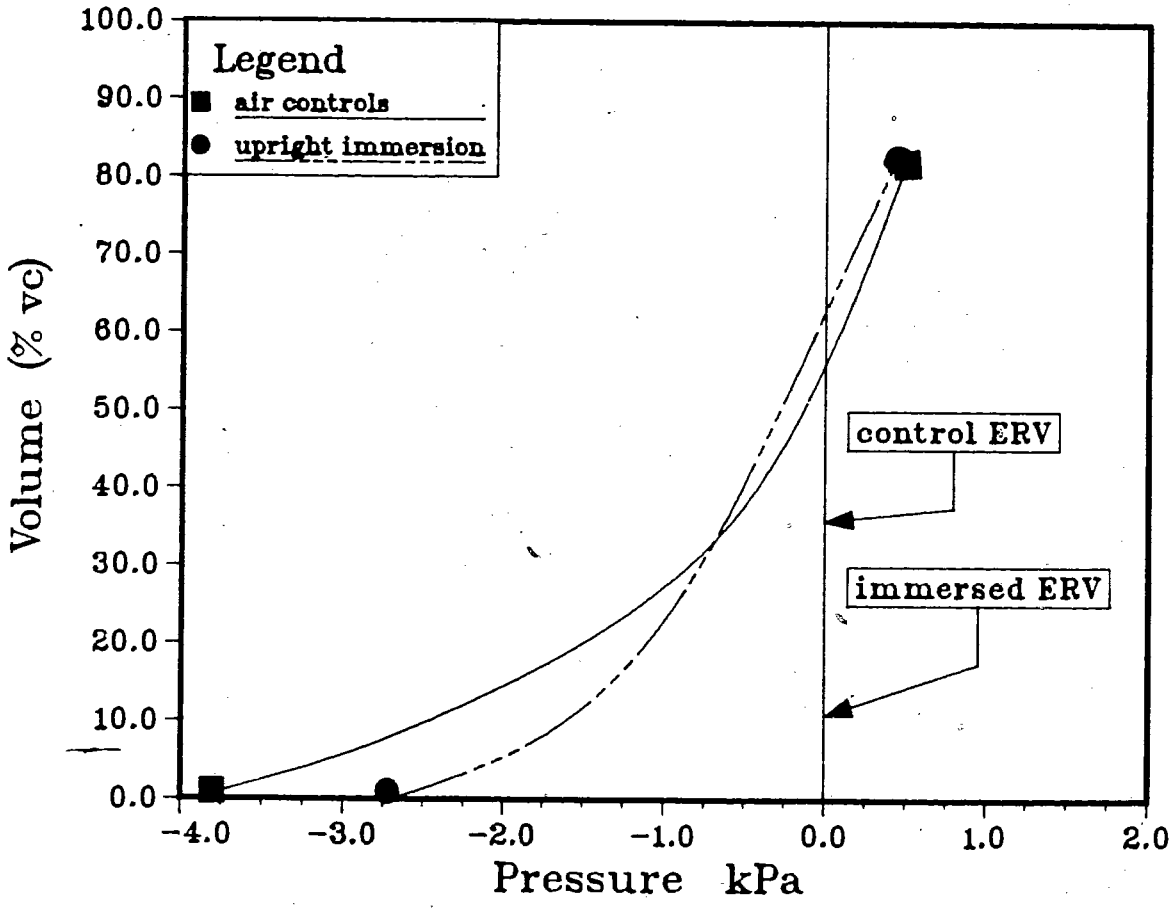


Figure 3.14: **Pressure–volume curves of the relaxed chest wall, in air, and during immersion to the *neck*.** [Modified from Agostoni *et al.* (1966). The immersion curve was originally referenced to P_A, but here it is referenced to P_{LC}, to illustrate compliance change. This data adjustment was performed using the current isovolume method.]

pulmonary engorgement (Dahlbäck *et al.* 1978, 1979). Current results fail to support this suggestion, however, they do not contest engorgement, merely, that in the present study, it did not modify static pulmonary compliance.

Numerous non-immersion studies have investigated the role of pulmonary blood on compliance (Table 3.8). While most groups found vascular congestion to reduce pulmonary compliance in air, this observation is not universally supported. Pulmonary compliance during immersion has been measured using dynamic (Bondurant *et al.* 1957, Bondurant *et al.* 1960b, Sterk 1970, Dahlbäck *et al.* 1979), and quasi-static (Sterk 1973, Burki 1976, Dahlbäck *et al.* 1978) pressure-volume techniques. Compliance reductions have not been unequivocally demonstrated (see: Section 3.1). One may speculate that technical differences may account for the lack of agreement between studies, and for the failure to observe a reduction in pulmonary compliance in the present investigation. It is suggested that non-static techniques, used by other research groups, may have produced data which is not directly comparable with results obtained in the current study using static manoeuvres.

All P_{tp} measurements assume P_{ao} equals P_{alv}, but only during truly static conditions can this assumption always be correct. In quasi-static techniques the airway is normally transiently occluded (1-2 seconds) while the subject continues to slowly exhale from TLC, without relaxing (Burki 1976). Sterk (1973) and Dahlbäck *et al.* (1978), however, used quasi-static manoeuvres without occlusion using the slowest possible flow rates (after Buytendijk 1949). The assumption that P_{alv} equals P_{ao} during these manoeuvres may be questioned. Gibson and Pride (1976) and Bake[38] have suggested quasi-static P_{tp} includes a resistive pressure drop at high lung volumes, due to laryngeal braking, and at low volume due to increased airway resistance. Furthermore, at high lung volumes, tissue stress-relaxation, acting to stretch lung tissue and to lower P_{tp}, is not complete due to the transience, or lack of a pause at each lung volume. Such technical limitations may well mean that $C_{q-s(l)}$ and $C_{st(l)}$ measure different mechanical attributes.

Similarly, the $C_{dyn(l)}$ method assumes airflow has ceased at both the alveoli and mouth, and the observed P_{tp} changes are applied only to elastic elements. Radford (1964) states the latter assumption is dependent on the lungs being a single elastic structure. Since this is not the case, parallel units must have equivalent time constants for $C_{dyn(l)}$ to be used as an index of lung elasticity (*i.e.* product of unit

[38] Personal communication.

Table 3.8: **Pulmonary compliance during pulmonary vascular engorgement in air.**

REFERENCE	Method	Subjects	Compliance	Observation
Bondurant *et al.* 1957	g-suit	H (4)	$C_{dyn(l)}$	45% reduction.
Borst *et al.* 1957	mechanical	D (8)	$C_{st(l)}$ & $C_{dyn(l)}$	$C_{st(l)}$ & $C_{dyn(l)}$ unchanged by flow; $C_{st(l)}$ fell as left atrial pressure increased, $C_{dyn(l)}$ fell 20-30%.
Cook *et al.* 1959	occlusion	D (12)	$C_{dyn(l)}$	18% reduction.
Bondurant *et al.* 1960b	g-suit	H (7)	$C_{dyn(l)}$	40% reduction, 26% when FRC controlled.
Ting *et al.* 1960b	negative pressure	H (3)	$C_{st(l)}$	no change.
Daly *et al.* 1963	g-suit	H (8)	$C_{dyn(l)}$	29% reduction.
Levine *et al.* 1965	saline	D (18)	$C_{dyn(l)}$	41% reduction.
Giutini *et al.* 1966	dextran	H (5)	$C_{st(l)}$ & $C_{dyn(l)}$	$C_{dyn(l)}$: 29% reduction. $C_{st(l)}$ non-significant.
Giannelli *et al.* 1967	mechanical	D (9)	$C_{dyn(l)}$	non-significant elevation.
Wood *et al.* 1971	mitral stenosis	H (23)	$C_{st(l)}$	reduced.
Goldberg *et al.* 1975	negative pressure	H (5)	$C_{dyn(l)}$ & $C_{q-s(l)}$	$C_{dyn(l)}$: 25% reduction. $C_{q-s(l)}$: 39.4% reduction at -4kPa.
Jones *et al.* 1978	occlusion	D (10)	$C_{dyn(l)}$ & $C_{q-s(l)}$	$C_{dyn(l)}$ reduced > $C_{q-s(l)}$.
Gautier *et al.* 1982	altitude	H (9)	$C_{st(l)}$ & $C_{dyn(l)}$	no change.

Method = method of obtaining engorgement: g-suit inflation, mechanical pumping, occlusion of systemic circulation, negative pressure breathing, saline and dextran infusions, patients with congestion, altitude. Subjects: H = humans, D = dogs; subject number in parenthesis. Compliance: $C_{st(l)}$ = static, $C_{dyn(l)}$ = dynamic, $C_{q-s(l)}$ = quasi-static.

compliance and resistance[39]). Otis *et al.* (1956) have demonstrated lung unit equivalence in normal subjects, however, disease states alter unit constants causing ventilation distribution to vary with respiration frequency (Radford 1964, Gibson and Pride 1976, Cotes 1979). Changes in parallel unit time constants during immersion are unknown. Under certain respiratory disease states (*e.g.* obstructive lung disease) $C_{dyn(l)}$ becomes an index of airway resistance, and Gibson and Pride (1976) have reported that minor airway dysfunctions may mediate $C_{dyn(l)}$ abnormalities. Distribution of ventilation (Daly *et al.* 1963), V_T, absolute lung volume (Ferris and Pollard 1960) and altered inertial forces also confound its value as an index of lung elasticity in the anormal lung[40]. Since immersion has been shown to elevate lung closing volumes and induce thoracic vascular congestion (see: Section 3.1) it is suggested that $C_{dyn(l)}$ during immersion may not adequately distinguish between elastic and dynamic mechanical attributes and should be viewed with some reservation.

In summary, it is believed that differences between immersion lung compliances observed in the current investigation ($C_{st(l)}$), and those of previous studies ($C_{dyn(l)}$ and $C_{q-s(l)}$), are attributable to inherent technical differences. If uncompensated immersion elevates airway resistance, then $C_{dyn(l)}$ techniques may simply provide a reflection of this change, and may no longer represent a measure of pulmonary elastic characteristics. It is postulated that during immersion, $C_{st(l)}$ techniques may be superior to non-static methods, and that the current observation of unaltered lung tissue compliance is a valid reflection of lung tissue elastic status during upright immersion.

3.5.4 Lung volumes during immersion.

Upright, head-out immersion reduces VC (Table 3.2). Qualitatively similar results were observed during the present investigation, and were attributed to hydrostatic compression countering inspiratory effectiveness, to the central translocation of peripheral blood (Agostoni *et al.* 1966, Hong *et al.* 1969, Dahlbäck 1975, Buono 1983), or possibly to a change in RV. In air the VC is limited by the inability of inspiratory muscles to expand the lungs beyond the elastic limit of the chest wall. Current data has shown that the chest wall expands to a larger relaxation volume during immersion. This observation has not been previously reported.

[39] Unit time constants reflect time required for that unit to attain equilibrium following constant pressure perturbation.

[40] Consequently, use of $C_{dyn(l)}$ to assess pulmonary elastic properties has been abandoned (Gibson and Pride 1976).

In air resting man breathes along the total respiratory compliance curve. The relaxation volume (V_R) of the latter normally equals the expiratory reserve volume. In the current project, ERV and V_R equivalence was observed during control trials and during immersion trials with air provided at the mean P_{LC} (Figures 3.9, 3.11). However, when air was not supplied at P_{LC} during immersion, ERV deviated from the total respiratory V_R, in accordance with the changes in air delivery pressure. Similar work by Thalmann *et al.* (1979) and Lundgren (1984) has shown this same pattern, when the static breathing pressure for prone subjects was altered.

When air pressures were less than P_{LC}, subjects appeared to defend end-expiratory lung volumes at about 500–600ml above the V_R, possibly by increasing inspiratory muscle tone, since by definition the system is not relaxed. With air delivery in excess of P_{LC}, subjects defended an ERV less than the V_R. In this case increased expiratory muscle tone would have been employed to counter the tendency of positive air pressure to hyperinflate the lungs. Thus, while the ERV increased with sequential breathing pressure increments (Figure 3.11), these changes were not passive.

The interrelationship of these volumes is illustrated by the tidal air band shown in Figure 3.15. Expiratory reserve volumes falling to the left of the relaxation curve indicate elevated inspiratory muscle tone. Conversely values to the right reflect a compensatory increase in expiratory tone. These relationships in air were first investigated and reported by Rahn *et al.* (1946), using breathing pressure manipulations on upright subjects in air. The original data points of Rahn *et al.* (1946) (normalised to the current control VC) are shown in Figure 3.15. The current observations are in close agreement with Rahn's pressure breathing data, which similarly demonstrated a tendency for subjects to defend an ERV that differed from the respiratory V_R during pressure breathing. Flynn *et al.* (1975) reported similar observations when breathing pressure changes were implemented during head-out immersion. However, data in the latter study were only collected for one subject.

The active defence of the ERV by intrinsic compensatory mechanisms possibly incorporates the rapid and/or slow adapting pulmonary stretch receptors during positive pressure ventilation (Cherniack and Atose 1981). During negative pressure air provision (relative to P_{LC}) the diaphragm adopts a more favourable length-tension relationship. This may enable the development of greater inspiratory force for a given level of neural activation (Agostoni *et al.* 1966, Minh *et al.* 1977, 1979, Banzett *et al.* 1985, Reid *et al.* 1985). Mead (1979) reported that expiratory duration was also prolonged, possibly due to inspiratory effort braking expiration. Bishop (1963) found continuous negative pressure breathing to

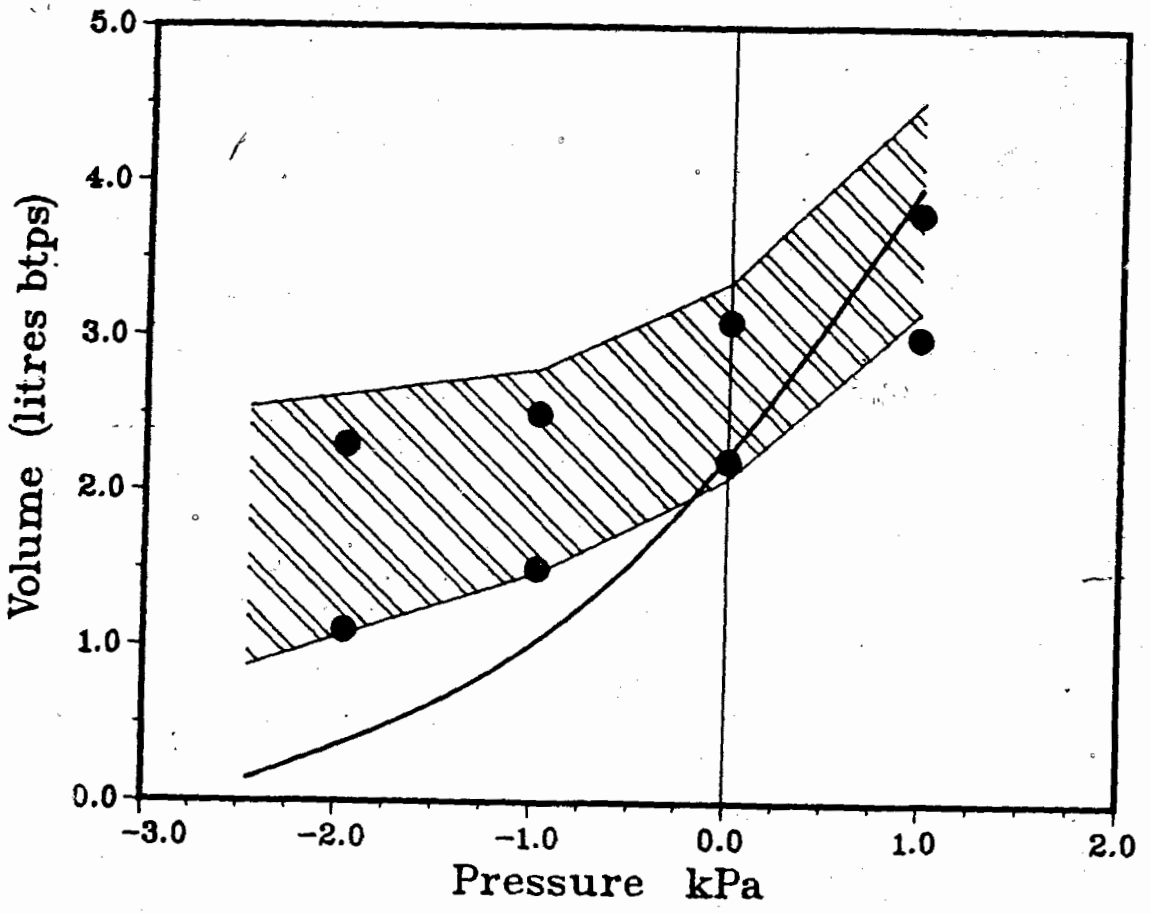


Figure 3.15: **The total respiratory pressure–volume diagram, showing expiratory reserve volumes and tidal volumes observed in air and during immersion at air delivery pressures of 0.98, 0, −0.98 and −2.43kPa relative to the lung *centroid* pressure.** [Data taken from mean respiratory relaxation volumes (thick curve), expiratory reserve volumes and tidal volumes obtained in the present investigation. The hatched region corresponds with the zone of tidal volume breathing. The line at the bottom of the hatched zone corresponds with the spirometrically measured ERV. • = original data points of Rahn *et al.* 1946).]

produce elevated diaphragmatic electrical activity. This change was observed even at the position of end–expriation, where the diaphragm is normally inactive. The maintenance of a constant inspiratory tone could have been created by muscle spindle reflexes, as suggested by Cherniack and Altose (1981). In both breathing pressure extremes, afferent feedback would probably have reached the level of the sensorimotor cortex, as subjects invariably reported having to resist both positive and negative pressure effects.

These active compensatory efforts to maintain ERV will alter muscular work during respiration. The current measures of inspiratory muscle work against the respiratory apparatus also fail to account for this work. Consequently, the respiratory muscle work during immersion (Figure 3.8) underestimates the actual muscular work required at all air delivery pressures less than P_{LC}.

During uncompensated immersion VC and ERV fell to 97% and 43% (respectively) of control values, while IC and IRV increased respectively to 124% and 114% of control levels. These data qualitatively support values reported in the literature (Table 3.2). Each of the investigations cited employed different depths of immersion. Since immersion depth determines the magnitude of thoracic compression, it will also determine lung volume subdivisions. The current project used total immersion with air pressure equivalent to the hydrostatic pressure head at the sternal notch, thus providing a minimal degree of pressure compensation. It is not surprising then that VC and ERV reductions were marginally less than the values cited in Table 3.2.

Static respiratory mechanics are ultimately dependent upon lung volume, due to the alinear nature of the three compliance curves. The current changes observed in lung volumes with altered breathing pressure during immersion, highlight the need to employ hydrostatic pressure compensation during upright immersion.

With air provided at P_{LC} during immersion, lung volume subdivisions and capacities returned towards control levels (Figures 3.10, 3.11). Beckman *et al.* (1961), and subsequently Thalmann *et al.* (1979), similarly observed positive pressure air supply (relative to hydrostatic pressure at the mouth) to return lung volumes to control values[41]. Current data, showing progressive increments in VC and ERV with elevated breathing pressure, qualitatively support the observations of Thalmann *et al.* (1979). It is postulated that lung volumes are restored because hydrostatic pressure compensation returns the normal transrespiratory pressure gradient to the lungs. Pulmonary blood will also be displaced back to the

[41] Thalmann *et al.* (1979) used three subjects during prone cycle exercise at 141 Watts.

peripheral venous pool. It is suggested that changes in pulmonary blood volume did not alter the elastic properties of the lung tissue in the current investigation, as evidenced by equivalence of the control and immersed compliance values. However, pulmonary blood may have influenced lung volumes. Alternatively, the total lung capacity may have been reduced due to hydrostatic compression of the thorax.

Fenn *et al.* (1947) first demonstrated the capacity of positive pressure ventilation to move blood from the thorax to the legs. Kilburn and Sieker (1960) demonstrated positive pressure lowered stroke volume, the cardiac index and central blood volume. More recently Orlov *et al.* (1985) found positive pressure ventilation during immersion reduced lung blood volumes, by shifting blood to the liver.

It is concluded that lung volumes, and consequently static respiratory mechanics, during immersion are best returned to normal levels when air supply pressure is modified to equate with a P_{LC} of 1.33kPa. The present observations therefore support the acceptance of hypothesis number two, and consequently indicate that underwater breathing apparatus should be modified to enable diver air supply at this pressure, when adopting an upright posture.

CHAPTER 4

RESPIRATORY DYNAMICS DURING WHOLE-BODY, UPRIGHT IMMERSION BREATHING AIR AT DIFFERENT STATIC RESPIRATORY LOADS.

Mechanically the respiratory apparatus may be considered a reciprocating bellows pump, (Mead and Milic-Emili 1964) driven by muscular contractions, transmitting forces via passive tissue elements to move air. According to Newton's third law of motion, such contractions will meet opposing internal forces, the magnitudes of which depend upon instantaneous position, velocity and acceleration of the respiratory apparatus and breathing gas.

Rohrer (1915, 1925) reported the first comprehensive analysis of respiratory dynamics. He identified elastic, flow-resistive and inertial forces during respiration. Morphometric analysis of autopsied airways enabled theoretical calculation of airway resistance from Poiseuille's Law [1], permitting separation of forces related to volume change (elasticity), and the rate of volume change (frictional resistance).

Rohrer's pupil, Wirz (1923), performed the first simultaneous measurements of pleural pressure (pneumothorax method) and airflow. Plotting pressure against volume at equivalent times provided a closed loop, as predicted by Rohrer (1925), the area of which described pressure changes necessary to overcome non-elastic (flow-resistive) respiratory forces. Subsequent investigations during spontaneous ventilation (Neergaard and Wirz 1927, Buytnedijk 1949, Mead and Whittenberger 1953, McIlroy *et al.* 1954, Margaria *et al.* 1960, Ballantine *et al.* 1970, Jonson and Olsson 1971, Field *et al.* 1984), artificial ventilation (Otis *et al.* 1950, Opie *et al.* 1959, Sharp *et al.* 1964b, Hedenstierna and McCarthy 1975, Zocchi *et al.* 1984), and spontaneous ventilation with gases of varied densities and kinematic viscosities (Bayliss and Robertson 1939, Fry *et al.* 1954, McIlroy *et al.* 1955) have substantiated Rohrer's observations, and have provided a basis for analysing respiratory dynamics.

Mechanical work of ventilation may be partitioned three ways:

(1) elastic work,

(2) flow-resistive work,

(3) inertial work (Figure 4.1).

The elastic work necessary to produce a static volume change against the resultant forces of the lungs and

---

[1] Pressure = $[\ 8 \cdot l \cdot u \cdot \dot{V}\ ] / \mathrm{Pi} \cdot r^4$
where: l = tube length, u = coefficient of viscosity, $\dot{V}$ = flow, r = tube radius.

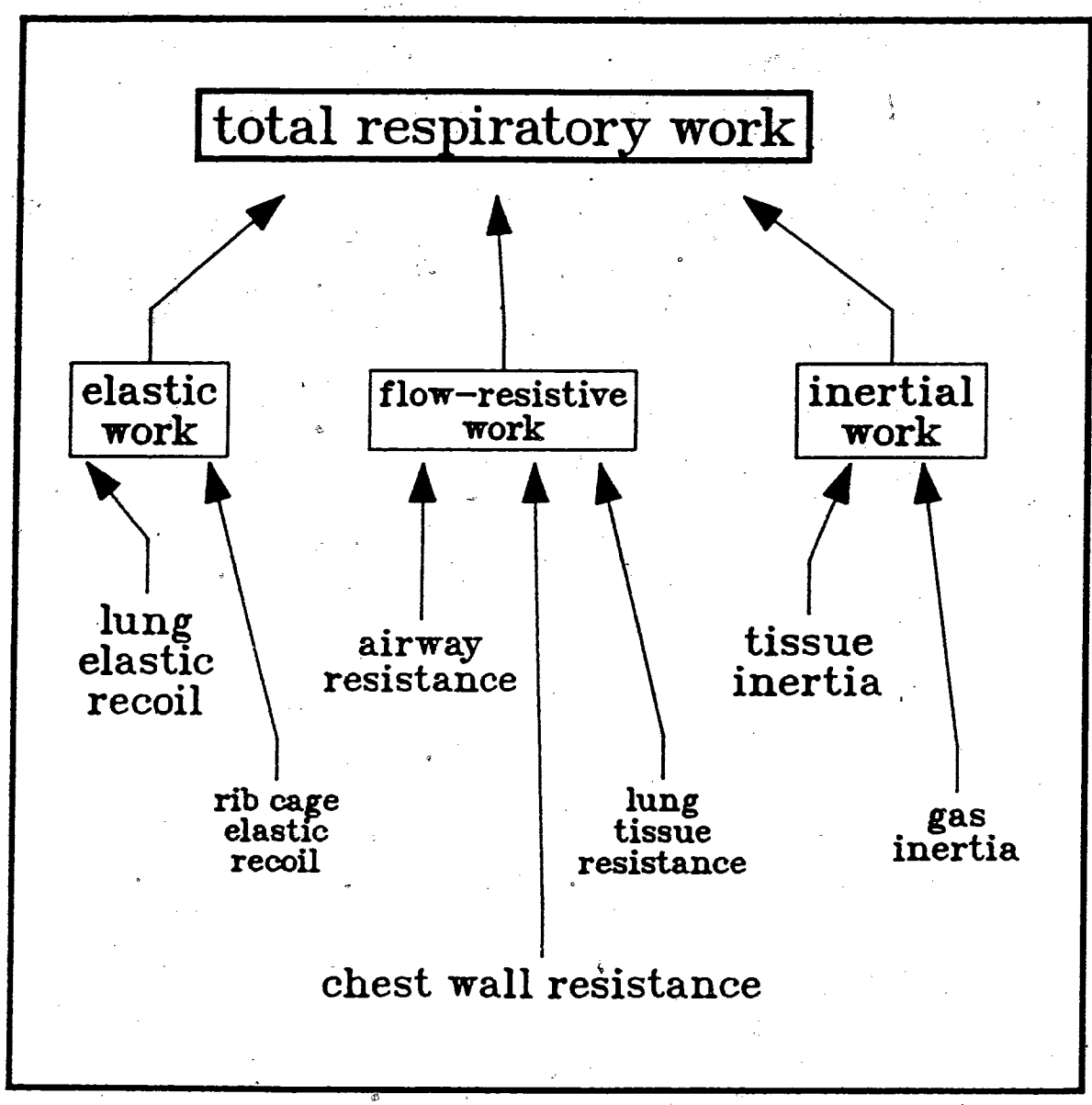


Figure 4.1: Mechanical work of ventilation and the forces from which it arises.

chest wall, increases as an approximate square function of V_T (Fenn 1951, McIlroy *et al.* 1954, Otis 1964). Flow-resistive work is expended against frictional forces in the airways and within lung and chest wall tissues [2] (*e.g.* tissue deformation). Inertial work is performed against air and body tissues, including the lungs, rib cage, diaphragm, and the abdomen, to initiate or change velocity. One may describe respiratory work as a function of lung volume (V) and its first ($\dot{V}$) and second time derivatives ($\ddot{V}$):

$$\text{Work} = f(V,\dot{V},\ddot{V})$$

Measurement of respiratory elastic work may be traced to Carson's (1820) initial analysis of lung elasticity, but is generally attributed to the independent works of Romanoff (1910–1911), Rohrer (1916) and Rahn *et al.* (1946). Opie *et al.* (1959) first attempted the division of flow-resistive work. Rohrer (1925) approximated inertance theoretically (though he neglected gas inertia (Mead 1956)), while DuBois (1953) and Mead (1956) subsequently provided its initial quantification. Table 4.1 illustrates the relative contributions to total respiratory work of the various sub-components. During resting ventilation about 60–70% of the total respiratory work is due to elastic forces, while flow-resistive forces account for the remaining work. Inertial forces represent negligible contributions (Rohrer 1925, DuBois 1953, Mead 1956, Sharp *et al.* 1964a, Dosman *et al.* 1975).

A consensus of available evidence indicates airway resistance (R_{aw}) is responsible for about 60–86% of flow-resistive work. However, the data of Opie *et al.* (1959) show the opposite trend (Table 4.1). Frictional losses due to lung tissue resistance (R_{lt}) reveal considerable variability. Only three studies could be located which employed techniques facilitating chest wall resistance (R_{wt}) measurement, of which only Ferris *et al.* (1964) partitioned work into three components. Thus, the relative R_{lt} and R_{wt} contributions remain incomplete.

Factors known to influence elastic work have been included in Table 3.1. Since R_{aw} accounts for 60–86% of flow-resistive work, it is apparent that the dynamic respiratory work is primarily regulated by the variables of flow mechanics. Since airways are short, elastic, curved, branched, of irregular cross-sectional geometry and possess uneven surfaces, they do not fulfil the requirements of *ideal tubes*. Flow patterns will vary with location and diameter. During quiet respiration, airflow in the trachea is

[2] Viscous resistance has been used to describe these forces, but this falsely implies viscosity is the only variable involved.

Table 4.1: **Percentage contribution of elastic and flow-resistive work components to total respiratory work as reported by previous investigators.**

SOURCE	ELASTIC WORK	RESISTIVE WORK				INERTIAL WORK		
	TOTAL	TOTAL	W_{aw}	W_{lt}	W_{wt}	TOTAL	gas	tissue
Otis *et al.* 1950	63%	37%	77.6%	22.3%	‡			
McIlroy *et al.* 1954	70%							
McIlroy *et al.* 1955				35%				
Marshall & DuBois 1956			86%	13.7%				
Attinger & Segal 1969		38%						
Opie *et al.* 1959			28.6%	71.4%	‡			
Ferris *et al.* 1964			60%	1%	39%			
Sharp *et al.* 1964a						†	84%	16%
Gautier *et al.* 1982			82%	18%				

‡ = combined lung tissue and chest wall resistance.

† = total inertance = 0.000918kPa.l^{-1}.$s.s^{-1}$.

Data under total columns show percentage of total respiratory work for each sub-component. Other columns show the percentage contribution of each component to each of the sub-totals.

partly laminar and partly turbulent [3] (Cotes 1979), and accounts for approximately 80% of R_{aw} (Hyatt and Wilcox 1961, Ferris *et al.* 1964, Macklem and Mead 1967, Hogg *et al.* 1968, Olsen *et al.* 1970). From the fourth to twelfth generation, flow is laminar, while beyond this point flow gradually becomes diffusive (Bouhuys 1977), due to the parallel arrangement of peripheral airways [4]. Such a flow arrangement of flow patterns has two implications: (1) it is not possible to assign a *single* number to resistance in a complex system of branching elastic airways; and (2) modifications to R_{aw} (*e.g.* during immersion) occur primarily in the larger airways, with contributions of the smaller airways becoming important with large lung dimensional changes.

Resistance will rise with higher flow rates (when turbulent flow is present), reduced airway diameter and lengthened airways. Briscoe and DuBois (1958) demonstrated an inverse relation between lung volume and R_{aw}. However, while lung volume increments widen, they also lengthen the airways. These changes have an opposite effect upon R_{aw}, the net result of which is a reduction in R_{aw}.

Flow-resistive work and power increases curvilinearly with minute ventilation ($\dot{V}_I$, Figure 4.2), however, variability between different studies is extensive, particularly as ventilation rises. Data from Otis *et al.* (1950) and Holmgren *et al.* (1973) indicate that flow-resistive work is best described as a power function of minute ventilation (r = 0.97 and 0.99 respectively). McIlroy *et al.* (1954) found a linear relation (r = 0.99), and Fritts *et al.* (1959) an exponential function (r = 0.99).

Upright immersion imposes a hydrostatic pressure imbalance on the lungs (relative to external thoracic pressure), which is physically analogous to negative pressure breathing. Hong *et al.* (1969) first studied respiratory work ($\dot{N}$ = 4) under these conditions using head-out immersion. Comparisons between 'xiphoid' and 'neck' immersions revealed a 107.2% rise in flow-resistive inspiratory work, and a 57.4% elevation in total elastic work; producing a reported 64.6% increment in total inspiratory work [5].

---

[3] Reynolds number is around 1500 (Cotes 1979).

[4] Weibel (1963) found cross-sectional area to decrease from 2.54cm^2 to 2.00cm^2 at the third generation, then to increase by a factor of about 1.4 per generation to 180cm^2 at the sixteenth generation. At this point there were about 65000 airways.

[5] Methodological description in these experiments lacks detail pertaining to dynamic pressure-volume measurements. Oesophageal balloons were used to measure static gastric pressures. One has to assume dynamic work was derived using transpulmonary pressure (P_{tp}) measurements.

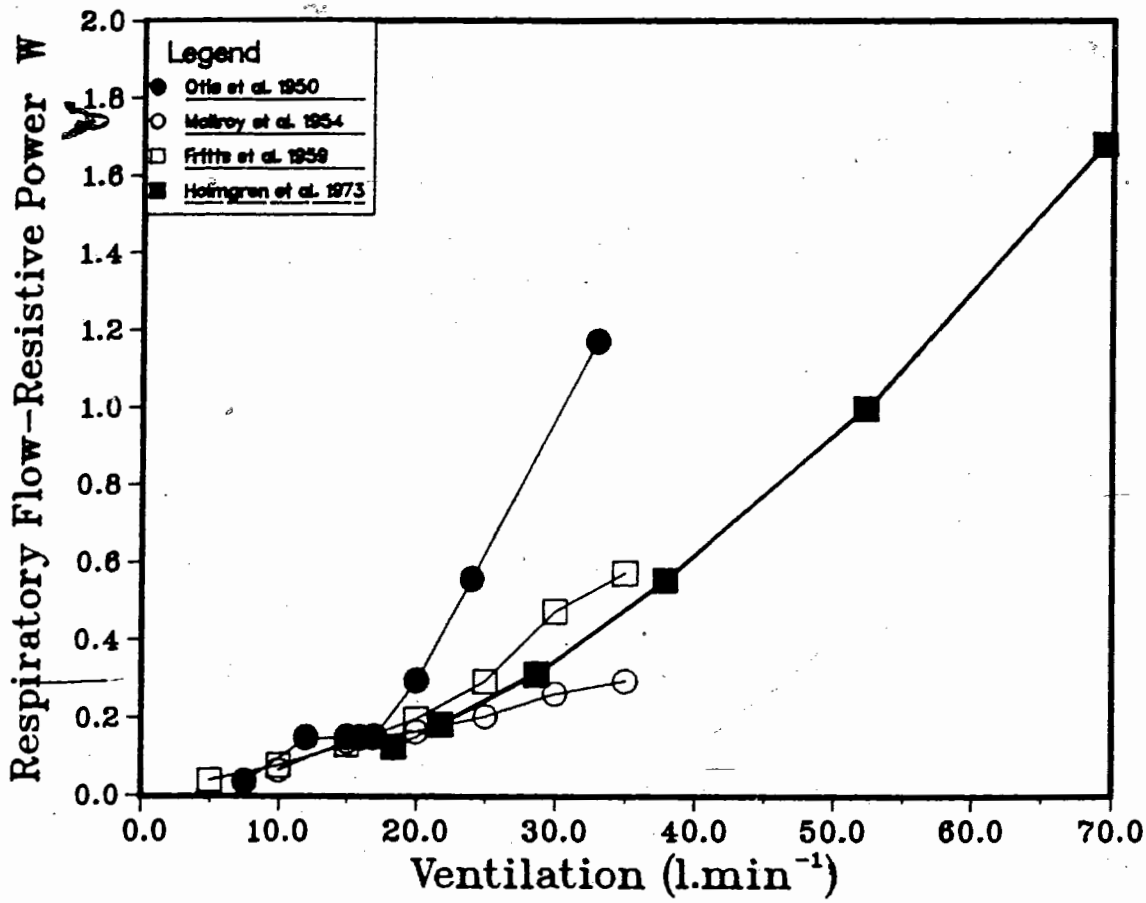


Figure 4.2: **Flow-resistive power of breathing with increased minute ventilation.** [Data obtained from: Otis *et al.* (1950), McIlroy *et al.* (1954), Fritts *et al.* (1959), Holmgren *et al.* (1973). Note: only data from Otis *et al.* (1950) includes chest wall resistance].

Agostoni *et al.* (1966) found 'neck' immersion increased R_{aw} 57.7% (interrupter technique), which the authors attributed to a reduction in functional residual capacity (FRC), mediated by a positive displacement of the static pressure–volume curve. Subsequent work by other groups has produced resistance increases during immersion of varying magnitudes (Table 4.2). Prefaut *et al.* (1976) observed no change in upstream R_{aw}[6] during upright immersion, which was interpreted as implying that airways upstream of the equal pressure point[7] were unaffected by immersion induced physiological changes. Resistance elevations during immersion (Table 4.2) must have been associated with downstream airway changes, possibly accompanying changes in lung volume[8].

Sterk (1970, 1973) analysed dynamic P_{tp} and airflow of divers wearing wet suits and semi-closed SCUBA. Flow-resistive work increased by 370% and 321% respectively, while elastic work showed respective increments of 115% and 131%. Sterk derived elastic work using only the lung tissue compliance curve. Since only P_{tp} was measured, elastic work did not make allowance for the participation of the chest wall. A large percentage of the flow-resistive gains may be attributed to extrinsically imposed work, associated with the breathing apparatus and diving apparel. External flow-resistive work presented by the SCUBA was beyond the recommended ideal limits (Morrison and Reimers 1982), confounding data interpretation with respect to immersion effects on dynamic respiratory mechanics. Thus, it is not possible to assign a quantity to the respiratory loading incurred by immersion alone. Similarly, flow-resistance changes (Table 4.2) may have been associated with thoracic constriction imposed by the diving equipment. Vital capacity (VC) in the latter study approximated 75% of dry controls, indicating restriction beyond that associated with immersion alone[9].

At present only one investigation has been located within the literature, in which dynamic respiratory work has been studied in isolation of breathing apparatus effects, during immersion (Hong *et al.* 1969). No material is presently available on dynamic changes associated with hydrostatic pressure compensation during immersion.

---

[6] Upstream airway resistance = static recoil pressure / maximal expiratory flow.

[7] The point of dynamic airway compression.

[8] The measurement of upstream R_{aw} is only an approximation of R_{aw}.

[9] Immersion normally reduces VC 5–10% (Table 3.2).

Table 4.2: **Changes in respiratory flow-resistance induced by upright immersion. Evidence from the literature.**

SOURCE	METHOD	R_{aw}	R_{pul}	$R_{f(pul)}$	Total
Agostoni *et al.* 1966	interrupter	57.7%			
Sterk 1970	dynamic P_{tp}			243.5%	
Sterk 1973	dynamic P_{tp}			185.4%	
Dahlbäck 1978	dynamic P_{oes}			31%	
Dahlbäck *et al.* 1979	dynamic P_{oes}			42.5%	
Löllgen *et al.* 1980	oscillation				57.4%

All percentages represent increments.
P_{oes} = oesophageal pressure, P_{tp} = mouth pressure minus P_{oes}, R_{aw} = airway resistance, R_{pul} = pulmonary resistance (R_{aw} plus lung tissue), $R_{f(pul)}$ = methods averaging inspiratory and expiratory R_{pul}, Total = R_{aw} + R_{pul} + chest wall resistance.

4.0.5 Purpose of the chapter.

The work performed in this chapter was directed towards analysis of pulmonary dynamics during upright, whole–body immersion with, and without hydrostatic pressure compensation. Of particular interest was the evaluation of the ability of breathing pressure compensation to return dynamic variables to levels which exist in air.

4.1 METHODS

4.1.1 Subjects

Ten male non-smokers (including eight divers), screened by questionnaire for normal lung function history participated as subjects. All subjects took part in experiments designed to characterise static respiratory attributes during immersion (Chapter 3). All received subject information packages and signed informed consent releases.

4.1.2 Apparatus

The apparatus used in this investigation was identical to that used in the previous chapter. Details of instrument calibration and testing are also described above (see: Sections 3.3.2 and 3.3.3).

The phase shift between the flow and transpulmonary pressure signals was minimised by adjusting the lengths of pressure probes to the pneumotachograph and transpulmonary pressure transducers. Data calculations incorporated a further phase shift allowance, to account for incomplete removal of phase distortion. The magnitude of the latter was evaluated using simultaneous application of an approximately sinusoidal air flow impulse to both the pneumotachograph and the oesophageal catheter.

4.1.3 Procedures

Subjects were required to perform spontaneous respiration using self-determined tidal volumes and timing patterns. All trials were completed with subjects seated upright and the hips flexed at 90°. Control experiments were performed in the laboratory using the apparatus modifications described above (Section 3.3.2).

Transpulmonary pressure (P_{tp}) was measured as the difference between oesophageal pressure (P_{oes}) and pressure at the mouth (P_{ao}), using an oesophageal balloon. Oesophageal balloons for each subject were positioned 40cm beyond the nares using the procedures detailed in Section 3.3.4. Balloon volume was verified between trials by having subjects perform valsalva manoeuvres. Oesophageal pressure at end-tidal lung volumes averaged −0.42kPa (SEM = 0.04), prior to commencing trials, and −0.41kPa (SEM = 0.05), immediately before catheter removal. From these observations it was deemed that balloons had not moved significantly during trials.

During submerged trials subjects wore a pressure compensated diving hood (Figure 2.4). Two demand regulators provided air (Figure 2.3). Regulators were covertly and randomly positioned at each of four depths: mouth level, 13.5cm below the sternal notch (mean lung *centroid* position, as determined in Chapter Two), and 10cm above and below the mean lung *centroid locus*. Between trials subjects breathed from regulator 2, to avoid CO_2 accumulation while submersed. The airway to regulator 1 was flushed with fresh air after each manoeuvre. Before commencing immersion trials at different delivery pressures, subjects breathed for about one minute, at the new air delivery pressure, to ensure respiratory mechanical adjustments were complete (Ferris and Pollard 1960).

4.1.4 Calculations

An analysis programme was written to calculate: respiratory timing, tidal volumes (V_T), dynamic lung tissue compliance ($C_{dyn(l)}$), pulmonary flow-resistive work (W_{pul}), pulmonary resistance (R_{pul}), and functional pulmonary resistance ($R_{f(pul)}$). Calculations were performed on each breath with data for trials averaged over multiple respiratory cycles.

Respiratory timing.

Algorithms defined end inspiratory and end expiratory flow points, and permitted data sectioning into regions containing complete respiratory cycles. Inspiration and expiration onset and termination were identified by detection of a voltage change at the pressure transducer channel measuring respiratory flow rate through the pneumotachograph. Unpaired inspirations or expirations, incomplete phases and false V_Ts [10] were not analysed. Identification of these points permitted computation of inspiratory (T_I), expiratory (T_E) and total cycle durations (T_{TOT}), and also the identification of marker points for subsequent computations.

Tidal volumes.

Inspired volumes were derived from the integration of flow with respect to time.

$$V_I = \int \dot{V}\, dt \qquad \textbf{Equation 1}$$

Integrated volumes were corrected to BTPS, by adjusting for regulator pressure variations, and

[10] Tidal volumes less than 200ml.

temperature and saturation differences between the inspirate and expirate.

$$V_I = V_I \cdot (P_A + \text{reg.pressure}) / (P_A) \qquad \textbf{Equation 2}$$

$$V_I BTPS = \frac{V_I \cdot 310 (P_A + P_{LC} - P_{H_2O})}{(273 + T_A)(P_A + P_{LC} - 47.1)} \qquad \textbf{Equation 3}$$

$$f_b = 60 / T_{TOT}(\text{sec}) \qquad \textbf{Equation 4}$$

$$\dot{V}_I = f_b \cdot V_I \qquad \textbf{Equation 5}$$

Dynamic lung tissue compliance.

Dynamic pulmonary (lung tissue) compliance ($C_{dyn(l)}$) was obtained from the ratio of lung volume change to transpulmonary pressure (P_{tp}) change, between points of zero gas flow at the end of expiration and inspiration, after the technique developed by Neergaard and Wirz (1927), Bayliss *et al.* (1939), and Mead and Whittenberger (1953). Assumptions inherent within the technique include: cessation of gas flow at mouth and alveoli; observed P_{tp} represents forces applied to overcome elastic properties only; and compliance between points of zero flow is linear.

Pulmonary work.

Work expended to overcome airway and lung tissue resistance (W_{pul}) was derived by integration of P_{tp} with respect to volume.

$$W_{pul} = \int P_{tp} \cdot dV \qquad \textbf{Equation 6}$$

Thus: $$W_{pul} = \int P_{tp} \cdot \dot{V} \cdot dt \qquad \textbf{Equation 7}$$

Four possible work subdivisions exist: (1) positive inspiratory; (2) positive expiratory; (3) negative inspiratory; and (4) negative expiratory work. Figure 4.3 illustrates that flow-resistive W_{pul} (represented by the area within the pressure-volume loop) may be derived from the integration of flow-resistive and

elastic work over a complete respiratory cycle[11]. Inspiratory and expiratory elastic work during this cycle are of opposite sign and of equal magnitude, unless lung volumes at the start and end of a breathing cycle are unequal.

Thus referring to Figure 4.3:

$$W_{pul} = (WIP - WEN) + (WEP - WIN) \qquad \textbf{Equation 8}$$

where:

WIP = positive inspiratory work (area HGDC).

WEN = negative expiratory work (area DCI).

WEP = positive expiratory work (area IFEA).

WIN = negative inspiratory work (area EAH).

Flow-resistive inspiratory and expiratory work ($W_{pul(insp)}$ and $W_{pul(exp)}$ respectively) were calculated. When end expiratory P_{tp} was zero or greater, inspiratory work ($W_{pul(insp)}$) equalled positive inspiratory work minus elastic work [12]. When subjects breathed at delivery pressures equal to the hydrostatic pressure at the mouth, end expiratory P_{tp} was negative, producing the situation in Figure 4.3. Here $W_{pul(insp)}$ again equals inspiratory work minus elastic work. Thus, inspiratory work equals area HGDC (positive inspiratory work) minus area BCD, plus area AEB minus area EAH (negative inspiratory work). Expiratory work was equal to area BCD minus area DCI, plus area IFEA minus area AEB. Total work was then obtained by addition of the inspiratory and expiratory areas.

Since tidal volumes were not of equal magnitude, flow-resistive work was divided by V_T to obtain work in $J.l^{-1}$ or kPa.

---

[11] Zero volume respresent the RV.

[12] Since $C_{dyn(l)}$ is assumed to be linear, elastic work equals the area of the triangle, or trapezium, created between the zero P_{tp} axis and the compliance cord.

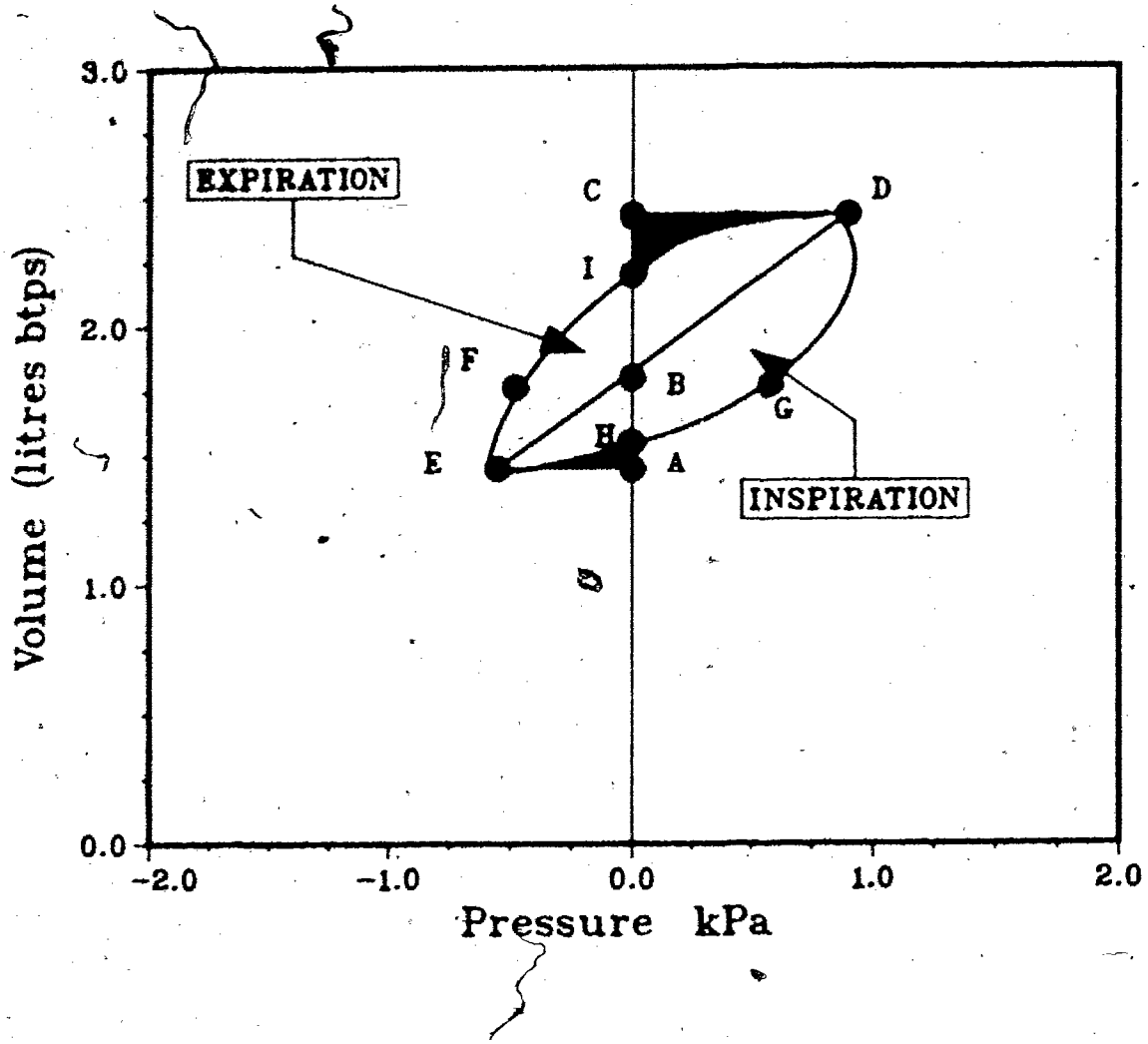


Figure 4.3: Partitions of pulmonary flow-resistive work using transpulmonary pressure–volume loops. [See text for a description.]

Pulmonary resistance.

Mechanical resistance to lung tissue movement and airflow was computed at 20msec intervals, after the subtractive technique initiated by Mead and Whittenberger (1953). Instantaneous P_{tp} may be attributed to elastic (volume related), resistive (flow related) and inertial (acceleration related) forces.

$$P_{tp} = f(V) + f(\dot{V}) + f(\ddot{V}) \qquad \textbf{Equation 9}$$

Inertial forces are negligible and ignored at normal gas densities, normal thoracic mass, at breathing frequencies less than 60b.min^{-1}, and during normobaric immersion (Mead and Whittenberger 1953, DuBois *et al.* 1956, Mead 1956, McKenna *et al.* 1973, Dosman *et al.* 1975). If P_{tp} attributable to lung volume (*i.e.* elastance) is known (i.e. $P = f(V)$), then the components of instantaneous P_{tp} responsible for overcoming pulmonary flow-resistive forces may be solved (i.e. $P = f(\dot{V}) = P_{fr}$):

$$P_{fr} = P_{tp(dyn)} - P_o - (dV/C_{dyn(l)}) \qquad \textbf{Equation 10}$$

where:

$P_{tp(dyn)}$ = dynamic P_{tp} during breathing,

dV = lung volume minus end-expiratory volume,

$C_{dyn(l)}$ = dynamic pulmonary compliance,

P_o = end-expiratory P_{tp}, and

P_{fr} = flow-resistive pressure.

It was assumed that $C_{dyn(l)}$ was linear over the V_T studied. Since instantaneous data exist for volume, $P_{tp(dyn)}$ and flow, one may compute instantaneous pulmonary resistance (R_{pul}):

$$R_{pul} = P_{fr}/\dot{V} \qquad \textbf{Equation 11}$$

Pulmonary resistance was measured using two methods. First, inspiratory and expiratory pulmonary resistances were measured over flow rates between 0.4 to 0.6 l.s^{-1}. These data were separately averaged and expressed as inspiratory, expiratory and total breath R_{pul}. Second, R_{pul} was averaged over all flow rates to yield mean inspiratory, expiratory and total breath R_{pul}. Units of measurement:

$kPa.l^{-1}.sec$.

Functional pulmonary resistance.

The concept of functional pulmonary resistance ($R_{f(pul)}$) was developed by Ahlström and Jonson (1974), and Jansson and Jonson (1975) based upon the assumptions that the lung may be modelled from sinusoidal flow (Jaeger and Otis 1964) with constant airway and lung tissue resistance. Using these assumptions pulmonary flow–resistive power could be obtained if minute ventilation and $R_{f(pul)}$ were known.

$$\dot{W}_{pul} = R_{f(pul)} \cdot \dot{V}_I^2 \cdot Pi^2 \cdot k/2 \qquad \textbf{Equation 12}$$

where:

$\dot{W}_{pul}$ = pulmonary power for a given breath (Watts),

$R_{f(pul)}$ = functional pulmonary resistance ($cmH_2O.l^{-1}.s$),

$\dot{V}_I^2 = (V_I \cdot f_b)^2$ ($l.min^{-1}$),

$Pi^2 = 9.8696$, and

k = common units conversion constant.

Since $R_{f(pul)}$ was an unknown, it could be obtained if $\dot{W}_{pul}$ was measured. [Equation 12 was taken from that developed by Ahlström and Jonson (1974) and Jansson and Jonson (1975)]. Thus:

$$R_{f(pul)} = (\dot{W}_{pul} \cdot 2)/(\dot{V}_I^2 \cdot Pi^2 \cdot k) \qquad \textbf{Equation 13}$$

$R_{f(pul)}$ calculations using this equation were incorporated to permit comparison with the work of Dahlbäck *et al.* (1979).

4.1.5 Technique delimitations.

Measurement of P_{tp} using oesophageal balloons depends upon the fidelity with which pleural pressure (P_{pl}) can be approximated by P_{oes}. P_{pl} gradients occur along both the sagittal (Mead 1961, Milic–Emili *et al.* 1964b, 1966, D'Angelo *et al.* 1970, Agostoni *et al.* 1970) and transverse axes (Agostoni and D'Angelo 1969, Miserocchi *et al.* 1981), thus P_{oes} is considered to reflect mean pleural surface pressure rather than to be an index of local P_{pl}. Transpulmonary pressure change attributable to random

compression artifact during the respiratory cycle, remains unknown (*e.g.* oesophageal contractions, cardiogenic compression).

Measurements of W_{pul} and R_{pul} do not include flow-resistive forces attributable to chest wall distortion (Opie *et al.* 1959, Otis 1964, Agostoni 1970, Goldman and Mead 1973, Goldman *et al.* 1976), which, during exercise may contribute up to 25% of total respiratory work (Goldman *et al.* 1976). Contributions of chest wall dynamics remain unknown in the current investigation.

Measurement of W_{pul} is subject to underestimation in cases where subjects do not inspire and expire identical volumes of air. Since integration procedures were triggered from flow changes measured at the pneumotachograph pressure transducer, a mid-inspiratory or expiratory pause or airflow reduction could result in termination of the integration process. This would occur if airflow fell below a predetermined basal limit. A recommencement of inspiration or expiration would then not be included in the integration, and flow-resistive work underestimation would result. Instances of this occurrence on expiration were observed when subjects breathed air at P_{LC} plus 0.98kPa.

Similarly, transient flow reductions to basal levels would confound R_{pul} measured from resistance averaging over all flow rates. In this case flow would be almost absent while P_{tp} would be at some negative or positive value close to zero. Division of the resultant flow-resistive P_{tp} by flow, could produce very large values, which may have the opposite sign for the part of the breaths for which they were to be added. Such data points may also be present at the start and end of each respiratory phase. Inclusion of such resistance values would falsely alter the mean R_{pul}.

4.1.6 Analysis

Analysis was based upon a repeated measures experimental design, with five levels of one within subjects factor (*i.e.* air delivery pressure). Subjects took part in all phases of the investigation. *A priori* probability significance was set at the 0.05 level. Three statistical analyses were performed: (1) paired or correlated t-tests for comparisons between group means on single variables; (2) Hotelling's T-squared analysis for simultaneous comparisons between univariate factors for several variables; and (3) multivariate analysis of variance for analysis over several levels of the treatment effect. *Post hoc* multiple comparison analyses (Tukey's HSD test) were employed to locate sources of significant variance obtained from MANOVA[13]. Where significant changes were not observed, computations of statistical power were

[13] Summary tables are contained within Appendix Three.

performed. Statistical power (ø) is defined as the probability of correctly accepting an experimental hypothesis (Keppel 1973, Gehring 1978).

4.2 RESULTS

4.2.1 Characteristics of subjects

Physical characteristics of subjects are detailed in Table 3.3. Subjects in this study took part in the above investigations of static mechanical attributes (Chapter Three) [14].

4.2.2 Pulmonary work of breathing.

Under control states, spontaneous breathing demanded a mean pulmonary power ($\dot{W}_{pul}$) to overcome flow-resistive forces, of 0.060 ± 0.026W. Ventilatory increments produced an exponential elevation in $\dot{W}_{pul}$ (r = 0.9939) as shown in Figure 4.4. During control trials, relative pulmonary flow-resistive work of spontaneous breathing (W_{pul}) [15] averaged 0.219kPa (SEM=0.019, Table 4.3 [16]). Flow-resistive pulmonary work (W_{pul}), in excess of that provided by passive recoil of the lungs [17], averaged 0.078kPa (SEM=0.015). In all subjects, control pulmonary elastic work ($W_{E(pul)}$), during spontaneous ventilation, exceeded expiratory flow-resistive work (Table 4.3) indicating ventilatory patterns were normal (Agostoni *et al.* 1970), and expiration was essentially a passive act, or expiration was braked by maintenance of a degree of inspiratory tone during expiration.

In all subjects control expiratory flow-resistive work ($W_{pul(exp)}$) exceeded inspiratory work ($W_{pul(insp)}$). The mean, control inspiratory:expiratory work ratio during spontaneous ventilation was 0.538 [18] demonstrating greater expiratory flow-resistance.

Spontaneous breathing during total immersion, breathing air at mouth pressure[19], elevated total, flow-resistive W_{pul} by a factor of 3.8 (Figure 4.5, $p<0.05$[20]) to 0.75kPa.

---

[14] Subject numbers used in Tables 3.3 and this chapter are identical, and remain consistent throughout the chapter.

[15] Relative W_{pul} represents work per litre of gas inspired. Units = $J.l^{-1}$, which are equivalent to units of pressure; thus, the kPa is used for relative W_{pul}.

[16] Standard error of the mean.

[17] Integration of the pressure-volume loop falling outside the zone of lung elastic recoil (see: Figure 4.3: areas BCD and BAE).

[18] Inspiratory:expiratory ratio = $W_{pul(insp)}/W_{pul(exp)}$ (dimensionless).

[19] The demand regulator was positioned in the same horizontal plane as the mouth.

[20] Overall MANOVA F = 26.5 (3,6), p=0.007.

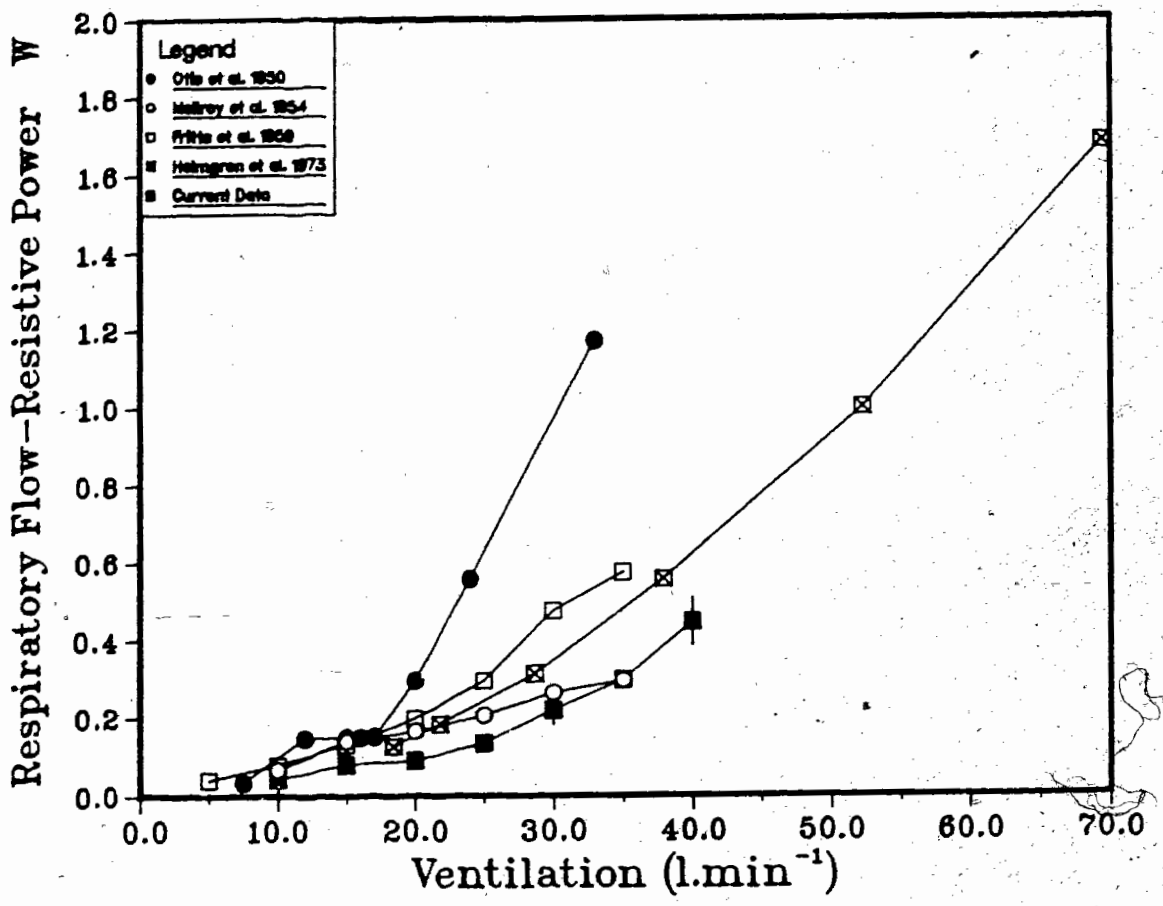


Figure 4.4: **Absolute pulmonary power as a function of ventilation.** [A comparison of current observations on subjects resting in air, with data from the literature. Subjects were studied during spontaneous ventilation and during trials where tidal volume was kept constant, but frequency was increased by following experimenter-generated timing patterns.]

Table 4.3: **Flow-resistive and pulmonary elastic work subdivisions during spontaneous ventilation in air.**

SUBJECT	FLOW-RESISTIVE WORK			PULMONARY ELASTIC WORK
	total	insp.	expir.	
1	0.274	0.126	0.148	1.102
2	0.139	0.057	0.083	0.390
3	0.242	0.108	0.134	0.638
4	0.165	0.054	0.111	0.923
5	0.177	0.020	0.158	1.625
6	0.114	0.033	0.081	0.251
7	0.159	0.055	0.103	0.562
8	0.254	0.077	0.178	0.680
10	0.306	0.112	0.194	0.652
$\bar{x}$	0.203	0.071	0.132	0.758
SEM	0.022	0.012	0.014	0.137

All units = kPa. (Work was calculated as $J.l^{-1}$ and expressed in kPa units).
Insp. = inspiratory flow-resistive work, Expir. = expiratory flow-resistive work.
Control data for subject nine was rejected due to artifactual error in pulmonary resistance data obtained at an airflow of 0.5 $l.s^{-1}$.

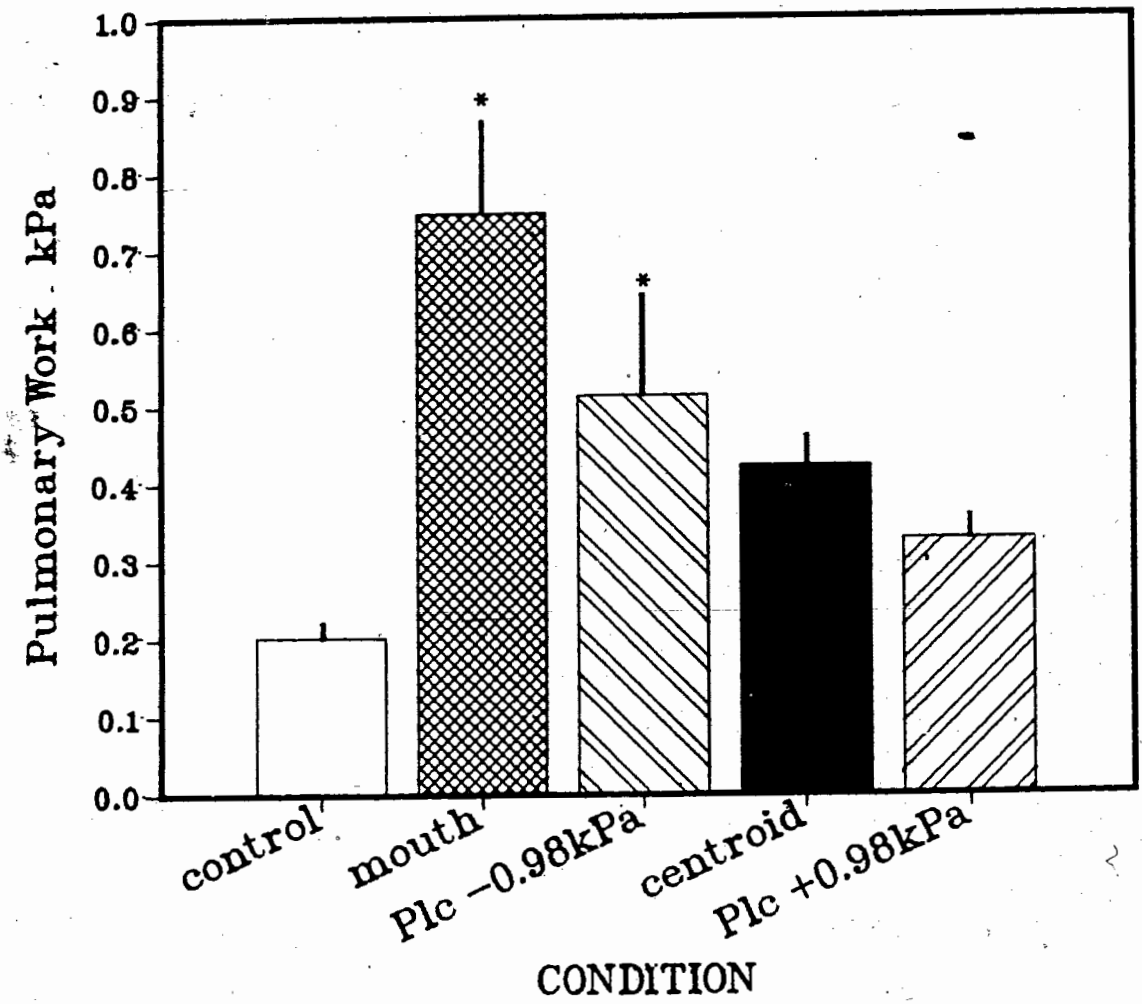


Figure 4.5: **Total flow-resistive pulmonary work during spontaneous breathing under control and immersion states, breathing air at four hydrostatic pressure loads.** [Work was calculated in $J.l^{-1}$ and is expressed in kPa units. Data represent means and standard errors. *=significantly different from control status ($p<0.05$).]

The effect of altered hydrostatic breathing pressure upon W_{pul} during immersion has not been reported previously. The present study reveals, in support of hypothesis three, that increasing air delivery pressure (relative to the mouth) returns total flow-resistive W_{pul} towards control levels (Figure 4.5). The use of hydrostatic pressure compensation, in the form of breathing pressure increments, sequentially reduced total flow-resistive W_{pul}. When air was delivered at P_{LC}, and above, total flow-resistive respiratory work was significantly less than observed in the uncompensated state ($p<0.05$), but remained greater than control levels ($p>0.05$, $\phi > 0.99$). [Appendix Three contains *post hoc* statistical summary.]

Analyses of inspiratory ($W_{pul(insp)}$) and expiratory ($W_{pul(exp)}$) flow-resistive work revealed immersion increased both components at all air delivery pressures (Figure 4.6). Flow-resistive work increased from 0.07 and 0.13kPa during control trials, to 0.29 and 0.46kPa during uncompensated immersion, for $W_{pul(insp)}$ and $W_{pul(exp)}$ respectively. Differences were significant ($p<0.05$), in both instances.

Expiratory flow-resistive work was reduced to levels not significantly greater than control values, with each of the three levels of breathing pressure compensation. Compensation pressures similarly resulted in significant reductions of $W_{pul(insp)}$, from values obtained during uncompensated immersion ($p<0.05$).

The control ratio of inspiratory to expiratory respiratory work was disturbed during immersion. Sequential increments in air delivery pressure produced inspiratory:expiratory ratios of 0.62, 0.47, 0.67 and 0.99 respectively. The results from the flow-resistive pulmonary work analysis are summarised in Table 4.4.

The purpose of this section was to determine whether air delivery pressures could be employed to offset the adverse respiratory mechanical changes attending whole-body, upright immersion reported in the literature (Hong *et al.* 1969, Sterk 1970, 1973). It appears that no single air delivery pressure provided the means to return all mechanical attributes to control status.

(1) P_{LC} plus 0.98kPa returned minimal total and expiratory flow-resistive work;

(2) P_{LC}, P_{LC} minus 0.98kPa and P_{LC} plus 0.98kPa minimised inspiratory flow-resistive work; and

(3) P_{LC} minus 0.98kPa best returned inspiratory:expiratory flow-resistive work ratio to control status.

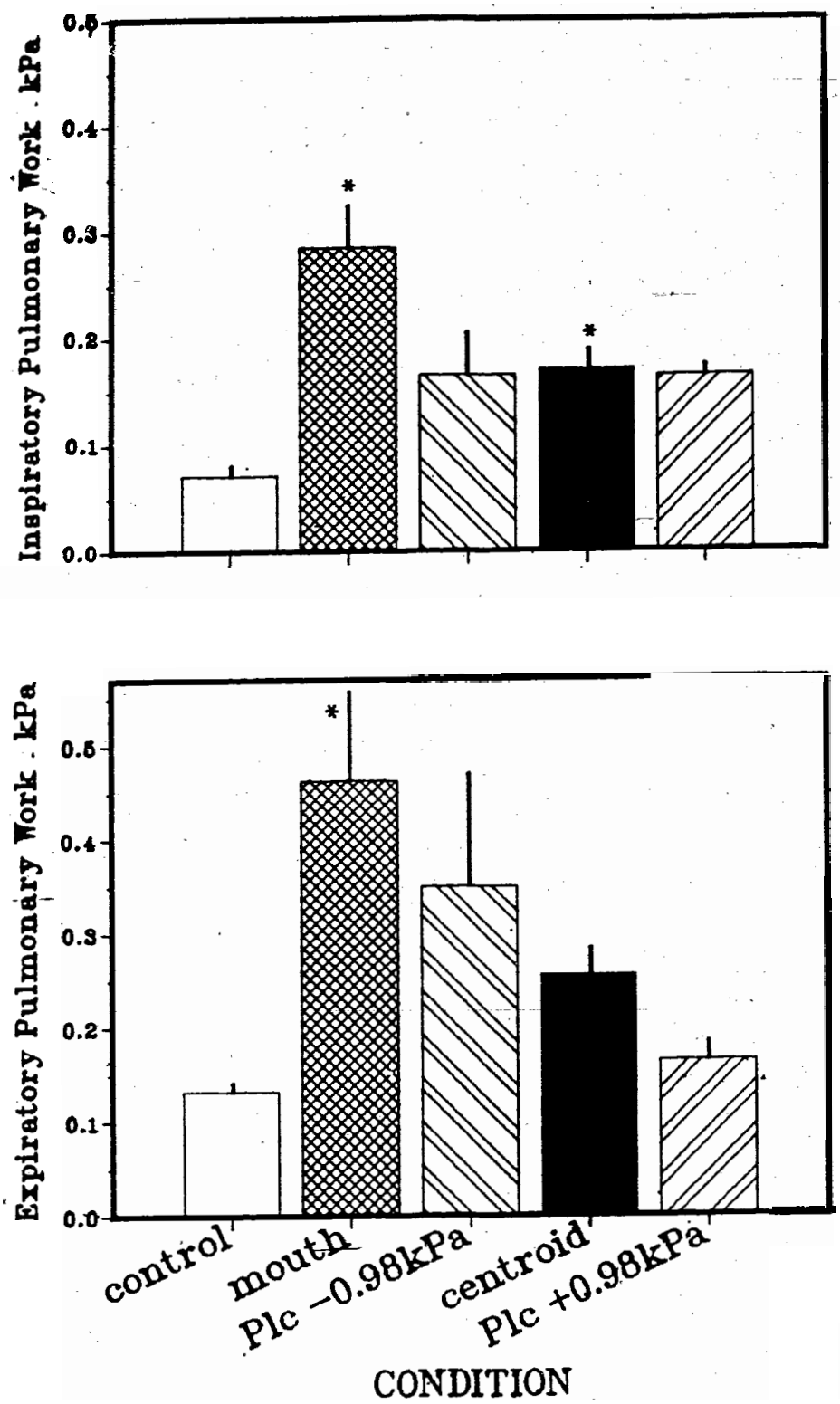


Figure 4.6: **Inspiratory and expiratory flow-resistive work during spontaneous ventilation under control and immersion states, breathing air at four hydrostatic pressure levels.** [Work was calculated in $J.l^{-1}$ and is expressed in kPa units. Data represent means and standard errors. *=significantly different from control status ($p<0.05$).]

Table 4.4: **Summary of flow-resistive pulmonary work partitions during control and experimental conditions at four air delivery pressures.**

CONDITION	FLOW-RESISTIVE WORK		
	insp.	expir.	total
control	0.071 (0.012)	0.127 (0.012)	0.197 (0.020)
mouth	0.285 (0.037)	0.462 (0.095)	0.748 (0.120)
P_{LC}−0.98	0.165 (0.039)	0.349 (0.120)	0.513 (0.133)
centroid	0.170 (0.019)	0.254 (0.034)	0.421 (0.042)
P_{LC}+0.98	0.163 (0.013)	0.164 (0.022)	0.327 (0.032)

All units = kPa. (Work was calculated as $J.l^{-1}$ and expressed in kPa units).
Means with standard error in parenthesis.
Conditions: control = upright in air; all others represent upright immersion breathing air at mouth pressure and at three pressures relative to *lung centroid* pressure P_{LC}.
Insp. = inspiratory flow-resistive work, expir. = expiratory flow-resistive work.

4.2.3 Pulmonary resistance.

Pulmonary and airway resistances are normally averaged from inspiratory data at a flow rate of 0.5 $l.s^{-1}$. Spontaneous ventilation, under control states, yielded a constant flow, inspiratory pulmonary resistance of 0.125kPa.l^{-1}.s (SEM = 0.022, Table 4.5). Expiratory R_{pul} exceeded mean inspiratory R_{pul} in six subjects.

Spontaneous breathing during total immersion, breathing air at mouth pressure, elevated inspiratory, expiratory and total R_{pul} (at 0.5 $l.s^{-1}$) by a factor greater than two, to produce respective resistances of 0.28, 0.56 and 0.44kPa.l^{-1}.s (Figures 4.7, 4.8, $p<0.05$[21]). Elevated air delivery pressures systematically returned resistances towards control status (Figures 4.7, 4.8). Air delivery at P_{LC} plus 0.98kPa best replicated control total, expiratory and inspiratory R_{pul}. Differences between values obtained at P_{LC} and P_{LC} plus 0.98kPa were non-significant ($p>0.05$, $\phi>0.99$), for each of the resistance measurements. [Appendix Three contains *post hoc* analysis summary.]

When analysed collectively across air delivery pressure during immersion, the expiratory R_{pul}, at a flow rate of 0.5 $l.s^{-1}$, was significantly greater than inspiratory R_{pul} ($p<0.05$). This trend is also reflected in the W_{pul} partitions in Figure 4.6. Expiratory R_{pul} and W_{pul} appeared more sensitive to changes in breathing pressure, when subjects were immersed.

Pulmonary resistance was also computed by averaging inspiratory, expiratory and total breath resistances over all flow rates. Mean inspiratory R_{pul} was lower in the control trials. Immersion, without hydrostatic breathing pressure compensation, increased mean inspiratory R_{pul} about 3.2 times to 0.29kPa.l^{-1}.s (Figure 4.9), compared with the 2.3-fold elevation recorded at a flow rate of 0.5 $l.s^{-1}$ (Figure 4.8). Expiratory R_{pul} obtained from all respiratory flow rates, similarly increased more than the corresponding change observed at a constant flow (3.2 versus 2.6 respectively, Figure 4.8), when uncompensated immersion was studied. Breathing pressure compensation lowered the mean expiratory R_{pul} to be non-significantly different from the control value ($p>0.05$, $\phi<0.30$). A sequential R_{pul} reduction with the breathing pressure increment was not as obvious for either inspiration or expiration, as that obtained at a constant flow (Figure 4.8).

The mean R_{pul}, averaged over the complete respiratory cycle (at all flow rates) reflected a weighted sum of the mean inspiratory and expiratory data. Changes observed at mouth pressure air

[21]Overall MANOVA $F = 26.08$ (7,2) $p = 0.037$.

Table 4.5: **Inspiratory, expiratory and mean total pulmonary resistance during spontaneous ventilation in air, calculated at a mean flow rate of 0.5 l.s^{-1}.**

SUBJECT	INSPIRATORY	EXPIRATORY	MEAN TOTAL
1	0.127	0.063	0.085
2	0.089	0.233	0.182
3	0.107	0.174	0.143
4	0.211	0.165	0.179
5	0.030	0.242	0.229
6	0.084	0.258	0.169
7	0.103	0.127	0.123
8	0.162	0.137	0.151
10	0.245	0.415	0.348
$\bar{x}$	0.129	0.202	0.179
SEM	0.022	0.034	0.025

All units = kPa.l^{-1}.s.
Mean total is a weighted average of inspiratory and expiratory resistances.
Control data for subject nine was rejected due to artifactual error.

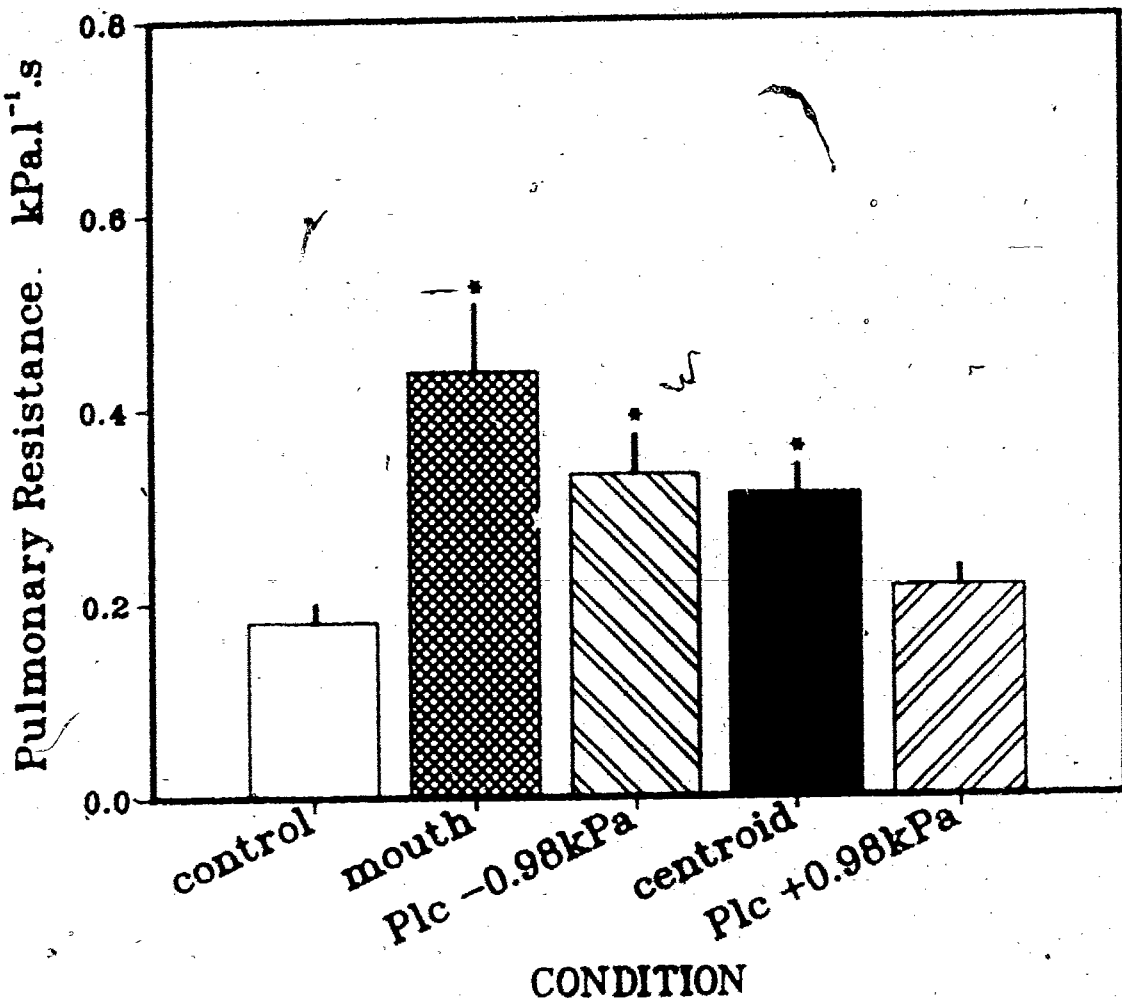


Figure 4.7: **Mean total pulmonary resistance during spontaneous breathing under control and immersion states, breathing air at four hydrostatic pressure loads.** [Resistance was averaged at a flow rate of 0.5 Ls^{-1}. Data represent means and standard errors. *=significantly different from control status ($p<0.05$).]

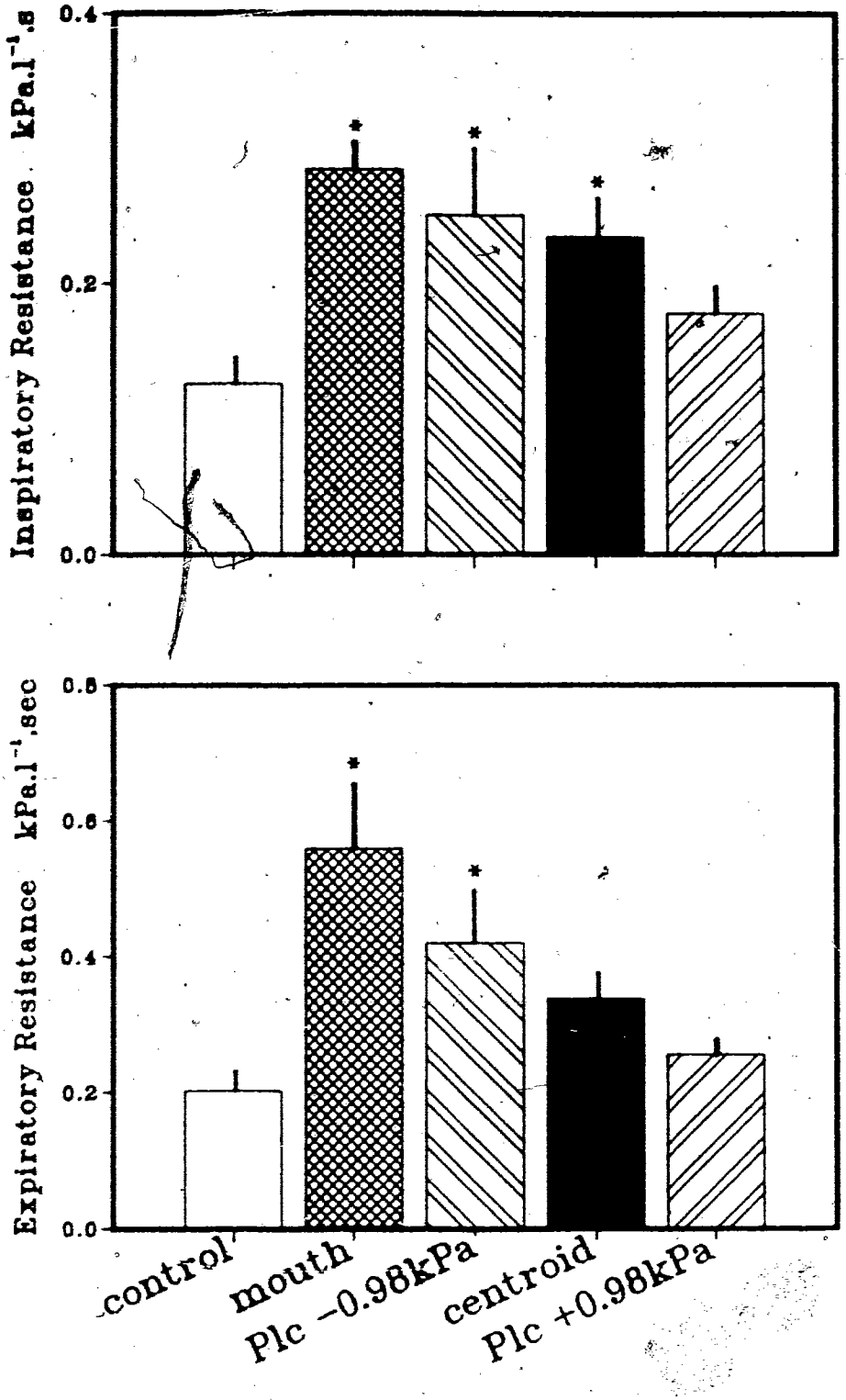


Figure 4.8: **Inspiratory and expiratory pulmonary resistances during spontaneous breathing under control and immersion states, breathing air at four hydrostatic pressure loads.** [Resistances were derived from measures taken at a flow rate of 0.5 l.sec^{-1}. Data represent means and standard errors. *=significantly different from control status ($p<0.05$).]

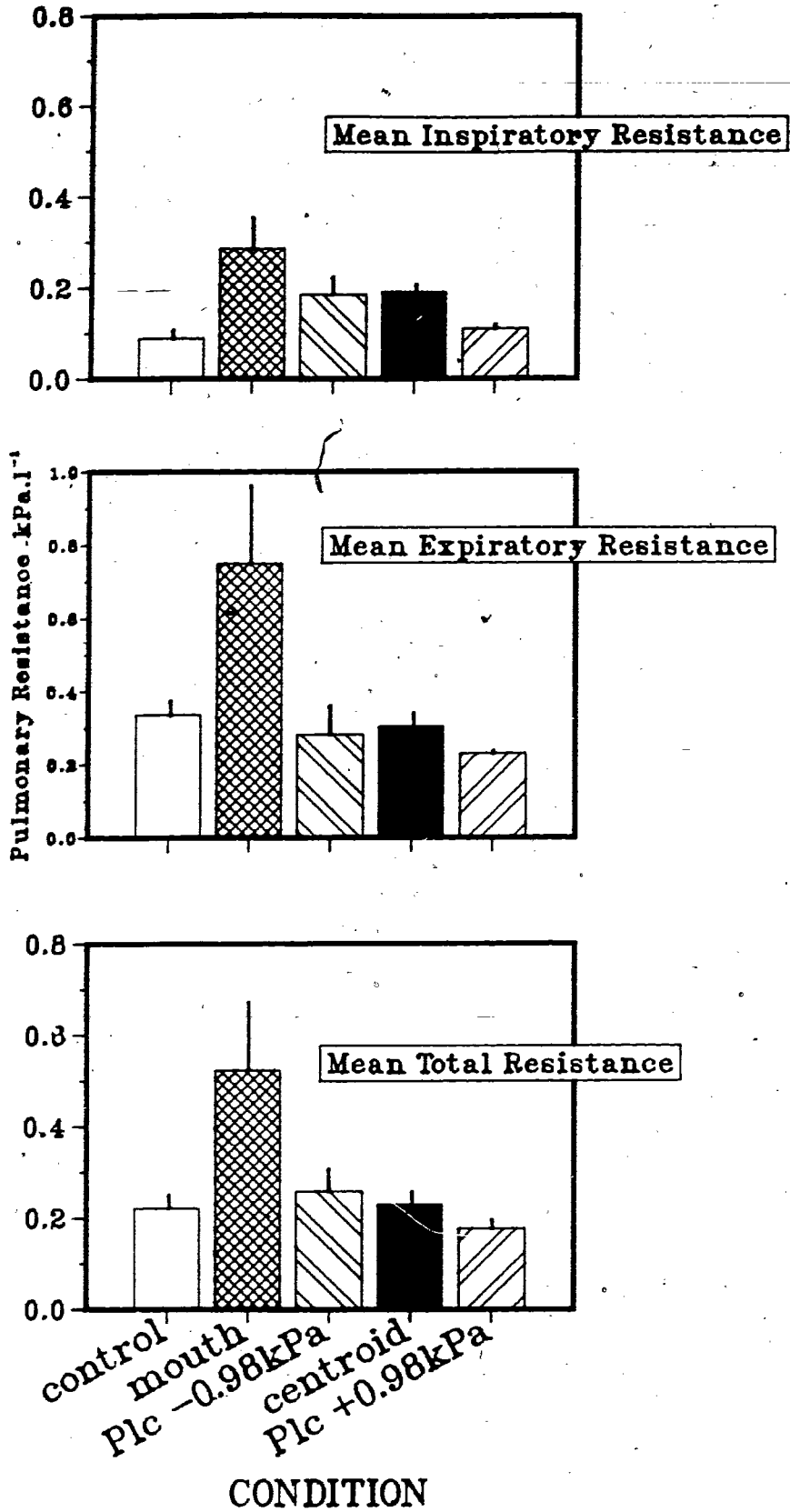


Figure 4.9: **Mean inspiratory**, expiratory and total breath pulmonary resistance, computed from resistances obtained over all flow rates during the respiratory cycle. [Data represent means and standard errors.]

delivery matched those recorded using the total R_{pul} at a constant flow rate (Figure 4.7). Resistance increased from 0.22kPa.l^{-1}.s in control trials, to 0.52kPa.l^{-1}.s during uncompensated immersion.

Functional pulmonary resistances ($R_{f(pul)}$) were computed to permit comparison with the immerson data of Dahlbäck *et al.* (1979). Results are shown in Figure 4.10. The control $R_{f(pul)}$ of 0.15kPa.l^{-1}.s were less than ($p>0.05$) total R_{pul} computed at either a constant flow rate (Figure 4.7), or when total R_{pul} was averaged over the full breathing cycle.

During total immersion, using mouth pressure air delivery, $R_{f(pul)}$ increased to 0.33kPa.l^{-1}.s. This change was about the same order of magnitude as seen when R_{pul} was averaged over the total respiratory cycle. Breathing pressure compensation did not significantly reduce $R_{f(pul)}$ accompanying immersion until air was breathed at P_{LC} plus 0.98kPa. The measurement technique appeared to lack the sensitivity necessary to detect differences between resistance at each of the compensation pressures. Means were significantly greater than control values ($p<0.05$), except during air breathing at 0.98kPa above P_{LC}.

4.2.4 Dynamic compliance.

Control static ($C_{st(l)}$) and dynamic pulmonary compliance ($C_{dyn(l)}$) values were not significantly different (3.24 and 2.91 l.kPa^{-1} respectively) ($p>0.05$, $\phi = 0.54$, Table 3.7, Figure 4.11). Values for $C_{dyn(l)}$ during immersion, when breathing at air pressures greater than mouth pressure, were not significantly different from controls ($p>0.05$, $\phi = 0.55$), in agreement with observations made during static manoeuvres (Chapter Three). Furthermore, $C_{st(l)}$ and $C_{dyn(l)}$, determined at P_{LC} were not significantly different ($p>0.05$, $\phi = 0.54$). These observations appear to validate $C_{dyn(l)}$ techniques during immersion when using these breathing pressures.

With air delivered at mouth pressure, $C_{dyn(l)}$ was reduced 43.8% to 1.63 l.kPa^{-1} ($p<0.05$). Every subject displayed this trend, which disagrees with the observations of no change in $C_{st(l)}$ made in Chapter Three.

4.2.5 Respiratory timing patterns.

Respiratory phase durations were computed to permit analysis of W_{pul} and R_{pul}. Since timing data was available, it was used to analyse variables commonly used as indices of respiratory control.

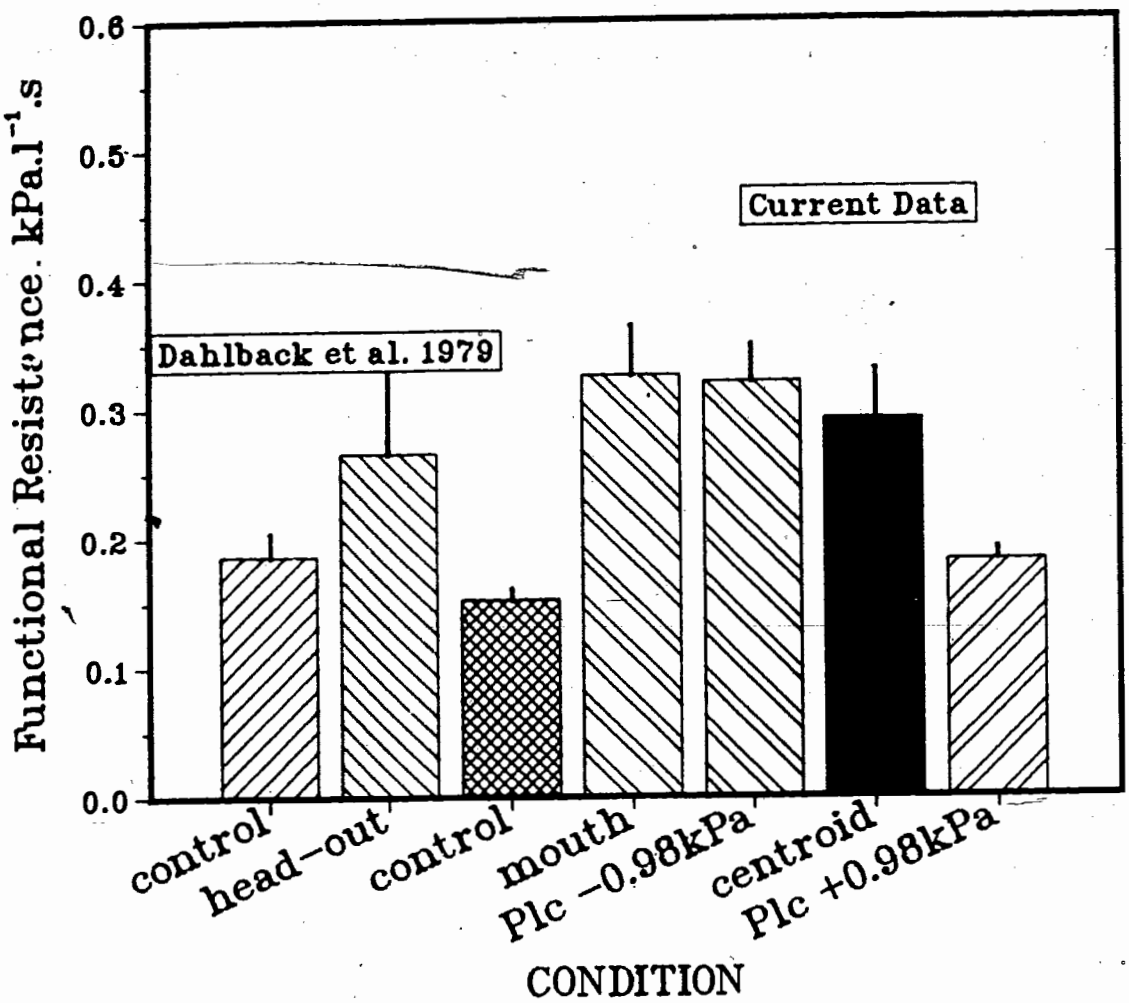


Figure 4.10: **Functional pulmonary resistance during spontaneous breathing under control and immersion states, breathing air at four hydrostatic pressure loads.** [Data represent means and standard errors.]

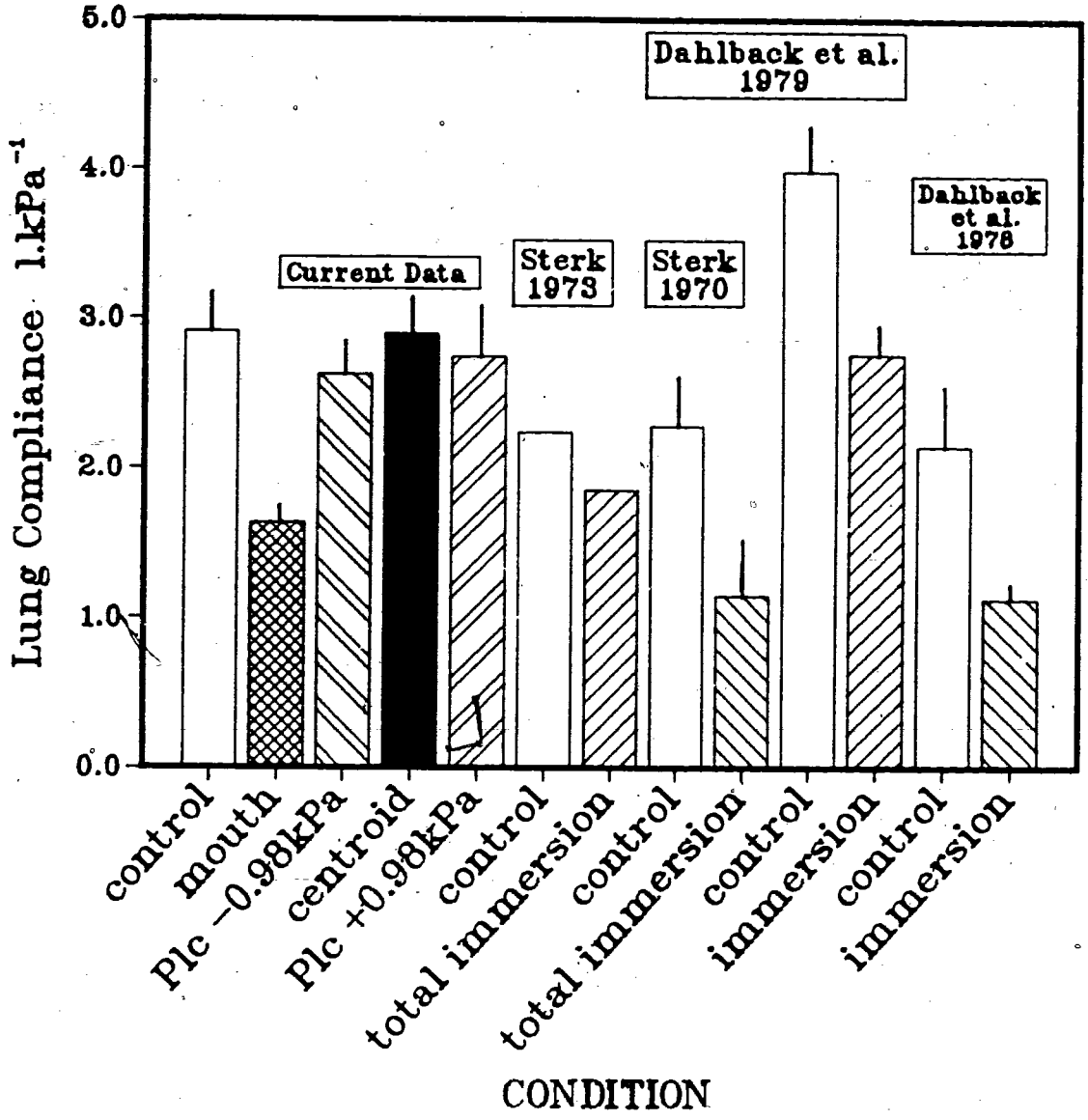


Figure 4.11: **Comparison of dynamic and quasi-static lung compliances measured upright in air and during immersion.** [Data from current investigation, Sterk (1970, dynamic; 1973, quasi-static) Dahlback *et al.* (1978, dynamic; 1979, quasi-static). Data represent means and standard errors.]

The ratio V_T/T_I (mean inspiratory flow – $l.s^{-1}$) has been used as an index of central inspiratory drive (Milic–Emili and Cajani 1957, Milic–Emili *et al.* 1975, Derenne *et al.* 1976, Milic–Emili *et al.* 1981). Uncompensated immersion increased V_T/T_I. Differences failed to achieve significance at the 0.05 level. Differences between experimental conditions, while showing a reduction at P_{LC}, were similarly not significant ($p>0.05$, $\phi = 0.47$, Table 4.6). The lack of significance between breathing pressures may be ascribed to impingement of experimental apparatus upon breathing patterns. For respiratory drive to be reliably assessed by V_T/T_I, subjects should not become consciously aware of their breathing patterns. This state was not achieved in the present experiments.

The ratio T_I/T_{TOT} is used as an index of respiratory timing or 'duty cycle' (Milic–Emili and Cajani 1957, Milic–Emili *et al.* 1975, Derenne *et al.* 1976, Milic–Emili *et al.* 1981). Differences between control and experimental means were similarly non–significant ($p>0.05$, $\phi=0.70$).

Table 4.6: **Respiratory timing patterns in air and during upright immersion, breathing air at four hydrostatic pressure loads.**

Condition	V_T	T_I	T_E	T_{TOT}	V_T/T_I	T_I/T_{TOT}	f_b	V_I
air	1.23	2.14	2.29	5.17	0.63	0.42	14.9	16.56
	(0.15)	(0.20)	(0.31)	(0.49)	(0.05)	(0.01)	(1.6)	(1.11)
mouth	1.75	1.73	1.61	3.84	1.00	0.45	18.0	27.29
	(0.38)	(0.14)	(0.22)	(0.33)	(0.18)	(0.01)	(2.0)	(4.78)
P_{LC}-0.98	1.28	1.59	1.37	3.62	0.84	0.44	18.8	22.87
	(0.20)	(0.07)	(0.14)	(0.22)	(0.13)	(0.01)	(1.7)	(3.82)
centroid	1.20	1.53	1.67	3.99	0.77	0.39	16.8	18.89
	(0.17)	(0.08)	(0.25)	(0.27)	(0.07)	(0.02)	(1.8)	(1.96)
P_{LC}+0.98	1.21	1.50	1.71	3.85	0.94	0.39	20.2	23.20
	(0.14)	(0.13)	(0.21)	(0.34)	(0.10)	(0.02)	(2.7)	(3.03)

Data represent means with standard error in parenthesis.
Abbreviations: V_T = tidal volume (l BTPS), T_I = inspiratory time (sec), T_E = expiratory time (sec), T_{TOT} = respiratory cycle duration (sec), f_b = breathing frequency ($b.min^{-1}$), V_I = minute ventilation ($l.min^{-1}$).
Units: pressure = kPa, V_T/T_I = $l.s^{-1}$, T_I/T_{TOT} = dimensionless.

4.3 DISCUSSION

4.3.1 Characterisation of the control flow-resistive mechanics in the current subjects.

Under control states, spontaneous breathing demanded a mean pulmonary power ($\dot{W}_{pul}$) to overcome flow-resistive forces, of 0.060 ± 0.026W. Evidence from the literature indicates that normal $\dot{W}_{pul}$ lies between 0.031 to 0.098W (McIlroy *et al.* 1954, Jaeger and Otis 1964, Agostoni *et al.* 1970, Ballantine *et al.* 1970, Holmgren *et al.* 1973), thus validating the current techniques and control values, and confirming respiratory normality of the subjects.

In all subjects, control expiratory flow-resistive work ($W_{pul(exp)}$) exceeded inspiratory work ($W_{pul(insp)}$). The mean, control inspiratory:expiratory work ratio during spontaneous ventilation was 0.538 [22] demonstrating greater expiratory flow-resistance. Expiratory R_{pul} (0.5 $l.s^{-1}$) exceeded mean inspiratory R_{pul} in six subjects. Few studies report such a comparison, however, Otis *et al.* (1950) and Mead and Whittenberger (1953) found approximately equivalent flow-resistances, while Uhl *et al.* (1972) reported a ratio of 0.68.

Spontaneous ventilation, under control states, yielded an inspiratory pulmonary resistance of 0.125kPa.l^{-1}.s (SEM = 0.022), when measured at a flow rate of 0.5$l.s^{-1}$. Normal pulmonary resistances (R_{pul}) range from 0.08 to 0.745kPa.l^{-1}.s (Marshall and DuBois 1956, DuBois 1964, Cotes 1979). These observations further validate the current techniques, and the normality of the flow-resistive status of the current subjects under control conditions (Table 4.5).

4.3.2 Respiratory flow-resistive work.

Hong *et al.* (1969) observed a two-fold increment in total flow-resistive respiratory work, moving from 'xiphoid' to 'neck' immersion. Sterk (1970) recorded a 4.7-fold rise, with total immersion using a mouth-held demand regulator. Some portion of this latter change is attributed to restrictive diving suits and SCUBA harness worn by the subjects[23]. In a subsequent report, Sterk (1973) observed a 3.5-fold

---

[22] Inspiratory:expiratory ratio = $W_{pul(insp)}/W_{pul(exp)}$ (dimensionless).

[23] This is evidenced by larger than normal reductions in vital capacity, which approximated 75% of control values. Immersion normally reduces VC 5-10% (Table 3.2).

pulmonary flow-resistive work elevation[24]. With allowance for immersion depth differences and thoracic restriction, the present data, showing an elevation of flow-resistive pulmonary work attending upright immersion of 3.8 times the control level, confirm these earlier observations. No previous investigators have studied the ability of breathing pressure manipulations to modulate W_{pul} during upright immersion, though several have hypothesised its implemention would be beneficial (Jarrett 1965, Craig and Dvorak 1975, Flynn *et al.* 1975). Such hypotheses are supported by the significant inspiratory, expiratory and total flow-resistive work reductions observed in the present investigation, which were mediated by air delivery at P_{LC} and P_{LC} plus 0.98kPa (Figures 4.5, 4.6).

Flow-resistive work is performed against friction in the airways, lung tissue and chest wall (Figure 4.1). Inherent limits of the current techniques meant only W_{pul} could be measured, without differentiation between airway and tissue components[25]. Airway friction is affected by changes in gas flow, density, viscosity and by airway dimensions. As downstream airways account for approximately 80% of frictional work (Hyatt and Wilcox 1961, Ferris *et al.* 1964, Macklem and Mead 1967, Hogg *et al.* 1968, Olsen *et al.* 1970), modifications during immersion might be expected to occur primarily in these larger airways. Work of Prefaut *et al.* (1976) has revealed no change in R_{aw} of the upstream airways during immersion[26], thus indirectly supporting the primary role played by the downstream airways.

The significant elevation of W_{pul} during uncompensated immersion, is ascribed primarily to volume-dependent reductions in airway diameter, concommitant with a decreased expiratory reserve volume (ERV). It is suggested that such volume changes elevate flow-resistance mainly by reducing the diameter of the larger airways.

This volume-dependence of R_{aw} was identified by Mead and Whittenberger (1953) and Briscoe and DuBois (1958). ERV decrements may be ascribed to a positive shift of the total respiratory compliance curve, chest compression, (Figure 3.11), and acute pulmonary vascular engorgement (Echt *et al.* 1974, Lange *et al.* 1974, Risch *et al.* 1978a, 1978b, Löllgen *et al.* 1980, Choukroun *et al.* 1983). The latter augments ERV changes, replacing air with blood (Dahlbäck 1975, Dahlbäck *et al.* 1978), and possibly elevating lung tissue frictional losses. Recent work by Jones *et al.* (1978) and Ishi *et al.* (1985)

---

[24] Sterk (1973) reported several values at different time intervals, the data above represents a weighted mean of these points.

[25] R_{aw} is assumed to represent 60-86% of W_{pul} (Table 4.1).

[26] Measured using the upstream R_{aw} technique.

has also implicated vagal participation in the airway narrowing that accompanies pulmonary blood volume elevation. Vagotomy was shown to decrease the effect of engorgement on airway diameter change (Ishi *et al.* 1985). Some interaction between these mechanical and neural mechanisms possibly occurs during immersion, the net result of which is an elevation in pulmonary resistance.

The importance of airway diameter changes to increments in W_{pul} is supported by the present observation of a significant elevation of constant flow R_{pul} attending uncompensated immersion (Figures 4.7, 4.8), and by previous investigators who found greater inspiratory R_{aw} (Agostoni *et al.* 1966, interrupter technique), $R_{f(pul)}$ (Dahlbäck 1978, Dahlbäck *et al.* 1979), mean R_{pul} (Sterk 1970, 1973), and total thoracic resistance (Löllgen *et al.* 1980) accompanying immersion.

The fact that the current change in R_{pul} (computed at a flow rate of 0.5 $l.s^{-1}$) accompanying uncompensated immersion exceeded observations of Agostoni *et al.* (1966), Dahlbäck (1978), and Dahlbäck *et al.* (1979), may be attributed to differences in measurement technique (see: Section 4.3.3), to the depth of subject immersion[27] and its concommitant influences upon lung volume, to differences in minute ventilation between control and experimental states (in each study), and to differences in the control resistance. Dahlbäck *et al.* (1979) reported only the mean minute ventilation, averaged over control and experimental trials. It can be seen in the present investigation that minute ventilation, and consequently airflow, increased about 65% (Table 4.7) for the same comparison. If airflow became more turbulent in the present study, it may account for the discrepency between the two investigations.

The current constant flow increments in inspiratory R_{pul} do agree with R_{aw} changes reported by Agostoni *et al.* (1966), using negative pressure breathing, a state which is at least physically analogous to the current uncompensated immersions.

Evidence supporting the lung volume-dependency of flow-resistive work is obtained from a comparison of the ERV and expiratory R_{pul} at a flow rate of 0.5 $l.s^{-1}$ (Figures 3.11, 4.8). For simplicity data has been reproduced in Figure 4.12. Expiratory R_{pul} altered inversely with ERV. The imperfect nature of the relationship may indicate involvement of other causal factors (*e.g.* vagally mediated airway narrowing) and/or the inability of ERV to reflect altered functional residual capacity (FRC). It is recognised that FRC is a better volume index with which to compare R_{pul}, however, it remained unknown in the current study, since residual volume (RV) was not measured. The addition of a constant RV to the ERV would alter the data shown in Figure 4.12. With air provided at mouth and P_{LC} minus

[27] No previous reports, except Sterk (1970, 1973), have employed total immersion.

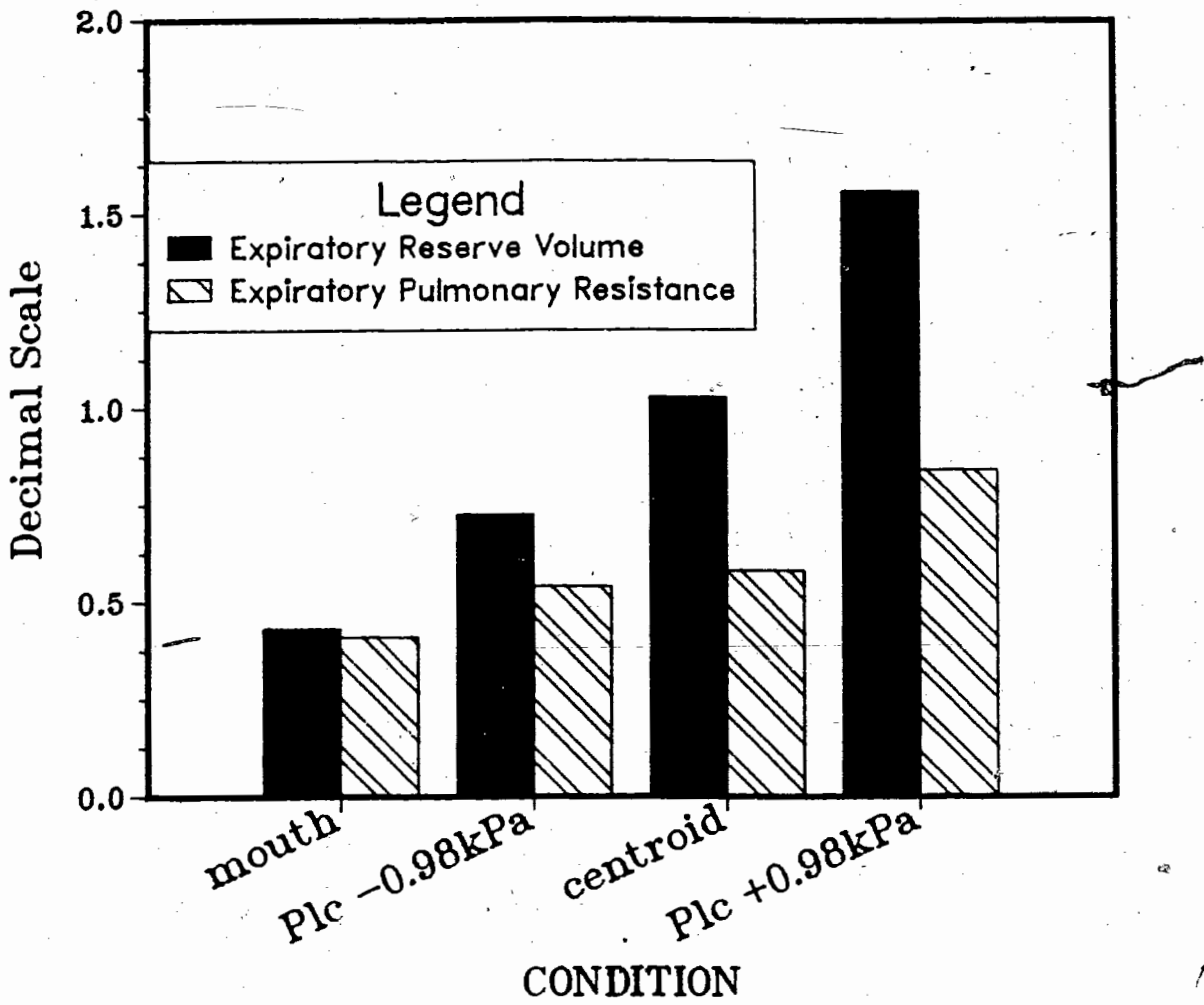


Figure 4.12: **Inter-relationship between expiratory reserve volume and the expiratory pulmonary resistance at four air delivery pressures, during upright immersion.** [ERV is expressed as a decimal relative to control; pulmonary resistance was calculated at a flow rate of 0.5 $l.s^{-1}$ and is expressed as a reciprocal of its decimal relation to control.]

0.98kPa, the ERV bars would lengthen, while at P_{LC} plus 0.98kPa, the ERV bar would shorten. The net result would be a profile similar to that of the expiratory R_{pul}. However, it is possible that RV also varied with delivery pressure, but since no previous works have been located which studied this relationship, the question is unresolved.

At low lung volumes during immersion, airway closure occurs (Dahlbäck and Lundgren 1972, Dahlbäck 1975), and is associated with narrowing of the larger airways (Dahlbäck 1978). Glaister *et al.* (1973) demonstrated that such a closure may further increase W_{pul}, since greater P_{tp} must be generated in order to open and ventilate occluded alveoli.

Analysis of $W_{pul(insp)}$ and $W_{pul(exp)}$ (Figure 4.6) revealed mouth pressure air delivery induced significantly greater flow-resistive work. Control inspiratory:expiratory W_{pul} ratio shifted from 0.54 to 0.62 in this condition due to proportionately greater $W_{pul(insp)}$ increments. Greater inspiratory flow-resistive work could induce respiratory discomfort and eventually produce respiratory fatigue.

It was observed that inspiratory W_{pul}, once reduced by the initial use of breathing pressure compensation, failed to respond to further modifications of the air supply pressure (Figure 4.6). Expiratory W_{pul} underwent greater changes during all air delivery pressures and always exceeded inspiratory W_{pul}. This observation may indicate that expiratory flow-resistance is more sensitive to lung volume changes. Theoretically, inspiratory resistance should demonstrate a similar sensitivity, but beyond the initial lung volume increments, experienced when air supply pressure was increased from mouth to P_{LC} minus 0.98kPa, it failed to respond to the changes in expiratory reserve volume (Figure 3.11). The absence of change with altered breathing pressure cannot be explained on the basis of the current measurements.

During the respiratory cycle, work is performed to overcome elastic and flow-resistive forces (Figure 4.1), thus a knowledge of total work is important. The summation of total W_{pul} and the inspiratory muscle work obtained from static pressure-volume curve analysis (Chapter 3) is inappropriate, since part of the flow-resistive work is performed by elastic recoil. However, in both cases, work partitions have been shown to increase (Figures 4.5, 3.8) when subjects were immersed without breathing pressure compensation. It is hypothesised that the combined effect of these elevations may increase ventilatory oxygen consumption ($\dot{V}O_2$) to the point where, during exercise at 1 ATA, it may impose a respiratory limitation on physical power.

In diving, particularly at depth, R_{aw}, and consequently flow-resistive work, becomes a major concern due to the added resistance of the underwater breathing apparatus (Morrison and Reimers 1982, Morrison *et al.* 1986), and density-dependent increments in R_{aw} (Buhlmann 1963, Glauser *et al.* 1967, Uhl *et al.* 1972, Vorosmarti *et al.* 1975, Van Liew 1982, 1983). It is postulated that the use of breathing pressures, designed to compensate the hydrostatic pressure imbalance encountered during upright immersion, would ameliorate the respiratory mechanical disadvantage normally encountered.

Evidence supporting this proposal is illustrated in Figures 4.5 to 4.8. In almost every instance W_{pul} and R_{pul} were diminished using air delivered at P_{LC} and P_{LC}±0.98kPa. At these pressures, W_{pul} and R_{pul} were not significantly greater than control levels.

It is suggested that the mechanism reponsible for improved respiratory dynamics is associated with ERV increments and vascular disgorgement. Positive pressure ventilation forces blood out of the lungs (Fenn *et al.* 1947, Kilburn and Sieker 1960, Orlow *et al.* 1985). Since compensation of a negative hydrostatic imbalance is analogous to the application of a positive pressure upon a balanced system, it is assumed that the use of P_{LC} air provision similarly translocates central blood to the periphery. Disgorgement may remove the vagally mediated airway constriction (Jones *et al.* 1978, Ishi *et al.* 1985), and reduce airway closure at low lung volumes (Dahlbäck and Lundgren 1972, Dahlbäck 1975) and its concommitant narrowing of the larger airways (Dahlbäck 1978). Perhaps of greater significance is the return of control ERV (Figure 3.11) mediated by P_{LC} air supply shifting the total respiratory compliance curve negatively, and increasing the thoracic relaxation volume.

It is concluded that demand regulator air supply pressures be adjusted, so that in the upright position, divers may obtain air at a positive pressure of 1.33kPa (relative to mouth pressure).

It appears from the preceding data that a higher delivery pressure may best replicate the pulmonary mechanical status that exists under control conditions. The decision to suggest that P_{LC} be adopted is based upon the following points:

(1) Differences between results obtained using P_{LC} and P_{LC} plus 0.98kPa air delivery pressure were non-significant in every instance.

(2) P_{LC} air provision resulted in a closer approximation of the control inspiratory to expiratory flow-resistive W_{pul} ratio.

(3) Air supply at P_{LC} plus 0.98kPa had been shown in Chapter Three to increase the expiratory

reserve volume to 50% of the vital capacity[11] (Figures 3.11, 3.12). Hyperinflation, whilst reducing R_{aw}, shortens the diaphragmatic fibres, and possibly reduces contraction efficiency (Evans and Hill 1914, Marshall 1962, Pengelly *et al.* 1971, Farkas *et al.* 1985), which may predispose subjects to respiratory fatigue (Roussos *et al.* 1979, Bye *et al.* 1983), and dyspnea (Killian and Jones 1984) during hyperpnea.

(4) As illustrated in Figure 3.15, subjects, when breathing air at P_{LC} plus 0.98kPa, appeared to defend an expiratory reserve volume 23% below the thoracic relaxation volume. This maintenance of end-expiratory muscle tone consumes energy, the magnitude of which is unable to be evaluated using the current W_{pul} measurements. Such metabolic work may enhance the onset respiratory fatigue.

(5) Finally, one must consider that air delivery at pressures in excess of P_{LC}, represent positive pressure ventilation, which has been shown to reduce cardiac output, to increase pulmonary shunting, and to precipitate pulmonary oedema (Sladen *et al.* 1968, Kuckelt *et al.* 1981, Venus and Jacobs 1984, Dreyfuss *et al.* 1985). Recent work by Dahlbäck and Balldin (1984) has, however, not found positive pressure breathing during heavy exercise to produce detrimental cardio-pulmonary effects.

4.3.3 Technical critique.

The measurement of pulmonary resistance using the functional resistance ($R_{f(pul)}$) method of Ahlstrom and Jonson (1974) and Jansson and Jonson (1975) was incorporated to facilitate data comparison between the current work and that of Dahlbäck *et al.* (1979). Results have shown divergence from resistance computed at a flow rate of 0.5 l.s^{-1}. Some disparity also exists between R_{pul} measured at a constant flow, and averaged over all flows.

The assumption of the $R_{f(pul)}$ technique is that pulmonary resistance may be described by a linear model. If correct, this implies that resistance will remain constant regardless of airflow changes. However, if a non-linear model is more appropriate, resistance will change as a function of the variations in respiratory flow. Since during normal spontaneous ventilation, respiratory airflow is best described as some combination of laminar and turbulent flow (Cotes 1979), the validity of the linear model was examined.

[11] In control trials ERV averaged 35% of VC.

The comparison of data for inspiratory R_{pul} obtained at a constant flow (Figure 4.8) and averaged over all flows (Figure 4.9), shows the latter to be generally smaller, except when subjects were provided air at mouth pressure. If R_{pul} is best described by a non-linear model then one would expect data to be influenced by airflow. Current mean flows exceeded 0.5 $l.s^{-1}$ in all conditions except the control trials. In a non-linear system this would have produced a resistance in excess of that measured at a constant flow. Since lower values were generally observed, the results indicate that while turbulence may have been present, flow patterns during spontaneous ventilation appeared to be predominantly laminar.

Functional resistance calculations rely upon pulmonary resistance being a constant, regardless of flow (Ahlstrom and Jonson 1974, and Jansson and Jonson 1975). In the present study, inspiratory R_{pul} data does not totally validate this assumption. When air was provided at mouth pressure, minute ventilation averaged 27.3 $l.min^{-1}$, while during control trials the ventilation was about 40% lower (16.6 $l.min^{-1}$). Under these conditions mean flow R_{pul} more closely approached R_{pul} measured at 0.5 $l.s^{-1}$, however, in all other trials mean flow R_{pul} underestimated that measured at a constant flow (Figures 4.8, 4.9). Ventilations were also lower than during uncompensated immersion (Table 4.6). If turbulent flow was present, R_{pul} at high flow rates should have increased to equal or exceeded constant flow R_{pul}. Since this was observed when subjects were delivered at mouth pressure, it was concluded that the amount of flow turbulence had increased with elevated minute ventilation. The significance of this observation for resting studies appears minor, however, it may become critical during ventilations encountered during exercise.

The assumption of sinusoidal breathing patterns used in computing $R_{f(pul)}$ (Ahlstrom and Jonson 1974, Jansson and Jonson 1975) was not directly evaluated in the present study. However, several subjects demonstrated breathing patterns which were not sinusoidal, particularly when breathing air at pressures in excess of P_{LC}. Any breathing pattern which is not sinusoidal, may produce differences between inspiratory and expiratory durations, peak and mean flows and pauses. Such changes may occur across experimental conditions, even though subjects maintain a fairly constant minute ventilation.

The disparity between the results of Dahlback *et al.* (1979, Table 4.2) and the current $R_{f(pul)}$, may be attributed to differences in minute ventilation and breathing patterns between studies, and between experimental conditions within the current investigation. The failure of the $R_{f(pul)}$ measurement to detect flow-resistance changes due to altered breathing pressure, is similarly ascribed to variations in respiration pattern between experimental conditions.

Dynamic pulmonary compliance ($C_{dyn(l)}$) was studied during the current series of experiments to answer technical criticisms raised within Chapter Three. Inadequacy of non-static compliance techniques was suggested since such methods: (a) may violate assumptions of equivalence between P_{ao} and P_{alv}; (b) may include pressure variance attributable to flow-resistance (Gibson and Pride 1976); (c) measurements are confounded when disease (or environmental) perturbations alter the time constants [29] of parallel lung units (Radford 1964, Gibson and Pride 1976, Cotes 1979); and (d) the results are sensitive to minor airway dysfunctions (Gibson and Pride 1976).

During immersion with each level of pressure compensation, differences between $C_{dyn(l)}$ and $C_{st(l)}$ were non-significant (Table 3.7, Figure 4.11). At mouth pressure, $C_{dyn(l)}$ was significantly reduced (43.8%), in agreement with data of Sterk (1970, 49.8%), Dahlbäck *et al.* (1979, 47.6%) and Baer *et al.* (1986, 36.2%).

Quasi-static compliance measurements have shown smaller reductions with immersion when compared with quasi-static control observations (Sterk 1973, 17.4%; Dahlbäck *et al.* 1978, 30.8%), while $C_{st(l)}$ was unchanged in this condition (Table 3.7). These observations are interpreted as indicating that the static lung compliance curve was essentially linear over the lung volumes encountered during the current investigation. Altered compliance obtained from the non-static techniques, indicates a non-linear pressure-volume relation over this lung volume range. It is apparent that static and non-static techniques, while being designed to quantify a common mechanical variable, appear to be measuring different attributes.

Upright immersions (head-out and total) produce increases in R_{pul}. Such a change may resemble that accompanying airway disorders. Supporting this proposition is the significant two to three-fold increase in W_{pul} and R_{pul}. It is suggested that mechanically and/or neurally mediated airway narrowing may be responsible for the apparent $C_{dyn(l)}$ reduction (lung stiffening), since in this state $C_{dyn(l)}$ may be measuring factors other than lung tissue elasticity (Gibson and Pride 1976).

The occurrence of negative, end-expiratory P_{tp}, in all subjects during uncompensated immersion, provides another possible explanation. Such pressures result from compression artifact and/or air trapping. If the former is correct, current $C_{dyn(l)}$[30], and possibly the data of other researchers, is invalid. In the latter case, the pressure-volume relation describes behaviour of compartments with and without

[29] Time required to equilibrate following a pressure change.

[30] $C_{dyn(l)}$ is computed using only end-inspiratory and end-expiratory P_{tp}.

direct communication with the mouth. In this instance the P_{tp} would reflect tissue and gas elasticity. The inclusion of such data may render the interpretation of $C_{dyn(l)}$ doubtful. Clearly there is need for further research and for caution in the use and interpretation of $C_{dyn(l)}$ during immersion, when subjects are not provided with air at pressures greater than the hydrostatic pressure at mouth depth.

CHAPTER 5

PHYSICAL POWER OF THE UPRIGHT, IMMERSED DIVER WITH AND WITHOUT HYDROSTATIC PRESSURE COMPENSATION

Normobaric physical power[1] is generally limited by cardiovascular rather than respiratory mechanisms (Asmussen 1964, Holmgren 1967, Saltin and Åstrand 1967, Saltin 1973, Åstrand 1976, Holloszy and Booth 1976, Saltin and Rowell 1980, Stubbing *et al.* 1980a) [2], however, patients with respiratory disorders consistently present with ventilatory mediated, exercise intolerance (Laurenco *et al.* 1965, Levison & Cherniack 1968, Clark *et al.* 1969, Leaver and Pride 1971, Potter *et al.* 1971, Stubbing *et al.* 1980b). It has been postulated that immersion without hydrostatic pressure compensation, may induce respiratory mechanical perturbations capable of reducing physical power (Thalmann *et al.* 1979, Lundgren 1984).

Numerous research groups have investigated aerobic power under hyperbaric conditions with divergent observations. Several groups have found aerobic power was not reduced when studied at depths to 6 ATA (Fagraeus *et al.* 1973, Fagraeus 1974, Anthonisen *et al.* 1976, Linnarsson and Fagraeus 1976). However, these studies investigated exercise with dry compression. Under these conditions, immersion-induced modifications to respiratory mechanics were absent.

Dressendorfer *et al.* (1976) found that maximal aerobic power during immersed exercise at 1 ATA (head-out cycling) was reduced along with minute ventilation, when compared with dry control data.

Several studies have investigated aerobic power of the immersed diver under hyperbaric states. Morrison (1973) reported prone subjects to experience respiratory difficulty. No consistent decrement in maximal aerobic power was observed with increased absolute pressure, however, not all subjects were able to complete the exercise protocol.

Dwyer *et al.* (1977) investigated upright, immersed divers at an absolute pressure of 43.4 ATA using a helium-oxygen breathing mixture. At this pressure the gas density was approximately equivalent to air density at 6 ATA. Unlike the dry experiments at this gas density (Fagraeus *et al.* 1973, Fagraeus

---

[1] The capacity to perform a given amount of work in a period of time.

[2] Recent work by Dempsey *et al.* (1981) has, however, demonstrated arterial hypoxaemia in elite endurance athletes. Such a ventilatory limitation is not applicable to normal, trained subjects.

1974, Anthonisen *et al.* 1976, Linnarsson and Fagraeus 1976), Dwyer *et al.* (1977) found maximal aerobic power to be reduced. The authors concluded that factors other than gas density were responsible for the observed power decrement.

Spaur *et al.* (1977) reported an inability of subjects to exercise at an $\dot{V}O_2$ in excess of 1.92 l.min^{-1} when immersed upright at 49.5 ATA, breathing a helium-oxygen air mixture. The authors concluded that reduced power was due to a density dependent increase in airway resistance (R_{aw}).

It has been demonstrated that uncompensated, upright immersion elevates total respiratory and chest wall elastic work (Hong *et al.* 1969, Flynn *et al.* 1975, Figures 2.12 and 3.7), flow-resistive respiratory work (Hong *et al.* 1969, Sterk 1970, 1973, Figures 4.5 and 4.6), and pulmonary and airway resistance (Agostoni *et al.* 1966, Sterk 1970, 1973, Dahlbäck 1978, Dahlbäck *et al.* 1979, Löllgen *et al.* 1980, Figures 4.7, 4.8). Previous chapters of this thesis have also illustrated the capacity of hydrostatic breathing pressure compensation to return respiratory mechanical attributes towards control status (Figures 2.12, 3.7, and 4.5 to 4.8). It is of interest to professional divers to know whether such compensation may improve respiratory mechanics during immersed exercise and in so doing, increase physical power.

Thalmann *et al.* (1979) studied the influence of breathing pressure (static load) on prone, immersed exercise at absolute pressures from 1.45 to 6.76 ATA. The authors reported a lower maximal minute ventilation when subjects were exposed to a negative breathing pressure[1]. Maximal aerobic power increased with absolute pressure, when breathing at a neutral static load. However, only one subject (N=3) was able to complete maximal exercise trials with a static breathing load of −0.98kPa at 6 ATA. All subjects completed submaximal trials at static loads of −1.98 and −0.98kPa. However, dyspneic sensations were greater and often severe.

It appears that static breathing load may influence physical power during immersed exercise with a raised absolute pressure. Pulmonary compliance is reported to be unchanged during normobaric exercise (Granath *et al.* 1959, Chiang *et al.* 1965, Olafsson and Hyatt 1969), yet recent work by Stubbing *et al.* (1980a) illustrated decrements with increased work intensity. Airway resistance (R_{aw}) is also reported to remain unaltered during exercise (Granath *et al.* 1959, Chiang *et al.* 1965, Olafsson and Hyatt 1969, Stubbing *et al.* 1980a). Of major concern remains the ventilatory changes associated with altered

[1] This would be physically analogous to breathing air delivered at a negative pressure relative to P_{LC}, when immersed upright.

barometric pressure (P_A).

The density of air changes as a function of barometric pressure (P_A) (Glauser *et al.* 1967). When air flow contains turbulence, Reynold's number will increase proportionately with gas density (all else remaining constant) [4]. Thus espiratory airflow becomes more turbulent as P_A increases, elevating R_{aw} (Buhlmann 1963, Glauser *et al.* 1967, Uhl *et al.* 1972, Vorosmarti *et al.* 1975, Peterson and Wright 1976, Clarke *et al.* 1982, Van Liew 1982, 1983, Dahlbäck 1985). P_A has almost no influence on gas viscosity (Hirschfelder *et al.* 1954), leaving R_{aw} to be primarily a function of gas density and ventilatory rate.

At 1 ATA dynamic airway compression[5] limits maximal ventilation (Fry *et al.* 1954, Hyatt *et al.* 1958, Fry and Hyatt 1960, Mead *et al.* 1967, Castile *et al.* 1980), however, during normobaric exercise, airway compression is not generally observed, as peak ventilations, and consequently peak expiratory transpulmonary (P_{tp}) pressures, are not attained (Olafsson and Hyatt 1969, Leaver and Pride 1971, Potter *et al.* 1971, Hesser *et al.* 1981). At raised ambient pressure, Wood and Bryan (1971) measured values of expiratory P_{tp} beyond levels associated with airway compression at 1 ATA. These values occurred at lower exercise ventilations as gas density increased, supporting earlier observations of density dependent expiratory limitations (Schilder *et al.* 1963).

One might predict R_{aw} increments, and associated dynamic compression, to limit exercise ventilation at depth. Numerous investigations support this prediction during chamber (Wood 1963, Taunton *et al.* 1970, Bradley *et al.* 1971, Uhl *et al.* 1972, Fagraeus 1974, Linnarsson and Fagraeus 1976, Thalmann *et al.* 1979, Hesser and Lind 1981, Morrison and Wood 1986) and open ocean dives (MacDonald and Pilmanis 1981) to moderate depths. Deep dives beyond 40 ATA revealed conflicting observations (Salzano *et al.* 1981, 1984), but comparison is confounded due to varied breathing gas densities and viscosities.

Lanphier (1963) postulated ventilatory decrements were due to depressed respiratory drive as a result of heightened O_2 partial pressures or narcosis. The latter has subsequently been discounted (Fagraeus and Hesser 1970, Lambertson *et al.* 1973, Gelfand *et al.* 1980), while the former was supported by studies utilising elevated O_2 concentrations (Bannister and Cunningham 1954, Lambertson *et al.* 1963,

---

[4] Reynold's number = linear velocity·density·tube diameter/viscosity.

[5] Dynamic airway compression is caused when expiratory effort is raised, in an attempt to overcome flow-resistance. The increased effort produces downstream points of negative pressure across the airway wall (*i.e.* airway pressure - pleural pressure), resulting in compression of these airways.

Miller *et al.* 1974). At submaximal hyperbaric exercise, reduced ventilation has been ascribed to diminished anaerobiosis, reducing *p*H respiratory drive, and to reduced response to CO_2 drive (Lanphier and Camporesi 1982). During anaerobic work, Whipp (1981) suggested high O_2 concentrations would impair the carotid body's capacity to compensate for the lactic acidosis.

Fagraeus (1974) described the hyperbaric ventilatory suppression as being either O_2-dependent (related to O_2 partial pressures) or non-O_2-dependent (related to R_{aw}), with both mechanisms contributing equally. It is important to identify conditions under which the non-O_2-dependent factors reach the point where ventilation is limited. Under these states, ventilation may reduce physical power. Identification of such conditions is essential if underwater breathing apparatus is to be designed which does not impose a further ventilatory hindrance to the working diver.

One approach to the evaluation of non-O_2-dependent variables has involved the comparison of exercise and maximal voluntary ventilations (MVV)[6]. MVV has been found to decline as a function of P_A, or gas density (Figure 5.1, $y=197.59 \cdot x^{-0.455}$[7]). The decrement is attributable to a density-induced rise in R_{aw}, accompanied by dynamic airway compression. Severe dyspnea accompanying ventilations approaching 95% MVV at depth (Anthonisen *et al.* 1976), has prompted speculation that exercise at depth is limited when the level of ventilation required, coincides with MVV for that depth (Van Liew 1982, 1983, Anthonisen 1984).

The applicability of the MVV gauge may be questioned at 1 ATA, since MVV cannot be maintained much beyond 20 seconds (Vorosmarti *et al.* 1975, Bye *et al.* 1983), and maximal exercise ventilations ($V_{I_{max}}$) range between 50-80% MVV (Zocche *et al.* 1960, Shephard 1966, 1967, Olafsson and Hyatt 1969, Fagraeus and Linnarsson 1973, Hesser *et al.* 1981). However, with the MVV decrement at depth, it becomes possible that $V_{I_{max}}$ may approximate MVV. Fagraeus and Linnarsson (1973) and Hesser *et al.* (1981) demonstrated that $V_{I_{max}}$ reached only 80% and 81.8% MVV at 3 and 6 ATA respectively. $V_{I_{max}}$ was derived, in the latter study, during the final 15-30 seconds of maximal exercise and would be expected to exceed the sustainable ventilation. On the basis of this data, one would not expect $V_{I_{max}}$ to reach MVV at 6 ATA, as the latter manoeuvre is performed at greater lung volumes (Hesser *et al.* 1981), permitting a lower R_{aw} at equivalent gas densities, due to airway enlargement.

---

[6] See Chapter One.

[7] This equation was obtained by regression analysis of the data presented in Figure 5.1.

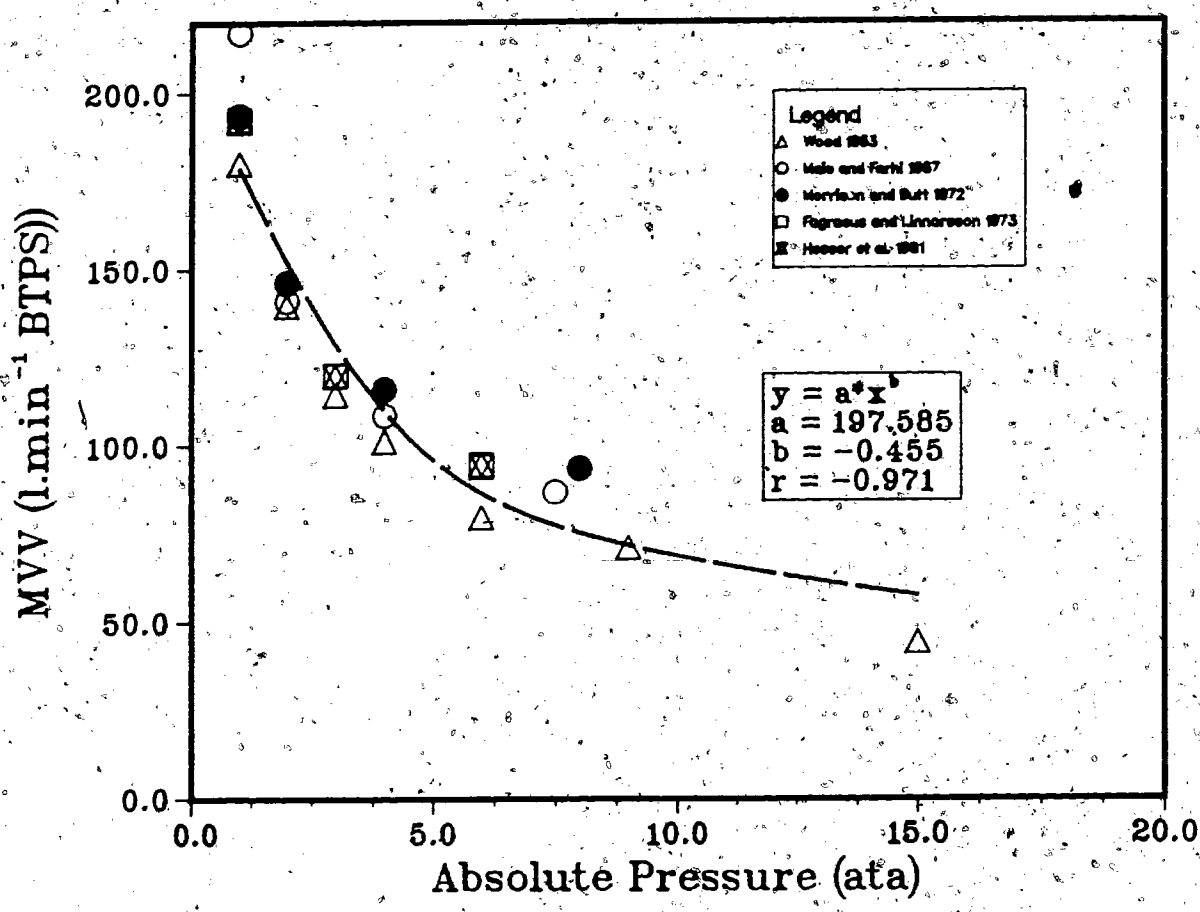


Figure 5.1: **Maximal voluntary ventilation as a function of atmospheric pressure.** [Data taken from: Wood (1963), Maio and Farhi (1967), Morrison and Butt (1972), Fagraeus and Linnarsson (1973), Hesser *et al.* (1981). The equation for the curve drawn was obtained by regression analysis using the published raw data.]

Another approach to the analysis of the non-O_2-dependent ventilatory reduction, has involved the investigation of P_{tp}. Wood and Bryan (1971) investigated maximal P_{tp} at various lung volumes during expiration over pressures from 1 to 10 ATA. Expiratory flow increased with driving pressure, until it reached a maximal flow plateau at the point where airway compression restricted expiration. The pleural pressure at which airway compression was initiated, was found to decrease with increments in absolute pressure (when measured at the same lung volume), and to decrease when lung volume was lowered. Both observations indicate an earlier onset of dynamic airway compression, and an effort-independent expiratory flow limitation. At 4 ATA they observed dynamic airway compression during exercise, and at 10 ATA it contributed to an exercise limitation.

Hesser *et al.* (1981) found that maximal P_{tp} increased with depth. Closer analysis of their pressure-volume and flow-volume diagrams shows the observation is not contradictory to that of Wood and Bryan (1971). The latter investigated P_{tp} at a constant flow. Data from Hesser *et al.* (1981) show peak P_{tp} was not associated with greater, but with smaller peak flow rates at increased depth. Thus these results only refer to the P_{tp} change in excess of that necessary to produce peak flows. Dynamic airway compression occurred in the Hesser *et al.* (1981) study, and was associated with $\dot{V}_{I_{max}}$ reductions of 35.7% and 48.1% at 3 and 6 ATA respectively.

A third approach to analysing the non-O_2-dependent ventilatory reduction has centered around the concept of *critical ventilation*. At normobaric rest, respiratory oxygen consumption ($\dot{V}O_2$) approximates 2.5ml.min^{-1} (Milic-Emili and Petit 1960, Otis 1964). Assuming resting total $\dot{V}O_2$ averages 250-300ml.min^{-1}, respiratory $\dot{V}O_2$ consumes 0.8-1.0% of this total. It was hypothesised that respiratory $\dot{V}O_2$ may increase, with ventilation, to reach a point beyond which further increments in ventilation fail to provide more O_2, but instead would result in excessive respiratory $\dot{V}O_2$ (Margaria *et al.* 1960, Otis 1964, Shephard 1966). This ventilatory level is called the *critical ventilation* , and has been measured and approximated at 120-160 l.min^{-1}. These figures correspond with 60-80% MVV.

Estimates of respiratory $\dot{V}O_2$ at maximal exercise at 1 ATA range from 7% to 20% of maximal $\dot{V}O_2$ (Milic-Emili *et al.* 1962, Shephard 1966, Bradley and Leith 1978, Pardy *et al.* 1984), representing substantial contributions. Macklem (1980) demonstrated that in resting subjects, fatiguing inspiratory resistance could elevate resting total $\dot{V}O_2$ by over 500ml.min^{-1}, presenting a possible respiratory $\dot{V}O_2$ ceiling. During exercise such a resistive load would impair work capacity, however, it is unknown whether resistive loads associated with raised absolute pressure can add sufficient respiratory $\dot{V}O_2$ to

compromise performance.

Ventilatory decrements lower the respiratory $\dot{V}O_2$ at depth, but density-dependent R_{aw} increments will counter this reduction. Evidence that permits evaluation of this exchange is not available. However, Glauser *et al.* (1967), using sulphur hexafluoride to simulate air densities of 4.5ATA (or heliox at 30 ATA), found respiratory $\dot{V}O_2$ at moderate hyperventilation (36 $l.min^{-1}$) did not limit O_2 available for other tissues. When breathing air at depth, the O_2 partial pressure is increased in proportion with the change in absolute pressure. This change may negate the importance of the *critical ventilation* theory during hyperbaric states.

It is apparent that ventilatory limitations at depth, whether related to MVV, maximal P_{tp} or respiratory $\dot{V}O_2$, are mediated by density-dependent elevations of R_{aw}. Such increments are associated with dynamic airway compression upon expiration. Fagraeus (1981) suggested this causes exercising subjects to prolong expiration to ensure adequate time for air expulsion, and then to shorten inspiration. Thus, while expiration limits ventilation, the extra inspiratory effort required would result in work-limiting dyspnea (Fagraeus 1981). In this respect, numerous research groups have reported subjects to feel limited by their inability to inspire an adequate tidal volume, or to 'catch up' on the ventilation required at a given exercise level (Dwyer *et al.* 1977, Spaur *et al.* 1977, Hesser and Linnarsson 1977, Thalmann *et al.* 1979, Salzano *et al.* 1981, Lundgren 1984).

5.0.4 Purpose of the chapter.

Previous sections of this thesis have illustrated the capacity of hydrostatic pressure compensation to return respiratory mechanical attributes towards control status (Figures 2.12, 3.7, and 4.5 to 4.8). One must now evaluate the physiological significance of such observations.[1] Given that uncompensated immersion is unfavourable to respiratory mechanics, one must determine whether such perturbations possess sufficient magnitude to either reduce physical power, or efficiency, and in so doing, to present a threat to diver safety.

Air breathing divers work at depths to 50m, where variations in pressure mediate concommitant changes in gas density, and oxygen and nitrogen partial pressures. Such changes interact with ventilatory modifications associated with uncompensated immersion. Divers were therefore studied during upright, immersed exercise and one and six atmospheres absolute (ATA) [1].

[1] 6 ATA = 50m simulated seawater.

It was hypothesised that elastic and flow-resistive respiratory work increments attending uncompensated, immersed exercise will magnify the non-O_2-dependent ventilatory limitations encountered at depth. Use of air delivery at lung *centroid* pressure was envisaged as ameliorating this restraint, thereby enabling divers to improve physical power and efficiency.

5.1 METHODS

5.1.1 Subjects.

Ten male, non-smoking sports divers participated as paid subjects. All were screened by questionnaire for normal lung function history, and previous and/or current physical disorders which would contraindicate hyperbaric exposure. Responses were reviewed by the departmental physician. All subjects were examined by a physician, to obtain medical clearance for diving, in accord with the Worker's Compensation Board of British Columbia. All received subject information packages, and signed informed consent releases.

To avoid age related pulmonary mechanical factors, subjects were all less than 35 years of age (Mittman *et al.* 1965). To ensure subjects were not sedentary, a selection criteria for aerobic capacity was set at greater than 40ml.kg^{-1}.min and determined using the Åstrand-Ryhming prediction (Åstrand and Rhyming 1954). Six subjects took part in all of the previous experiments, and three took part in only the current experiments investigating physical power. The tenth subject participated in the determination of lung *centroid* pressure.

5.1.2 Apparatus

All experiments were performed in the wet chamber of the hypo-hyperbaric chamber complex described in Chapter Two. Water temperature was regulated at 28.6±0.9° C.

A stainless steel, underwater cycle ergometer was designed to fit the confines of the wet chamber. The unit was coupled via an hydraulic pump, to an hydraulic circuit. The power output of the subjects was controlled by adjusting the back pressure acting on the system. A large adjustable seat positioned the subject 45cm above the floor. Legs were extended forward and down to adjustable pedal positions, producing a semi-reclining lower body position. The back was vertical. Feet were protected with diving boots and held loosely to the pedals using cycle toe clips. A quick release belt positioned across the hips secured the subjects to the seat.

A neckseal diving helmet (Kirby Morgan *super-light* , U.S. Divers) was modified to facilitate the following requirements:

(1) attachment to a movable demand regulator, to enable respiratory pressure variations through vertical displacement of the regulator;

(2) equilibration of breathing gas delivery pressure and facial surface pressure, as suggested by Thompson and McCally (1966, 1967);

(3) measurement of airway pressure between the mouth and demand regulator; and

(4) collection of end-tidal gas samples (Figure 5.2).

The fixed regulator was replaced by a brass pipe sealed into the helmet. To this were attached two sample ports: the port proximal to the subject was used to draw end-tidal gas; the other permitted measurement of airway (P_{ao}) pressure. An oronasal mask was secured to the pipe within the helmet, while a length of low resistance, corrugated tubing connected the pipe to the movable regulator.

The airway and oronasal mask were always at the air pressure delivered by the demand regulator. A small mask aperture (2mm diameter) ensured air pressure equilibration between the facial cavity and the mask. Vertical displacement of the movable regulator provided values of P_{ao} and helmet pressure equal to the hydrostatic pressure acting on the body surface within the range of mouth depth, to 10cm below lung *centroid*.

A Conshelf 30 (U.S. Divers) open circuit demand regulator was used as the movable regulator for all trials. Prior to use, the regulator was serviced and cleaned ultra-sonically. Morrison *et al.* (1986) found this same regulator to fulfil tolerance and comfort requirements at depths to 50m (simulated seawater depth) and at ventilations to 70 $Lmin^{-1}$ (Figure 5.3), in accordance with prescribed performance specifications of Morrison and Reimers (1982) and the Department of Energy, U.K. and Norwegian Petroleum Directorate (1984).

Pedal frequency was regulated by the experimenter, using function generators (Hewlett Packard function generator, 3310A; Interstate Electronics Inc. 20MHz sweep function generator, F74) to produce a periodic low pitch tone through the helmet earphones.

End-tidal gas samples were drawn through the diving chamber wall and analysed for CO_2 concentration (Beckman CO_2 medical gas analyser LB-2). Sample lines of two internal diameters were used: 2.117mm for surface trials, and 0.022mm for trials at 50m. These diameters were chosen, with suitable lengths, to maintain the fidelity of the CO_2 signal. Both systems produced exponential CO_2 growth curves during expiration, with a rapid return to a stable baseline on inspiration. It was concluded that gas mixing was minimal prior to analysis.

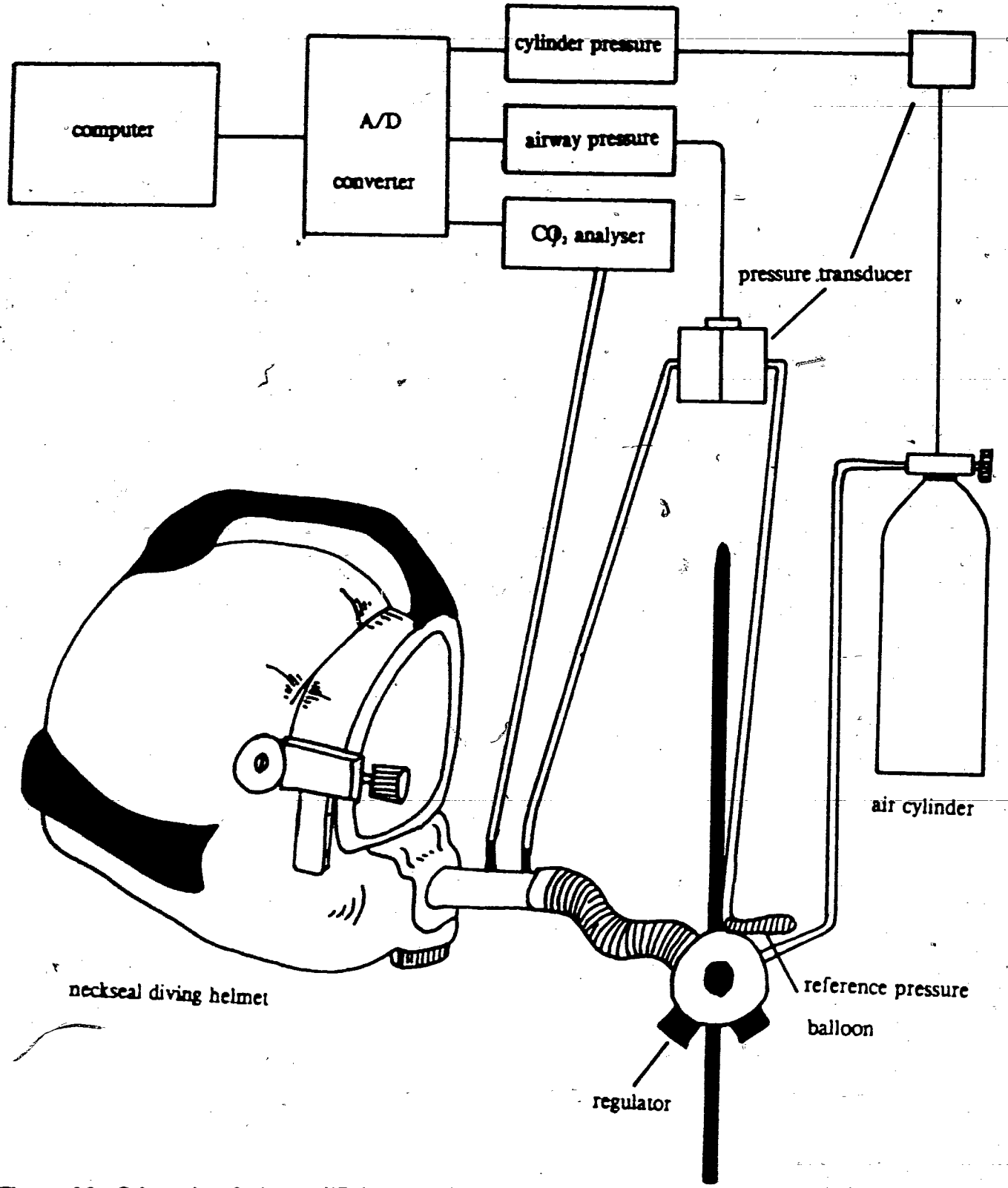


Figure 5.2: Schematic of the modified neckseal diving helmet (Kirby Morgan *super-light* . U.S. Divers) used for all experiments.

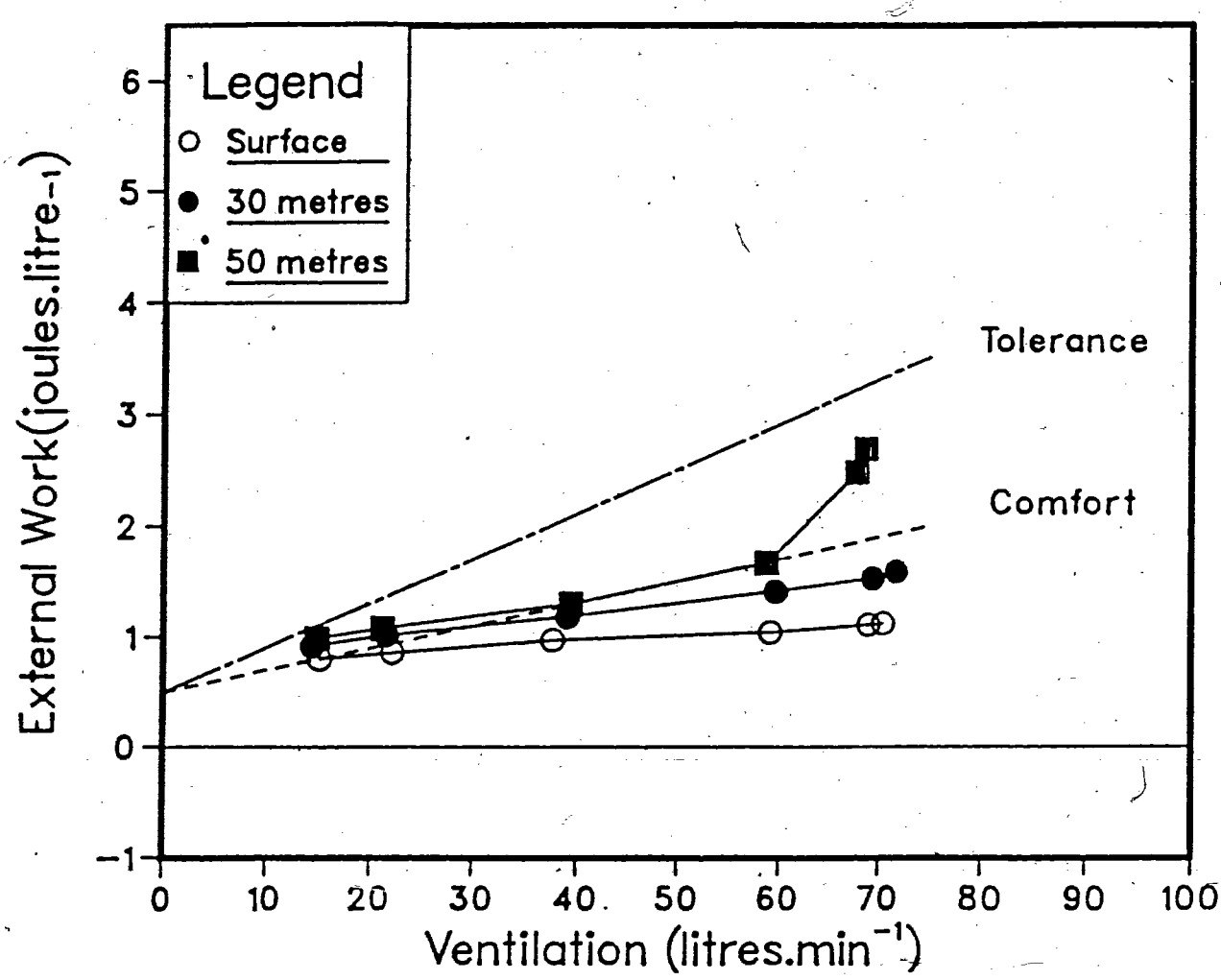


Figure 5.3: **External respiratory work using a Conshelf 30 (U.S. Divers) open circuit demand regulator, at simulated depths ranging from 0.5 to 50m, employing ventilations between 15 to 70 Lmin^{-1}.** [Data reproduced with permission from Morrison *et al.* 1986].

Airway pressures (P_{ao}) were measured as differentials, relative to pressure at the regulator depth. Tygon® tubing connected the airway port to one side of a differential pressure transducer (Validyne MP45 ±3.92 kPa (±40cmH_2O)). The reference side was connected to a 12cm rubber balloon mounted on the regulator, and containing a small volume of air equilibrated with chamber pressure.

Heart rate was recorded directly from exercise electrocardiographs. Electrodes were attached and sealed to the right scapula, and to right and left midaxillary ribs. Immersion traces were taken only after mains power was disconnected from the recorder (Fukuda Denshi Electrocardiograph FD-13).

Air supply to the subjects was provided from two calibrated gas cylinders. At the end of each minute subjects were switched from one cylinder to the next, with cylinder pressures being recorded at the start and end of each minute (Celesco Transducer Production Inc., Model PLD, ± 34.530MPa (±5000PSI)). The cylinder not in use was refilled to a pressure of 13.8MPa (2000PSI) from the reserve cylinder bank. Between the calibrated cylinders and the subject was a small surge bottle, to prevent transient pressure drops at the regulator while changing cylinders. For surface dives 0.566m^3 cylinders were used to enhance pressure measurement accuracy. At 50m, 2.265m^3 cylinders were necessary to supply air requirements during heavy exercise.

Total air supply volumes, including cylinders, pressure hosing and the surge bottle, were 12.01 and 3.68 litres for surface and 50m systems respectively. Ventilation was computed each minute from measurements of system volume, and initial and final cylinder pressures (see: Section 5.1.6).

Electrical connections for the helmet earphones, EKG cables and differential pressure transducer were made through waterproof, high pressure connectors mounted above the water. Pressure signals were amplified externally (Daytronic LVDT model 9130) and passed to a micro-computer (Digital Equipment Corp. PDP-11/03), via an analog to digital convertor (Digital Equipment Corp., LSI-11 RT-1251), for storage. Cylinder pressure and CO_2 signals were similarly sampled and stored. Sampling was at 50Hz. McCall *et al.* (1957) demonstrated peak frequency content of tidal volume and vital capacity manoeuvres to be 3.5 and 4Hz respectively. EKG signals were not stored on computer. A chart recorder (Hewlett Packard 7404A) provided for on-line visual inspection of data and computer back-up.

5.1.3 Calibration.

Calibration was performed daily, or more frequently if apparatus was shut down, or when large time gaps occurred between successive experiments. CO_2 analyser calibration was performed at flow rates encountered during testing. All conditions, except surface trials at mouth pressure, resulted in a positive pressure differential between P_{ao} and the external P_A, producing gas flow. These flow rates were measured with a rotameter, and replicated during calibration to avoid delivery pressure biasing during gas analysis.

P_{ao} calibration was performed at zero and +4.9kPa (+50cmH_2O) using a water manometer (±19.6Pa). Transducer linearity was established by repeat pressure applications over the range −5.88 to +4.90kPa using the above manometer. The system remained linear ($r > 0.999$).

Cylinder pressure calibration was performed with cylinders open to P_A, then pressurised to a variable, but known pressure, around 13.8MPa (2000PSI). Step increments over this range provided linearity checks ($r = 1.0$). Cylinder pressures were also measured using a pressure test gauge (Marsh Instr., type 200−3) connected in parallel with the transducer.

Pilot studies revealed ergonometric load increments in pressure steps of 1.04MPa (150PSI) induced fatigue at about minute ten in three subjects. To accomplish this, subjects had to pedal at different frequencies, according to their level of cardiovascular fitness. These pedal frequencies were determined during preliminary trials. To evaluate absolute external work, the cycle was calibrated to measure ergonometric work. Resistive work imposed by the water was also assessed for each subject.

Ergometer calibration was performed over pressure loads from zero to 10.01MPa[9]. Pedals were placed horizontally with a spirit level along the pedal arm. At each hydraulic pressure load, calibrated weights (±2g) were positioned directly over the pedal axis until movement against the fluid flow commenced. At this point the weight was removed and added to reconfirm the load. Calibration weights, and ergonometric power at four pedal frequencies are shown in Table 5.1. The relationship between fluid pressure and weight to initiate pedal motion was linear ($r = 1.0$).

To evaluate power expended against water resistance, subjects pedalled without load at pre−determined frequencies [10] in chin−depth water. Pedal frequency was controlled by a metronome.

[9] This was the maximal pressure which could be generated within the system.

[10] The frequency found necessary to produce fatigue at about minute ten.

Table 5.1: **Cycle ergometer calibration data and ergonometric power at four pedal frequencies for nine increments in hydraulic pressure.**

PRESSURE* (kPa)	WEIGHT (kg)	POWER AND PEDAL FREQUENCY (W and r.p.m.)			
		70	75	80	85
2072	1.96	24.0	25.7	27.4	29.1
3108	3.40	41.5	44.5	47.4	50.4
4414	4.83	59.0	63.3	67.5	71.7
5449	6.26	76.6	82.0	87.5	93.0
6215	7.70	94.1	100.8	107.5	114.2
7521	9.13	111.6	119.6	127.4	135.5
8287	10.56	129.1	138.4	147.6	156.8
9323	12.00	146.7	157.1	167.6	178.1
10014	12.95	158.3	169.6	180.9	192.3

*The initial test period was conducted at rest so only nine increments are shown.

After five minutes, a one minute gas sample was collected in a meteorological balloon, and analysed for CO_2 and O_2 concentrations (Beckman CO_2 medical gas analyser LB-2, Ametek Applied Electrochemistry O_2 analyser S-3A/I) and volume (American Meter Division, DTM-325). From unloaded $\dot{V}O_2$ data, approximations of external resistive power were derived (Table 5.2A). The $\dot{V}O_2$ and total power output of the final minute of exercise were also approximated using these data (Table 5.2B).

At each increment of ergonometric load, power output was obtained from the sum of unloaded and ergonometric power. When immersed at 1 ATA (with demand regulator at mouth depth) terminal power averaged 333.0±24.6W, while terminal $\dot{V}O_2$ averaged 4.25±0.29 $l.min^{-1}$.

5.1.4 Procedures.

All procedures were reviewed and approved by Simon Fraser University Ethics Review Committee. Subjects were healthy, and physically active divers.

Each subject took part in five preliminary and five experimental dives. Preliminary dives served several purposes:

(1) to act as training to ensure subjects did not improve endurance as a result of the experiments;

(2) to familiarise subjects with the cycle ergometer, underwater cycling, the diving helmet and other experimental apparatus;

(3) to train the subjects to pedal in time with the helmet tone;

(4) to determine the pedal frequency necessary to fatigue subjects at minute ten; and

(5) to familiarise and train subjects in the use of a subjective scale to rate respiratory comfort.

Subjects were asked to rate respiratory comfort each minute using a scale from zero to five (Table 5.3). During the final preliminary trial subjects rated comfort when the regulator was randomly and covertly positioned at mouth level, lung *centroid* level and at 10cm above and below lung *centroid* . Rating took place at rest, and cycling at a mean total power output of 279.6±22.8W.

Lung *centroid* position was taken at 13.5cm below the sternal notch (Sections 2.2, 3.4). Seven current subjects took part in lung *centroid* determination studies. Mean *centroid* position for these subjects was 13.8cm (SEM=1.08) below sternal notch.

Table 5.2A: **Pedal frequency, oxygen consumption and power output during unloaded pedalling.**

SUBJECT	PEDAL FREQUENCY (r.p.m.)	$\dot{V}O_2$ ($l.min^{-1}$)	POWER (W)
1	85	1.79	150.4
2	80	2.11	177.6
3	85	1.82	153.3
4	85	2.03	170.6
5	80	2.11	177.6
6	70	1.65	138.9
7	85	1.59	133.8
8	75	1.76	148.0
9	80	1.35	113.8
10	75	2.01	168.7
$\bar{x}$	80	1.82	153.3
SEM	1.7	0.08	6.6

$\dot{V}O_2$ = oxygen consumption.
Calculation of power from $\dot{V}O_2$ was performed using the assumption of 25% metabolic efficiency.

Table 5.2B: **Oxygen consumption and power output for the final minute of exercise.**

SUBJECT	PEDAL FREQUENCY (r.p.m.)	$\dot{V}O_2$ ($l.min^{-1}$)	POWER (W)
1	85	4.37	342.7
2	80	4.46	350.2
3	85	4.40	345.6
4	85	4.60	362.8
5	80	4.55	358.6
6	70	3.78	293.1
7	85	4.17	326.1
8	75	4.07	317.6
9	80	3.80	294.7
10	75	4.32	338.4
$\bar{x}$	80	4.25	333.0
SEM	1.7	0.09	7.8

$\dot{V}O_2$ = oxygen consumption.

Terminal power was approximated using ergometer calibration data and $\dot{V}O_2$ during unloaded pedalling. Terminal $\dot{V}O_2$ was approximated from terminal power using the following assumptions: (a) resting $\dot{V}O_2$ = 0.3 $l.min^{-1}$, (b) respiratory quotient = 0.82, (c) 25% metabolic efficiency.

$\dot{V}O_2$ approximations for subjects 5, 8 and 10 exceeded values quoted in Table 5.5. The mean overestimation is 14%, and was probably within the limits of accuracy for such approximations.

Table 5.3: **Subjective rating scale for perceived respiratory comfort.**

SCALE	VERBAL EQUIVALENT	MEANING
0	very, very light	just noticeable.
1	light	very little effort, easy to breathe.
2	moderate	at the limit of comfort, would prefer no greater.
3	heavy	hard but acceptable.
4	very heavy	at the limit of tolerance, but could sustain.
5	very, very heavy	very uncomfortable, interferes with breathing, could sustain for a few seconds.

Scale is identical to that used by Morrison and Wood (1986).

With deeper regulator positions, there was a tendency for the helmet to lift off the head. Similar problems encountered in pressurised flight helmets (Dhenin 1978) were solved by elimination of the airspace on top of the head. Helmet lift had no associated discomfort and was not countered, but should be considered for industrial applications.

Experimental dives on submerged subjects were conducted at a surface pressure of 1 ATA, and at a simulated seawater depth of 50m (6 ATA). Subjects breathed air at each of two pressures (mouth pressure and lung *centroid* pressure (P_{LC})) at both the surface and 50m dives. Subjects were initially randomly assigned to experimental conditions. However, to equate the number of subjects experiencing mouth and P_{LC} air delivery pressures on the first surface and 50m dive, subjects tested later in the study were assigned a specific test sequence. Due to technical and support staff arrangements, no attempt was made to balance the diving depth sequence. The fifth dive was performed at 1 ATA with $\dot{V}O_2$ being measured during unloaded pedalling.

Time to complete all dives (plus repeats due to occasional equipment failure) averaged 18±3 days (including week-ends) for nine subjects. Subjects generally rested for 24hr between trials. Subject two took longer due to injury, and was required to undertake additional training trials before recommencing the experiments.

During experimental dives one to four, subjects started with a one minute rest, then cycled at the appropriate and constant pedal frequency until fatigued, or until ten minutes had elapsed. Workload was incremented at the end of each minute (Table 5.1) from outside the chamber. A computer generated signal to the dive tender prompted signalling for perceived comfort rating at the end of each minute. CO_2 and P_{aO} signals were sampled continuously. EKG traces were taken over the last ten seconds of each minute. Cylinder pressure was sampled by computer at the start and end of each minute. Subjects were free to terminate experiments due to fatigue, respiratory discomfort, narcosis, or anxiety due to any cause.

Immediately after each dive subjects were questioned regarding subjective impressions during the experiment.

(1) Why did you stop exercising? (a) general fatigue, (b) respiratory discomfort, (c) respiratory fatigue, (d) local muscle fatigue, (e) narcosis, (f) time had elapsed.

(2) Which did you find easier? (a) inspiration, (b) expiration, (c) neither.

(3) Did you find inspiration a limiting factor? (a) yes, (b) no.

(4) Did you find expiration a limiting factor? (a) yes, (b) no.

Subjects were given the questionnaire prior to experimentation, and asked to be aware of each question during the experiment.

5.1.5 Experimental safety.

Prior to commencing trials subjects were given information regarding the risks of compressed air diving. After 50m dives subjects received decompression checklists and medical alert bracelets. The latter were worn for 24 hours.

During every experiment a cardiovascular *crash cart* was positioned in the laboratory. A physician was in attendance during all 50m dives. Following such dives a mandatory one hour *bends watch* within the environmental laboratory, was enforced for all divers and chamber operators. The physician was always within three minutes of the chamber during this period.

Two chamber operators controlled the diving facility. Two diver tenders accompanied each subject, one in the water and one inside the entry lock, where an emergency harness and pulley block was secured. This second tender was generally a subject for a second experiment, and was in verbal contact with the outside operators, except during the subject change-over.

Subjects were familiarised with the helmet's quick release mechanism in case of air supply failure. Two emergency demand regulators were positioned beside the ergometer at all times. During every experiment the diver and tender reciprocally signalled "all is well" at least once per minute.

Decompressions took place inside the main (dry) chamber. The chamber facility was equipped with an external Environmental Control System, which removed expired CO_2 from the air, and controlled air temperature and humidity. Divers dried and dressed in fire retardant clothing. All decompressions were monitored on closed circuit television and adhered to the 1986 Canadian Forces air diving tables and procedures. Using Table 2 of these procedures, divers were brought to 30 feet using air, with decompression stops at 60, 50 and 40 feet. The final decompression stop was at 30 feet. Immediately upon reaching this point, divers commenced O_2 breathing, which continued until they reached the surface. During O_2 use divers wore oronasal masks equipped with an overboard dump. Emergency air masks were also available.

5.1.6 Calculations.

Calculations were performed by computer programme. Data acquisition, calibration and analysis programmes were obtained from Morrison and Wood (1986).

Ventilation.

Minute ventilations ($\dot{V}_I$) were computed as dry volumes at standard pressure ($\dot{V}_I SP$) from changes in air cylinder pressure. Since the cylinder volume was known, $\dot{V}_I SP$ was proportional to the change in cylinder pressure. Air cylinders were kept in a water bath, at room temperature, to minimise adiabatic temperature change during discharge and refilling, but pressure data still required correction to provide isothermal expansion. A series of tests were performed in which the effect of temperature on cylinder pressure change, over one minute, was studied (Morrison and Wood 1986). The results produced linear correction coefficients for both cylinder sizes. These coefficients were used within the present calculations.

$$\dot{V}_I BTPS = \dot{V}_I SP[760/(PD + P_A - 47.1)] \cdot [310/(273+T)] \qquad \textbf{Equation 1}$$

where:

PD = (chamber depth + regulator depth)(mmHg)

T = temperature of cylinder bath.

P_A = barometric pressure (mmHg).

End-tidal CO_2 tension.

CO_2 concentrations were computed as three point, moving averages to identify peak (end-tidal) and trough (baseline) concentrations during the middle 30 seconds of each recording minute. During this time the first and last peaks were discarded, the remainder were averaged.

$$P_{ET}CO_2 = F_{ET}CO_2 [P_A / (P_A - P_{H_2O})] (P_A + PD - 47.1) \quad \textbf{Equation 2}$$

where:

$F_{ET}CO_2$ = mean end-tidal fractional CO_2 concentration.

P_{H_2O} = water vapour pressure (mmHg).

Airway pressure.

Maximum and minimum values for P_{ao} were derived from three point moving averages. The mean of these values was obtained over the middle 30 seconds of each minute. P_{ao} was then derived as the difference between mean peak and mean trough pressures for each minute.

5.1.7 Analysis.

Analysis was based on a repeated measures experimental design containing two within-subjects factors (experimental conditions and workload). Perceived respiratory comfort was regarded as parametric data since it possessed both interval and ratio characteristics. *A priori* probability significance was set at the 0.05 level. Multivariate analysis of variance was used to search for significant interactions between experimental conditions and workloads. Analysis was also performed using a paired model, following the attainment of a significant overall *F* statistic (Hotelling's T-squared correlated), since variability between subjects was large, while changes between conditions were consistently in the same direction. Paired comparisons were made at each work rate, and for the combined data of all work rates, to search for differences between data obtained at 1 ATA and 6 ATA, and for differences when breathing air supplied at P_{LC} and mouth pressure. Analysis of variance lacks the necessary sensitivity to analyse data with these characteristics. Appendix Three contains the summary tables for the MANOVA analysis and the Hotelling's T-squared comparisons.

5.2 RESULTS

5.2.1 Characteristics of subjects.

Physical characteristics of subjects are detailed in Table 5.5[11]. All compressions and decompressions proceeded without incident. No post-dive decompression symptoms were observed.

5.2.2 Subjective evaluation of hydrostatic pressure compensation.

Using the respiratory rating scale (Table 5.3), subjects evaluated comfort at four regulator depths at rest, and cycling at a mean total power output of 279.6 W. At all breathing pressures subjects perceived respiration to be significantly more comfortable ($p<0.05$) at rest, than during exercise (Figure 5.4).

Analysis of the combined exercise and resting data showed that differences in perceived comfort between P_{LC} and P_{LC} minus 0.98kPa air delivery, were non-significant ($p>0.05$, $\phi=0.65$). However, subjects perceived respiration to be significantly more comfortable at these air supply pressures than at either mouth or P_{LC} plus 0.98kPa ($p<0.05$[12]).

During incremental exercise trials, subjects consistently perceived respiration to be more comfortable for all levels of exercise, and at both diving depths, when the regulator was positioned at lung *centroid* depth (Figure 5.5)[13]. Both at 1 and 6 ATA subjects rated respiration at the final minute of exercise, as "hard but acceptable", using mouth pressure delivery, and at the "limit of comfort" with P_{LC} air provision[14].

The paired data for all levels of exercise was examined for differences in perceived respiratory comfort associated with a change in absolute pressure (1 ATA versus 6 ATA), and for differences associated with altered air delivery pressure (mouth versus P_{LC} air provision). Subjects perceived respiration to be significantly less comfortable at 6 ATA than at 1 ATA ($p<0.05$). Subjects rated P_{LC} air

---

[11] Subject numbers in Table 5.5 remain consistent throughout this chapter.

[12] MANOVA and Hotellings paired analysis summary tables are contained in Appendix Three. An asterisk is used in figures to represent points of significant difference at the 0.05 level.

[13] All plots during incremental exercise trials contain time on the abcissa. Times represent points of sequential workload increments averaging 0%, 54%, 60%, 66%, 72%, 78%, 84%, 90%, 96% and 100% of the terminal workload (respectively).

[14] Overall MANOVA $F = 6909.23$ (5,75), $p<0.05$.

Table 5.4: **Physical characteristics of subjects.**

SUBJECT	Age (years)	Years diving	Mass (kg)	Maximal $\dot{V}O_2$ ($l.min^{-1}$)
1	27	13	70.7	4.70*
2	24	2	81.5	4.75*
3	26	10	71.9	4.94*
4	23	1	71.7	4.80*
5	23	1	84.1	4.00
6	33	13	77.5	4.75*
7	27	1	82.0	4.41*
8	28	3	70.0	3.60*
9	26	5	75.4	3.90
10	26	4	89.3	3.70
$\bar{x}$	26.3	5.3	77.4	4.36
SEM	0.9	1.5	2.1	0.16

$\dot{V}O_2$ = oxygen consumption.

* = $\dot{V}O_2$ measured using breath-by-breath analysis for a separate investigation conducted simultaneously, other $\dot{V}O_2$s were predicted (Åstrand-Ryhming prediction).

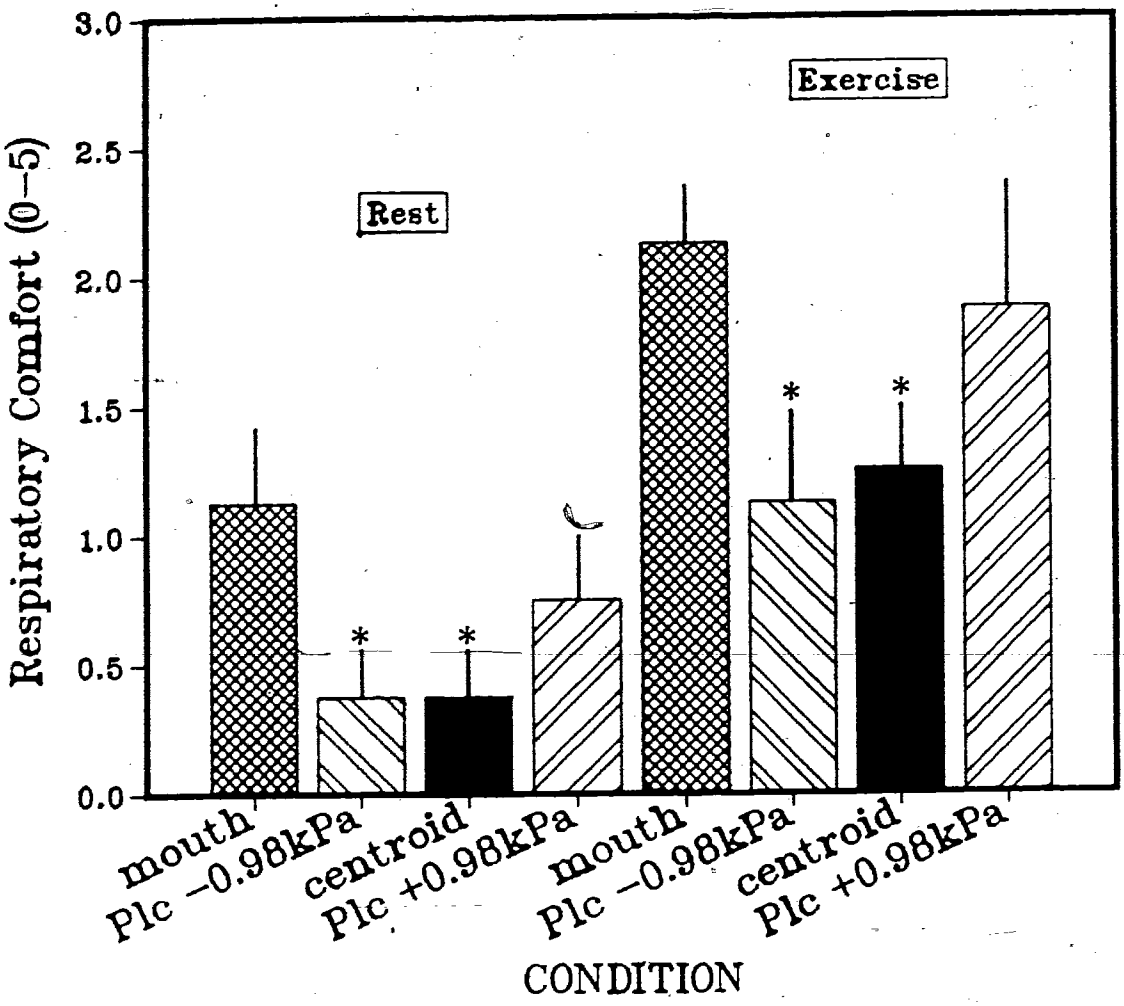


Figure 5.4: Assessment of respiratory comfort rating for four regulator depths, with subjects resting and cycling at 279.6±22.8 Watts. [Data represent means and standard errors.]

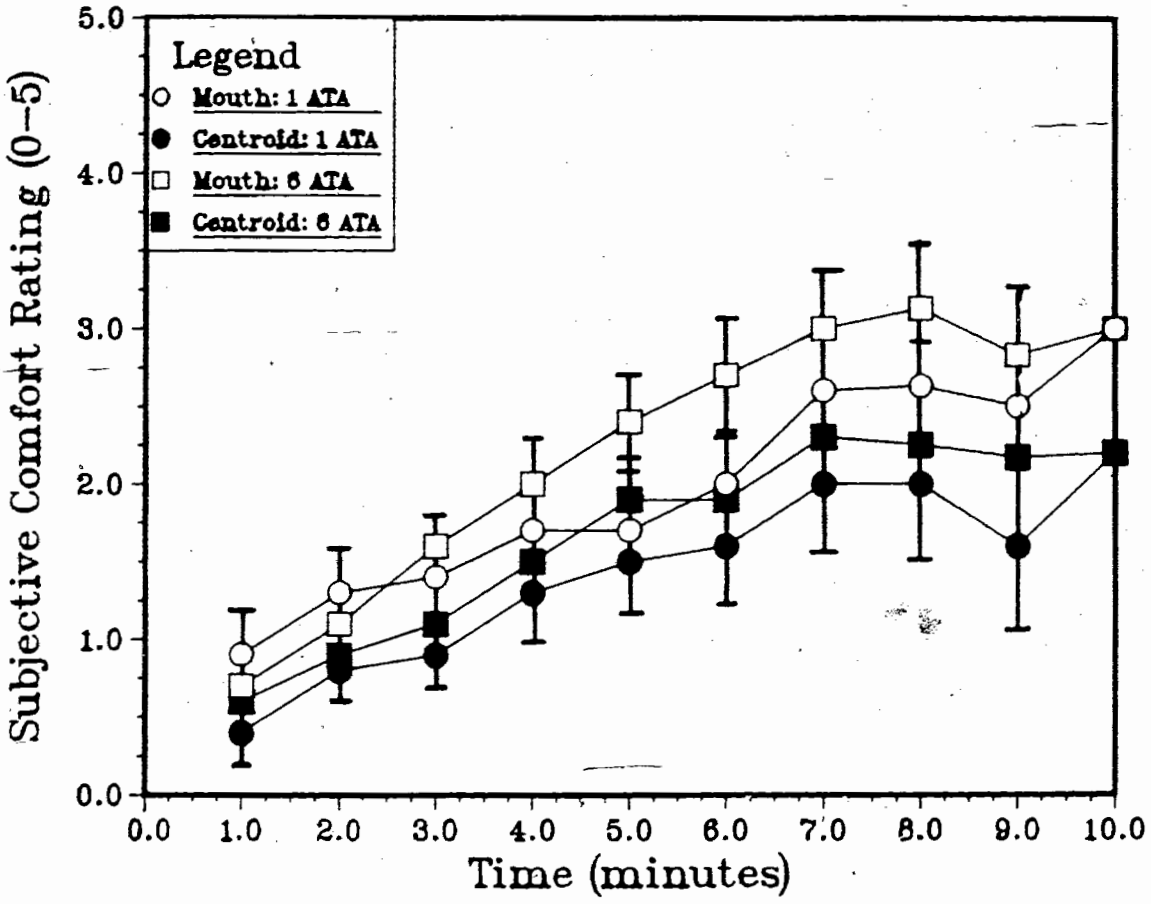


Figure 5.5: Assessment of respiratory comfort during exercise at 1 and 6ATA, with air delivered at mouth and lung *centroid* pressures . [Data represent means and standard errors.]

provision to be more comfortable ($p<0.05$). This latter trend was apparent during trials at both absolute pressures.

Post-experimental questionnaires (Section 5.1.4) were designed to evaluate perceived differences between inspiratory and expiratory effort. Subjects reported inspiration to be a limiting factor when receiving air at mouth pressure. Six subjects reported inspiration as limiting at both 1 and 6 ATA, while only 1 felt compromised when breathing air at P_{LC} at 1 and 6 ATA.

No subjects ceased exercising specifically because of dyspnea or perceived respiratory fatigue. Subject six failed to complete 10 minutes at 1 ATA using mouth pressure delivery. At 6 ATA five trials were terminated due to narcosis; two at mouth pressure and three at P_{LC} air delivery.

When questioned after completing both deep dives, 7 subjects reported feeling better when breathing air at P_{LC}[15]. They reported being less narcotic, more aware of the physical work and passage of time, and had better recall of events upon completion of the trial. At mouth pressure air provision, two subjects had to be helped off the cycle, four reported impending loss of consciousness, and five reported visual, auditory and mental aberrations. All symptoms disappeared immediately the helmet was removed. One of these subjects regained sensual acuity within one minute of trial termination, as evidenced by the capacity to locate and refit a helmet locking nut (10mm diameter) which had dislodged and fallen to the bottom of the chamber. Only one subject perceived narcosis to be greater when breathing air at P_{LC}.

5.2.3 Minute ventilation.

Minute ventilation ($\dot{V}_I$) was significantly reduced at 6 ATA when compared across workloads ($p<0.05$), in accord with previously reported observations (Lanphier 1963, Wood 1963, Bradley *et al.* 1971, Morrison *et al.* 1976, Thalmann *et al.* 1979, Hesser *et al.* 1981, Morrison and Wood 1986) (Figure 5.6).

Subjects achieved significantly greater exercise $\dot{V}_I$ when air was supplied at P_{LC}, when compared across workloads ($p<0.05$). This trend was primarily attributed to differences observed at 6 ATA, where $\dot{V}_I$ for P_{LC} air supply exceeded that with mouth pressure air provision, at all points of measurement. These differences were significant for all points of measurement beyond minute three. Similar

[15] Regulator positioning was performed covertly, but most subjects were able to judge regulator position from inspiratory effort.

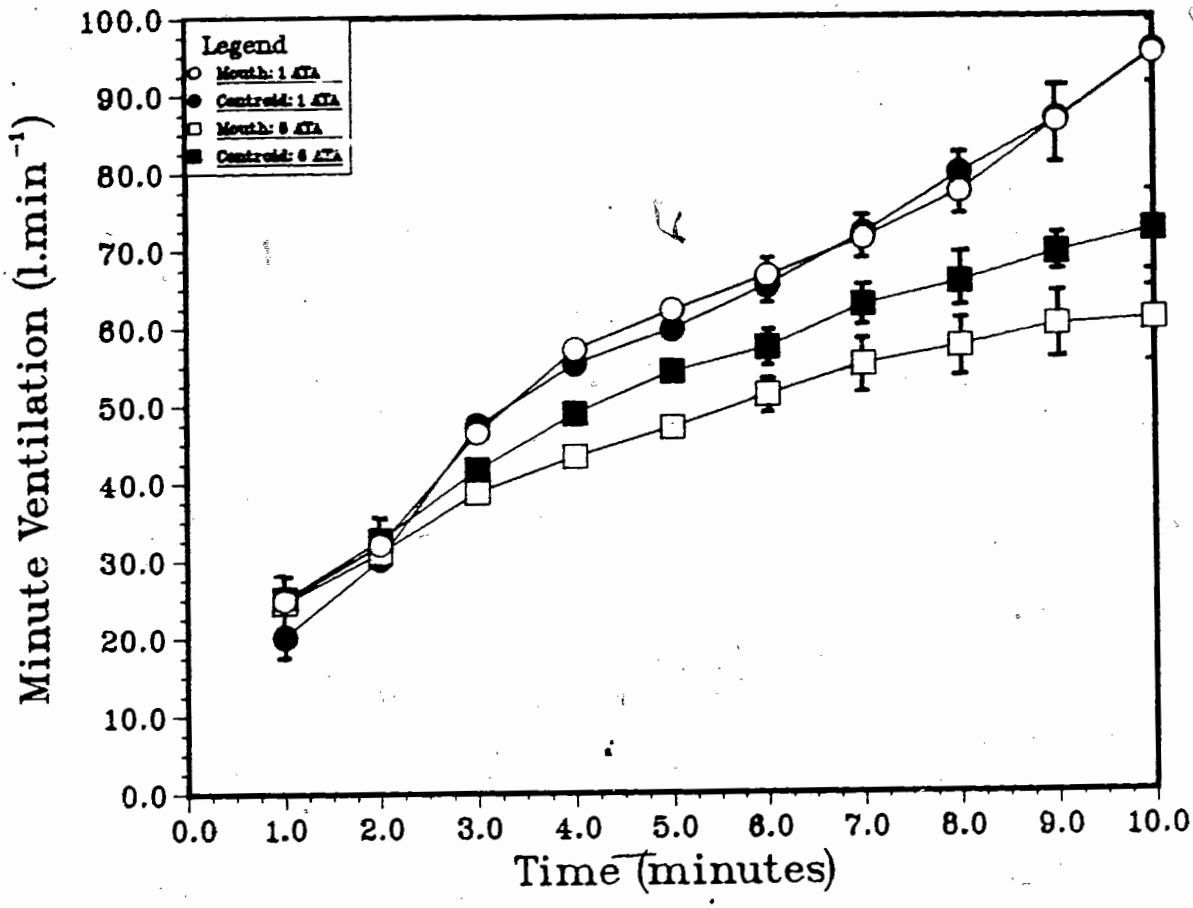


Figure 5.6: **Minute ventilation during exercise at 1 and 6 ATA using air delivered at mouth pressure and lung *centroid* pressure.** [Data represent means and standard errors.]

differences were not apparent at 1 ATA.

5.2.4 End tidal carbon dioxide tension.

During 1 ATA trials $P_{ET}CO_2$ followed an inverted-U relation with work increments, as reported in the literature (Wood and Bryan 1971, Fagraeus and Linnarsson 1973, Linnarsson and Fagraeus 1976, Thalmann *et al.* 1979, Morrison and Wood 1986) (Figure 5.7). Resting $P_{ET}CO_2$ at 1ATA averaged 38.43 and 38.49 mmHg for mouth pressure and P_{LC} respectively, representing expected normal observations (Cotes 1979) and confirming technical validity. Exercise $P_{ET}CO_2$ similarly matched reported observations, despite the added respiratory dead space (Jarrett 1966, Jones *et al.* 1979, Macdonald and Pilmanis 1981).

At 6 ATA $P_{ET}CO_2$ continued to rise throughout the course of the experiments. All subjects, except subject nine, generally experienced a progressive $P_{ET}CO_2$ elevation at each level of exercise, supporting observations from previous investigations (Wood and Bryan 1971, Fagraeus and Linnarsson 1973, Linnarsson and Fagraeus 1976, Thalmann *et al.* 1979, Morrison and Wood 1986). Subjects completed exercise with an average $P_{ET}CO_2$ of 62.7 and 62.6 mmHg for mouth and P_{LC} air delivery respectively. Differences between $P_{ET}CO_2$ at 1 and 6 ATA were significant ($p<0.05$), as were the differences between the two air supply pressures, when paired data was analysed collectively ($p<0.05$). The latter observation was created by greater CO_2 retention during exercise at 1 ATA when air was provided at P_{LC}. This trend was opposite to that observed at 6 ATA, where greater CO_2 retention was associated with mouth pressure air delivery.

5.2.5 Heart rate.

Bradycardia is normally observed under conditions of raised ambient pressure (Shilling *et al.* 1936, Hesser *et al.* 1968, Bradley *et al.* 1971, Macdonald and Pilmanis 1981). In the current investigation a significant bradycardia was observed at 6 ATA ($p<0.05$), when paired data obtained with mouth and P_{LC} air delivery were analysed collectively (Figure 5.8).

In general, heart rates with mouth pressure air provision were greater than values observed when breathing air at P_{LC}. When analysed collectively over workload and absolute pressure, the mean difference was 5b.min^{-1}, and was significant at the 0.05 level.

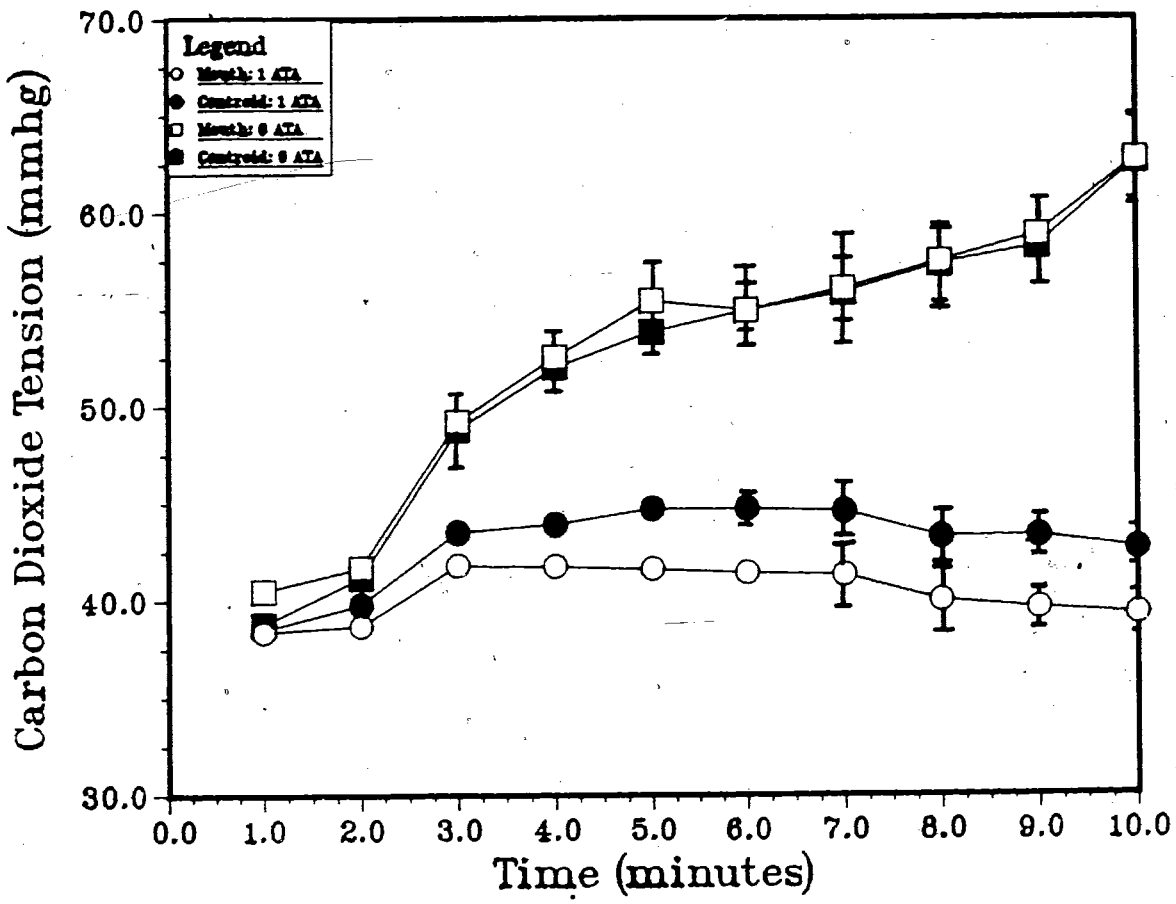


Figure 5.7: **End tidal carbon dioxide tension during exercise at 1 and 6 ATA, using mouth pressure and lung *centroid* pressure air delivery.** [Carbon dioxide tension was obtained as an average of the values recorded for the middle 30 seconds of each minute of exercise. Data represent means and standard errors.]

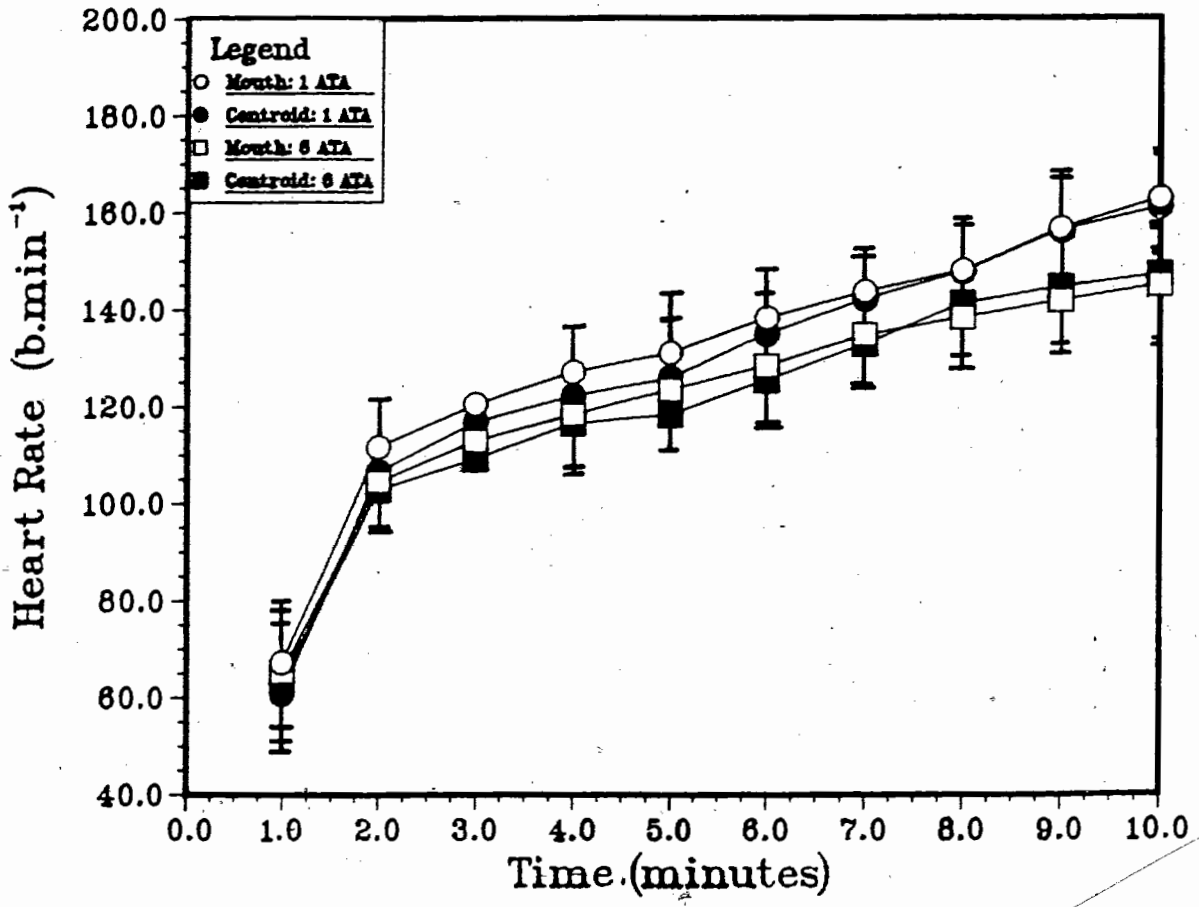


Figure 5.8: **Heart rate during exercise at 1 and 6 ATA, using mouth and lung *centroid* air delivery pressure.** [Data represent means and standard errors.]

5.2.6 Airway pressure.

Airway pressure was monitored to provide a gauge of inspiratory effort required to overcome the external resistance of the breathing apparatus. Observations at 1 and 6 ATA revealed a progressive airway pressure increase with workload (Figure 5.9). This trend reflected increments in ventilation, as the workload was elevated.

At 6 ATA, the airway pressure was significantly greater than that observed at 1 ATA when compared across workloads ($p<0.05$).

5.2.7 Differences at equivalent ventilations.

Since the major interest in this study was the investigation of ventilation and respiratory effort, perceived breathing comfort, heart rate, end-tidal CO_2 tension and airway pressure were compared between conditions using minute ventilation as the independent variable. Because $\dot{V}_I$ was uncontrolled, it was only possible to analyse unpaired data (using MANOVA) where minute ventilations were approximately equivalent. Ten sites of equivalent ventilation were obtained for subjects at 1 ATA, and seven locations were similarly achieved at 6 ATA. All subjects provided data at these points, except for the final ventilatory level at 1 ATA, and the final point at 6 ATA when breathing air at P_{LC}.

From MANOVA analysis of all variables at equivalent minute ventilations at 1 ATA, it was found that differences at each air supply pressure were non-significant, when analysed collectively and at each ventilatory level[16] ($p>0.05$; $\phi=0.52$). At 6 ATA, comparisons between data measured at mouth and P_{LC} air delivery, produced a significant divergence for perceived breathing comfort and heart rate, both when analysed collectively and at different levels of ventilation[17] ($p<0.05$).

[16] Overall MANOVA F = 9859.73 (4,86), $p<0.05$. This indicates a significant time interaction only.

[17] Overall MANOVA F = 3772.83 (4,55), $p<0.05$.

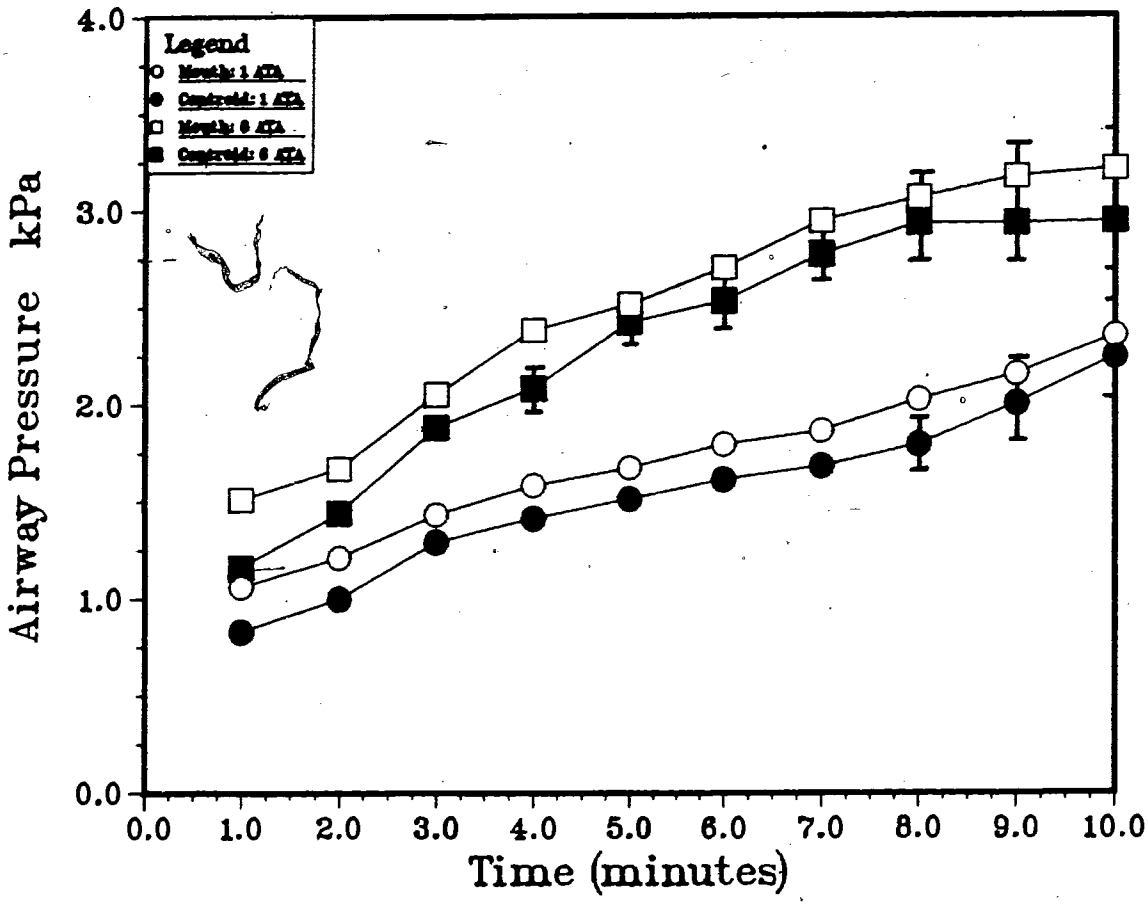


Figure 5.9: **Airway pressure during exercise at 1 and 6 ATA, using mouth and lung *centroid* air delivery pressures.** [Pressures were obtained from the difference between the average peak and trough pressures, obtained over the middle 30 seconds of each minute of exercise. Data represent means and standard errors.]

Perceived respiratory comfort.

At 6 ATA respiration was perceived to be significantly more comfortable at P_{LC} when data, analysed across equivalent ventilations, were compared with those at mouth pressure air supply (6 ATA, $p<0.05$). Incremental differences were significant at minute ventilations of 42.5, 47.9, 57.4 and 61.9 $l.min^{-1}$ ($p<0.05$, Figure 5.10), when breathing air at P_{LC}. The last two ventilatory levels at mouth pressure produced comfort ratings above three, indicating that respiration was "hard but acceptable". At equivalent ventilations, when subjects breathed air at P_{LC}, respiration was rated as requiring "very little effort, and easy to breathe", and at the "limit of comfort" respectively for the final two measurement points.

Since ventilations were only equivalent between air delivery pressures, at points up to 61.9 $l.min^{-1}$, assessments of respiratory comfort beyond that ventilation are not provided in Figure 5.10. These data are illustrated in Figure 5.11, and provide information up to ventilations of 72.4 $l.min^{-1}$, thereby permitting comparison with the observations of Morrison *et al.* (1986, Figure 5.3) obtained from the demand regulator used in the current investigation. It is noted that a plateau of comfort was apparent for both air supply pressures, at the higher ventilatory levels. This may simply reflect subject attrition at the higher workloads, rather than the attainment of a stable level of breathing comfort. Though ventilations beyond 60.9 $l.min^{-1}$ were not attained using mouth pressure air supply, it appears that differences between comfort ratings at the two air delivery pressures, are maintained even at high ventilatory levels.

Heart Rate.

When heart rate data at 6 ATA were analysed across levels of equivalent ventilation, the results obtained when subjects breathed air at P_{LC} were significantly lower than with mouth pressure air provision ($p<0.05$, MANOVA). Incremental differences between heart rates recorded at equivalent ventilations, when using the two air supply pressures, were significant for ventilations in excess of 40 $l.min^{-1}$ (Figure 5.12).

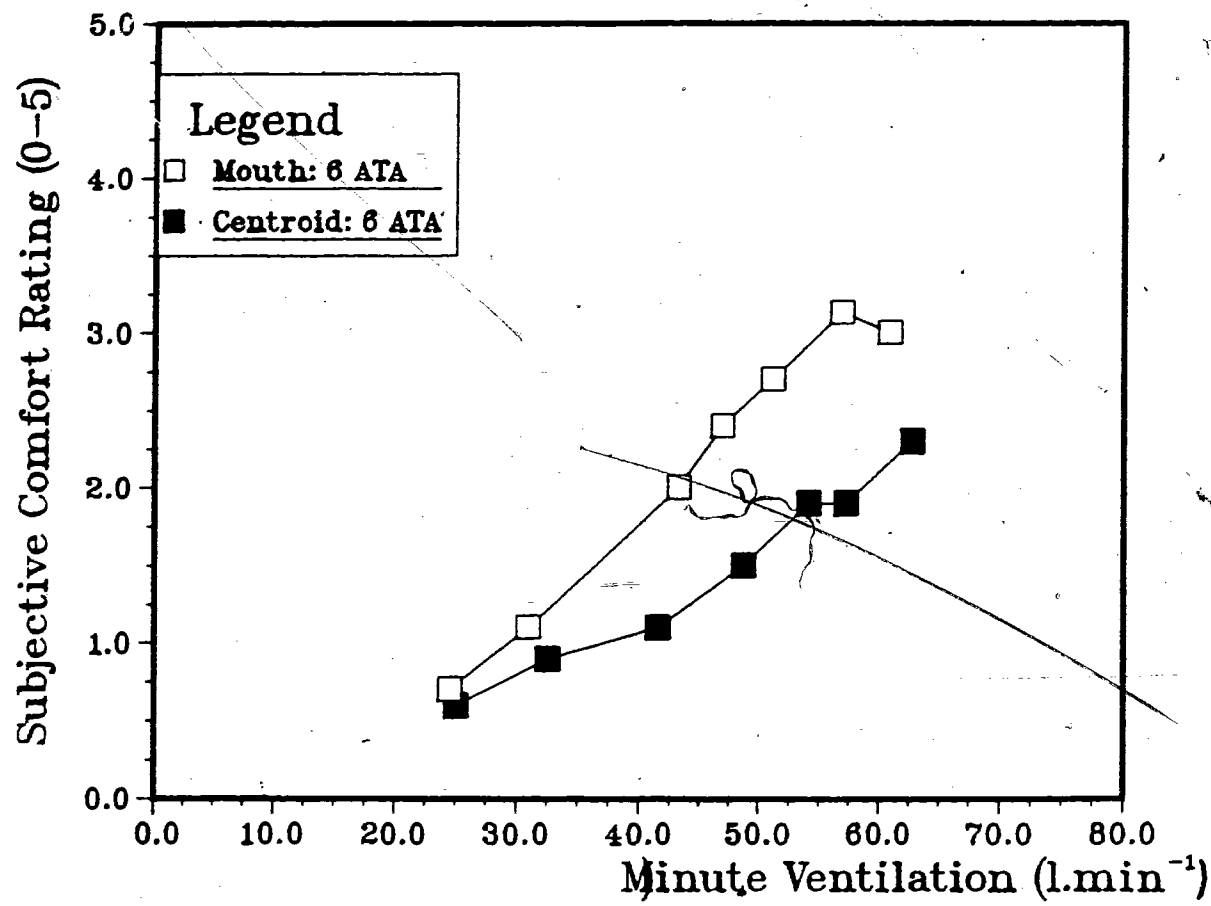


Figure 5.10: **Assessment of respiratory comfort for matched minute ventilations, using air delivered at mouth and lung *centroid* pressures at 6 ATA.** [Data represent mean values only. Standard errors are not indicated as each was previously reported in Figures 5.5 and 5.6.]

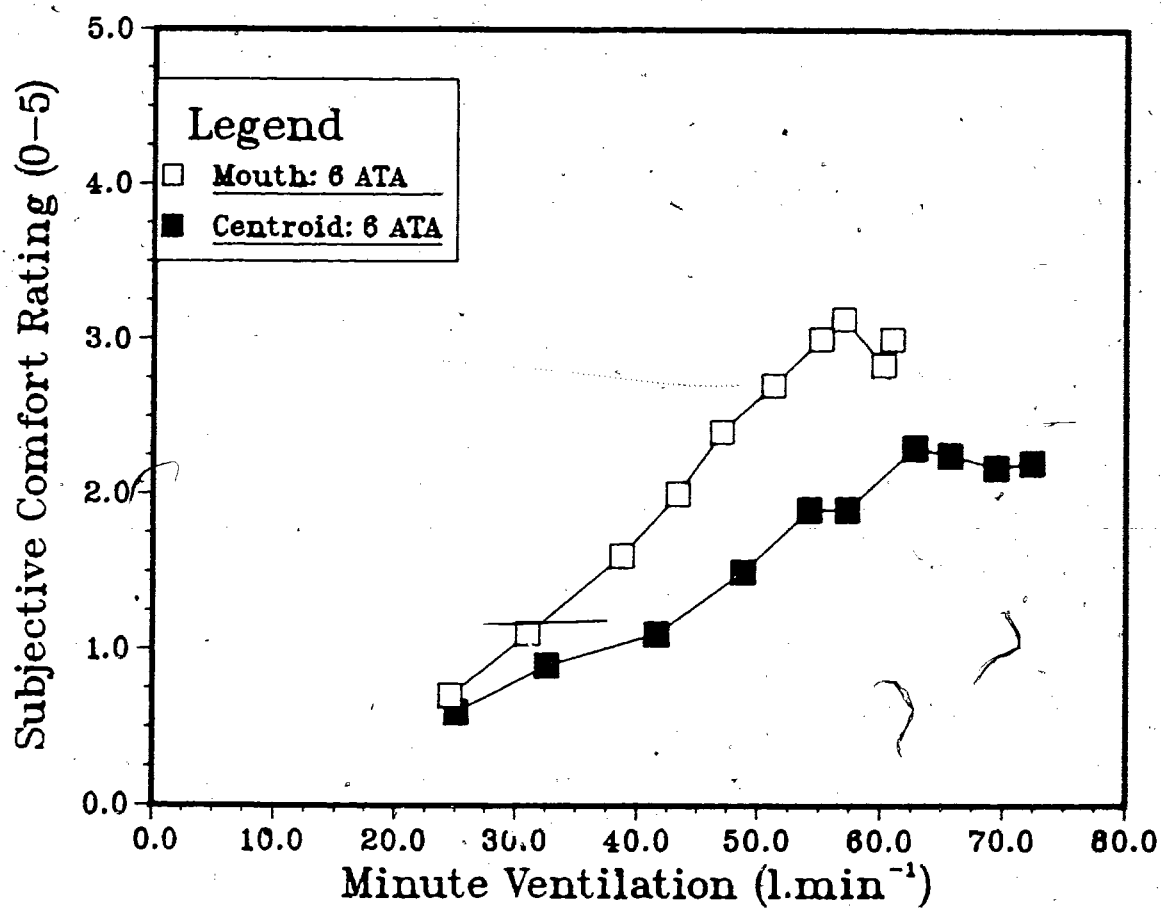


Figure 5.11: Assessment of respiratory comfort at all minute ventilations, using air provided at mouth and lung *centroid* pressures at 6 ATA. [Data represent mean values only. Standard errors are not indicated as each was previously reported in Figures 5.5 and 5.6.]

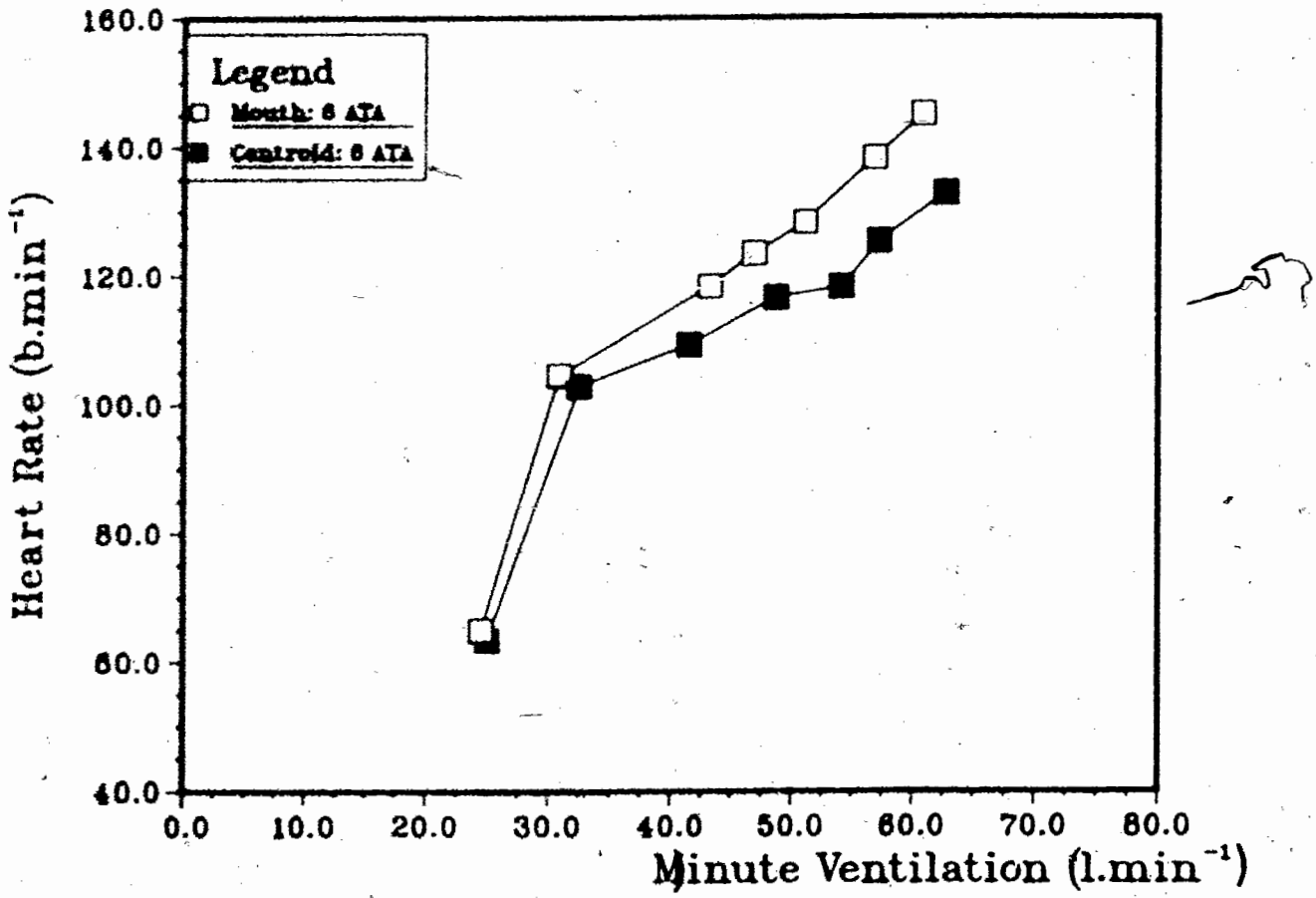


Figure 5.12: Heart rates for matched minute ventilations, using air delivered at mouth and lung *centroid* pressures at 6 ATA. [Data represent mean values only. Standard errors are not indicated as each was previously reported in Figures 5.6 and 5.8.]

5.2.8 Perceived respiratory comfort at equivalent airway pressures.

At 1 and 6 ATA, subjects tended to rate similar airway (mouth) pressures as equally comfortable at the lower work rates[11] regardless of air supply pressure (Figure 5.13). At higher work rates, equivalent airway pressures were perceived to be less comfortable when air was provided at mouth pressure. At 6 ATA there appeared to be a shift in the position of the comfort–airway pressure relationship, towards a lower (more comfortable) rating for any given airway pressure. This observation supports the comfort rating index reproducibility, with and without the influence of narcosis (Figure 5.13).

[11] Work rate is not recorded on either the abcissa or the ordinate, however, moving to the right, each point represents a sequential increment in work rate, and serial points are joined by a straight line.

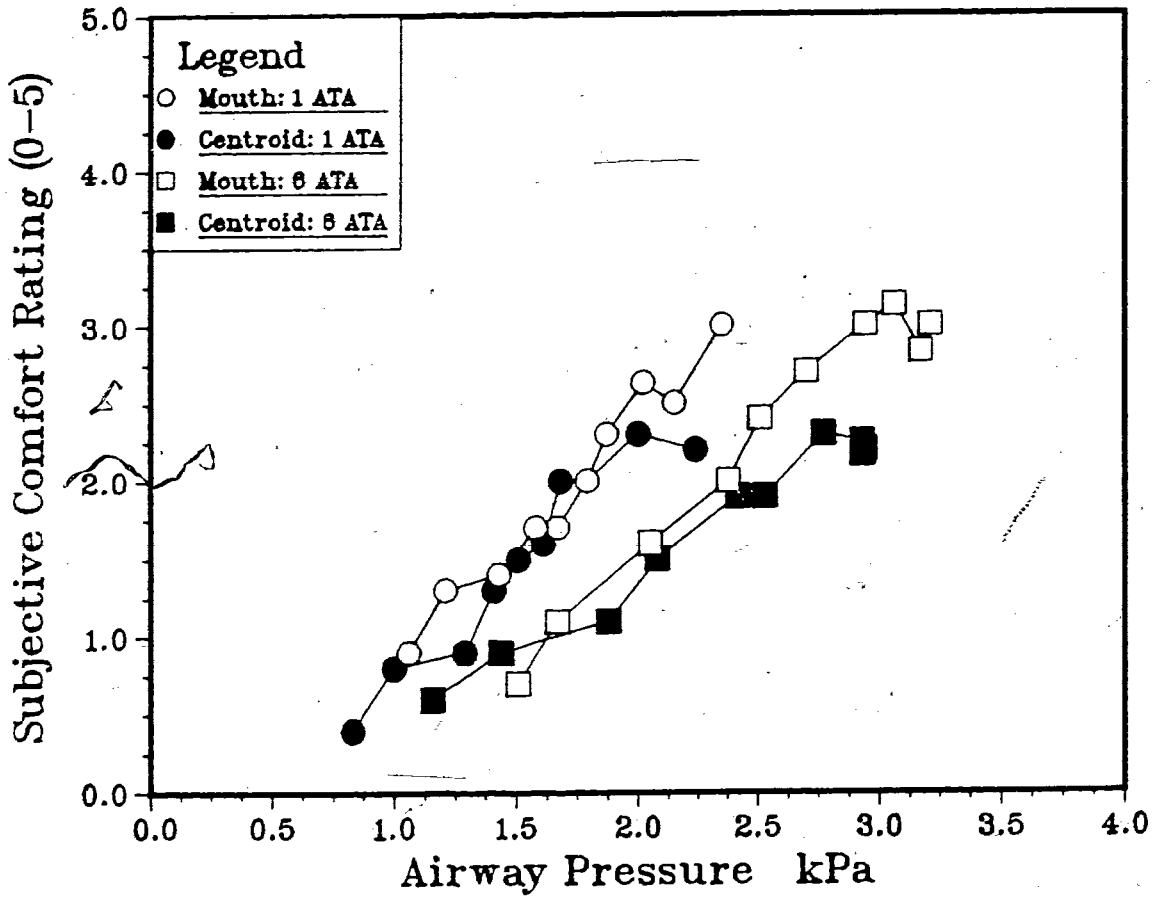


Figure 5.13: Perceived respiratory comfort with increments in airway pressure at 1 and 6 ATA, during upright, immersed exercise. [Data represent mean values only.] Standard errors are not indicated as each was previously reported in Figures 5.5 and 5.9.]

5.3 DISCUSSION

5.3.1 Physical work capacity.

Normobaric physical work capacity is usually limited by cardiovascular mechanisms (Asmussen 1964, Holmgren 1967, Saltin and Åstrand 1967, Saltin 1973, Åstrand 1976, Holloszy and Booth 1976, Saltin and Rowell 1980). However, many subjects, unless highly motivated, do not reach workloads where the cardiovascular system becomes a limitation. Subjects stop due to local fatigue, general unpleasantness, perceived maximal effort and so on. Subjects vary considerably on threshold levels for stopping, possibly as a function of previous experience. It is possible that immersed exercise at 50m may magnify the importance of these sensations, causing subjects to terminate exercise prior to reaching levels where physiological limits restrict physical power.

Dyspnea, defined as the feeling of breathlessness, is associated with respiratory perturbations occurring during or prior to exercise, which cause a mismatch between the demanded and obtainable ventilation (Derenne *et al.* 1978, Jones *et al.* 1984). Fagraeus (1981) emphasised that subjects may perceive varied dyspnea at identical workloads (relative and/or absolute), with perception being related to previous experience. It is recognised that commercial divers are a highly experienced population, however, novice divers find exercise-induced hyperventilation at depth to induce a variety of novel physiological and proprioceptive sensations. Thus one cannot ignore the potential of respiratory discomfort to limit both work capacity and physical power.

The subjects in this study reported air delivery at P_{LC} and P_{LC} minus 0.98kPa, to be significantly more comfortable than air provision at mouth or P_{LC} plus 0.98kPa, during rest and heavy steady-state exercise at 1 ATA (Figure 5.4). During incremental exercise at 1 and 6 ATA, P_{LC} air delivery was perceived to be more comfortable than air delivery at mouth pressure (Figure 5.5). When comfort was analysed at matched minute ventilations at 6 ATA (Figure 5.10), air delivery at P_{LC} was significantly more comfortable at four levels of ventilation up to 61.9 l.min^{-1}BTPS. These results would indicate that air provision at P_{LC} is preferable, since it reduces dyspneic sensations during exercise.

Thalmann *et al.* (1979) investigated underwater work in the prone position using positive and negative static breathing loads (relative to the centre of thorax). They reported a tendency towards greater dyspnea at negative static loads[19]. Direct comparison between studies is not possible because

[19] Negative static loads are analogous to regulator positions above lung *centroid* .

Thalmann *et al.* (1979) used only three subjects, without statistical analyses, and used a two point dyspneic scale. However, current results are in accord with those of Thalmann *et al.* (1979).

Thalmann *et al.* (1979) reported that only one subject could perform maximal work when breathing at a static load of −0.98kPa at 6.76 ATA. Subjects in the present study, when breathing air at mouth pressure, were exposed to a hydrostatic respiratory imbalance of about −2.4kPa[20]. At 1 and 6 ATA all but two were able to complete the exercise protocol. The combined work against the cycle and water drag, averaged 330W for the final minute of exercise in the current study. Subjects in the Thalmann study averaged 246W for five minutes, however, work against the water was not reported. It is possible that the latter subjects were exercising at greater levels, but given that the mean maximal aerobic power of the two groups was about equal[21], and that both groups were performing power tests, this conclusion may not be appropriate. It is postulated that the greater tolerance of the breathing pressure imbalance in the current investigation may be due to differences in pressure across the oropharangeal regions.

Thompson and McCally (1967) investigated differences between airway pressure and external surface pressure, and its influence upon subject comfort. Subjects consistently found an intrapulmonary pressure slightly in excess of facial surface pressure to be more comfortable than negative transpharyngeal pressures. It may be possible that in the prone position, a negative transpharyngeal pressure is less tolerable. In the upright posture, when breathing air at mouth pressure, negative pressure gradients are primarily centered in the thorax. The transpharyngeal pressure gradient is minimal, as both internal and external surfaces are at the same pressure. Thompson and McCally (1967) found their subjects were more comfortable with negative transthoracic than with negative transpharyngeal pressures.

A certain degree of scepticism surrounds the concept of perceived exertion and the reliability of such measurements (Borg 1973). It is also possible that extraneous influences may invalidate breathing comfort perception. The present subjects were given practice at rating breathing comfort during preliminary trials to minimize such influences. Furthermore, regulator pressures were varied covertly in an attempt to reduce subject bias. Both procedures were designed to increase the reliability of the comfort measurements.

At 6 ATA, the curves for respiratory comfort as a function of airway pressure, moved positively (Figure 5.13). Thus, when the mean comfort rating was two, airway pressure was about 1.5 times greater

[20] Taken from average dimensions of subjects used in Chapters 3 and 4.

[21] $\dot{V}O_2$ = 4.36 l.min^{-1} (current subjects) and 3.86 l.min^{-1}.

than the pressure recorded at 1 ATA. The corresponding reduction in minute ventilation at this point of measurement at 6 ATA was much smaller. This indicates that narcosis may have reduced the perception of discomfort. Differences between levels of narcosis at the two air delivery pressures, may have a similar tendency to reduce discomfort; whether or not this occurred is unknown. When breathing air at mouth pressure subjects consistently reported being more narcotic. If such an influence was present, the observed differences between comfort rating at 6 ATA (Figure 5.5) would underestimate differences that may have existed, if subjects were breathing an equally dense, but non-narcotic gas mixture.

From Figures 5.5 and 5.6 it is observed that breathing becomes less comfortable when ventilation is reduced by the combination of increased gas density and hydrostatic imbalance between breathing pressure and external thoracic pressure. One may conclude that for diver comfort, underwater breathing apparatus should undergo modification to enable air delivery at P_{LC} for upright postures.

Morrison and Reimers (1982) prescribed physiological specifications for underwater breathing apparatus, and provided recommended limits of comfort and tolerance for the work of breathing performed on such apparatus. The current study used a Conshelf 30 (U.S. Divers) open circuit demand regulator. Testing of this regulator at ventilations up to 70 l.min^{-1} at 6 ATA, has shown that it conformed to these specifications (Figure 5.3). By comparing the respiratory comfort data in Figure 5.11 with Figure 5.3, two observations may be made which are relevant to demand regulator selection.

First, when divers breathed air at P_{LC}, the mean respiratory comfort rating did not exceed 2.3. This corresponded with subjects reaching their limit of breathing comfort (Table 5.3). In this situation the combined internal and external respiratory work did not impose undue discomfort on the divers for ventilations up to 72 l.min^{-1}.

Second, when the divers were provided with air at mouth pressure, respiratory comfort exceeded 2.9 when ventilation was above 55 l.min^{-1}. This rating indicated that divers perceived respiration as heavy but acceptable. The difference between comfort rating at the two air supply pressures, reflects changes in the internal contribution to total respiratory work. With the Conshelf 30 regulator, the total work was within the limits of tolerance. However, with a demand regulator that requires a greater external respiratory work, the total work may become intolerable. In this situation external respiratory work may combine with increased pulmonary flow-resistance and elastic work to further impair ventilation. This may indicate, for divers working upright and using mouth-held demand regulators, that the specifications of Morrison and Reimers (1982) are not stringent enough.

Differences between the ventilatory response to exercise under normobaric and hyperbaric states have been attributed to increased inspired O_2 tensions (Bannister and Cunningham 1954, Lambertson *et al.* 1963, Lanphier 1963, Miller *et al.* 1974), reduced anaerobiosis (Taunton *et al.* 1970, Lanphier and Camporesi 1982), and increased R_{aw} (Fagraeus 1974, Van Liew 1982, 1983, Anthonisen 1984). Current comparisons between $\dot{V}_I$ at different absolute pressure and the same air delivery pressure, revealed two important observations. First, $\dot{V}_I$ decreased at depth, as reported in the literature (Lanphier 1963, Wood 1963, Bradley *et al.* 1971, Morrison *et al.* 1976, Thalmann *et al.* 1979, Hesser *et al.* 1981, Morrison and Wood 1986). Second, at 6 ATA when breathing air at mouth pressure, $\dot{V}_I$ diminished to a greater extent than at P_{LC}, as observed by Thalmann *et al.* (1979) for prone subjects.

The observation that at 6 ATA, the hydrostatic imbalance created a further $\dot{V}_I$ reduction (Figure 5.6) implies an additional ventilatory impedence. One may assume that at this depth, subjects were receiving equivalent neuro-humoral respiratory drive[22]. However, this drive did not produce equivalent ventilations.

Doell *et al.* (1979) demonstrated that equivalent hypercapnic respiratory drive results in equivalent minute ventilation, only when respiratory mechanics remain constant. They found that gas density elevation reduced ventilatory response to CO_2 by increasing airway resistance. Thus equivalent drive produced a smaller output. This observation may be applied to the current density-dependent ventilatory reduction, observed when the absolute pressure was elevated. However, at 6 ATA a second ventilatory reduction was seen, when subjects breathed air at mouth pressure (Figure 5.6). The magnitude of the latter change was approximately equivalent to the reduction attributable to raised gas density, for all but the final two minutes of exercise. It is concluded that differences in pulmonary resistance, between the two delivery pressures, caused a further mechanical lowering of the ventilation that a given respiratory drive could produce. This conclusion is supported by significant increases in pulmonary resistance observed during immersion when air was breathed at mouth pressure (Figure 4.7). It is believed that dynamic airway compression accompanying active and/or passive increments in airway resistance may be responsible for the additional ventilatory reduction.

The mechanical perturbations associated with breathing air supplied at mouth pressure at depth, may reach the levels found in patients with respiratory disorders (*e.g.* chronic obstructive lung disease). If this occurs, divers may experience a respiratory limitation of physical power, not unlike that observed

[22] Subjects were performing an identical exercise protocol, at equivalent absolute pressures, using the same breathing apparatus.

in these patients (Stubbing *et al.* 1980b).

A restriction to physical power attending air supply pressure change, was not observed in the current study. This may have resulted from the increased respiratory effort producing a reduced minute ventilation at the higher workloads, which had the effect of compensating for the external work. At 6 ATA subjects may also have experienced an elevation in aerobic capacity due to increased inspired O_2 tension. However, even without modifications to the physical power, altered air supply pressures were able to significantly change respiratory comfort, minute ventilation and the level of narcosis.

Perhaps of greater pertinence to diver safety than the ventilatory limitation *per se* is the accumulation of CO_2 in the blood. Carbon dioxide has been shown to potentiate narcosis (Case and Haldane 1941, Lanphier 1963, Jarrett 1966, Hesser *et al.* 1971). Conditions reducing $\dot{V}_I$ or elevating CO_2 production ($\dot{V}CO_2$), predispose to CO_2 accumulation.

Exercise at depth was accompanied by a progressive CO_2 retention as previously reported (Lanphier 1963, Wood and Bryan 1971, Linnarsson and Fagraeus 1976, Thalmann *et al.* 1979, Morrison and Wood 1986). Studies at 1 ATA in air have similarly shown CO_2 retention to accompany respiratory flow-resistive and elastic loading (Axen *et al.* 1983, D'Urzo *et al.* 1985).

Elevated $P_{ET}CO_2$ at depth is attributed to hypoventilation (Jarrett 1966, Wood and Bryan 1971). While $P_{ET}CO_2$ is not strictly a function of $\dot{V}_I$, but rather alveolar ventilation, $P_{ET}CO_2$ accumulation may be ascribed to ventilatory reductions accompanying the pressure related elevation in gas density. Subjects in the current study ended exercise at 6 ATA with a mean $P_{ET}CO_2$ of 62.7 and 62.6mmHg for mouth and P_{LC} air delivery respectively. These levels are in excess of those normally reported (Linnarsson and Fagraeus 1976, Morrison and Wood 1986), though Lanphier (1963) reported data exceeding 70mmHg. The present high values are attributed to the additional respiratory dead space that existed, due to the use of a movable demand regulator (see Section 5.1.2).

The consequences of progressive CO_2 retention are potentially lethal. CO_2 acts as a narcotic (Severinghaus 1974, Hesser *et al.* 1978), and at depth it potentiates nitrogen narcosis. In the present work, 70% of the subjects felt less narcotic with P_{LC} air delivery, however, lower $P_{ET}CO_2$, in response to the change in minute ventilation, was not observed. Subjects reported being more alert, more aware of physical work and time, and better able to recall events. At mouth pressure, four felt they would soon lose consciousness and five reported visual, auditory and mental aberrations. Only one subject felt more

narcotic at depth when breathing air at P_{LC}. These subjective observations do not agree with the measurements of $P_{ET}CO_2$ (Figure 5.7), which show an approximately equivalent $P_{ET}CO_2$ for each air supply pressure at 6 ATA. Three explanations may be possible. There may have been measurement errors. Subjects may have felt more comfortable with P_{LC} air provision, and this may have falsely changed their perception of narcosis. Alternatively, the $P_{ET}CO_2$ may have failed to be a valid measure of arterial and cerebral CO_2 tension.

It is concluded that the use of a mouth-held regulator at 6 ATA, without breathing pressure compensation, impairs exercise ventilation and increases respiratory discomfort. Both of these changes are attributed to increased pulmonary flow-resistance accompanying uncompensated immersion. It is therefore suggested that underwater breathing apparatus be modified to enable air provision at a positive pressure of 1.33kPa (relative to the mouth), when divers adopt an upright working posture.

CHAPTER 6

CONCLUDING STATEMENTS

6.1 CONCLUSIONS

6.1.1

Analysis of pressure–volume relaxation curves revealed a positive and approximately parallel displacement from control status, when subjects were immersed upright in thermoneutral water. As a direct result of this shift, the total respiratory relaxation volume decreased ($p<0.05$) while the total respiratory elastic work increased ($p<0.05$).

The use of facial counter–pressure to facilitate relaxation was found to produce superior pressure–volume relaxation curves, and its implementation is recommended for future investigations.

Lung *centroid locus* was positioned an average of 13.6cm below the sternal notch while upright, and 7.0cm above the sternal plane while prone, representing mean hydrostatic pressures of 1.33 and −0.69kPa, relative to these respective anatomical points. These data do not support the accepted upright lung *centroid locus* of 19cm inferior to the sternal notch (Jarrett 1965).

6.1.2

Although upright immersion caused a positive displacement of the total pressure–volume relaxation curve, it did not invoke significant changes in the static compliance of the total respiratory system, lung tissue or chest wall, when compared with control data obtained at equivalent lung volumes.

6.1.3

Upright immersion perturbed lung volume compartments and subdivisions. The application of progressively greater breathing compensation pressures, sequentially returned subdivisions towards control status.

The use of breathing pressure compensation also reduced the inspiratory muscle work performed against elastic tissues during upright immersion. At lung *centroid* pressure air delivery, inspiratory muscle work was equivalent to work obtained in air.

6.1.4

Uncompensated upright immersion significantly elevated inspiratory, expiratory and total flow-resistive pulmonary work, as well as inspiratory and expiratory pulmonary resistance measured at a constant airflow rate ($p<0.05$).

Air delivery at lung *centroid* pressure reduced all pulmonary work and resistance variables, to levels not significantly greater than obtained under control conditions. It is suggested that pulmonary work was lowered following expiratory reserve volume increments, which accompanied relaxation volume increases, vascular disgorgement, and possibly, the removal of vagally mediated airway constriction.

Expiratory pulmonary flow resistance was greater than inspiratory resistance at all breathing pressures during immersion. Inspiratory resistance and flow-resistive work appeared less sensitive to changes in air supply pressures.

6.1.5

During incremental exercise at six atmospheres absolute, a greater exercise minute ventilation was achieved when air was provided at *centroid* pressure. These higher levels of ventilation were found to be associated with greater respiratory comfort at this air supply pressure. Differences in ventilation were not apparent during surface experiments, however, respiratory comfort was significantly greater at *centroid* pressure. Since physical power was not shown to be limited when breathing air without pressure compensation, hypothesis four may not be fully accepted.

6.1.6

A greater narcosis was experienced during trials at 6 ATA with the uncompensated air delivery pressure. This was attributed to augmentation of nitrogen narcosis subsequent to hypoventilation and the retention of carbon dioxide.

6.2 PHYSIOLOGICAL MODEL SUMMARISING CONCLUSIONS

The results and conclusions from this series of investigations have been summarised and combined into a physiological model (Figure 6.1), illustrating the mechanisms through which exercise intolerance may accompany respiratory mechanical perturbations observed during upright immersion. The model contains primary observations from the present dissertation (symbolised: †) and secondary observations from other research reports (symbolised: ‡). The proposed sites through which air delivery at lung *centroid* pressure may ameliorate these modifications are also indicated (■).

The model is centered around the three components of respiratory work: elastic, flow-resistive and inertial work. Immersion and raised absolute pressure produce physical changes which may be associated with physiological perturbations. The latter either directly or indirectly, via secondary physiological changes, influence one or more of the respiratory work components, and thus elevate the total work of breathing.

Increased respiratory work may adversely affect exercise tolerance by one or more of three channels: hypoventilation (and ultimately narcosis), dyspnea and inspiratory fatigue.

6.3 RECOMMENDATIONS

The following recommendations represent possible directions of future research in this field. They include projects designed to solve questions unanswered by this dissertation, and questions raised from the results.

6.3.1

It is recommended that air delivery mechanisms of underwater breathing apparatus be modified, to enable air provision at lung *centroid* pressure to working divers, when adopting upright postures. The most immediate problems are those of the man/machine interface. Clearly, the incorporation of an airway extension, to link the diver to the demand regulator positioned at the lung *centroid* , is both cumbersome and physiologically inferior, as it involves carbon dioxide rebreathing which potentiates depth-related narcosis. The helmet mounted regulator, adapted to supply air at ambient pressure plus 1.33kPa, represents a more appealing solution. Such a mechanism would require re-designing the demand diaphragm. It is possible that a manual or automatically adjusted mechanism may be suitable.

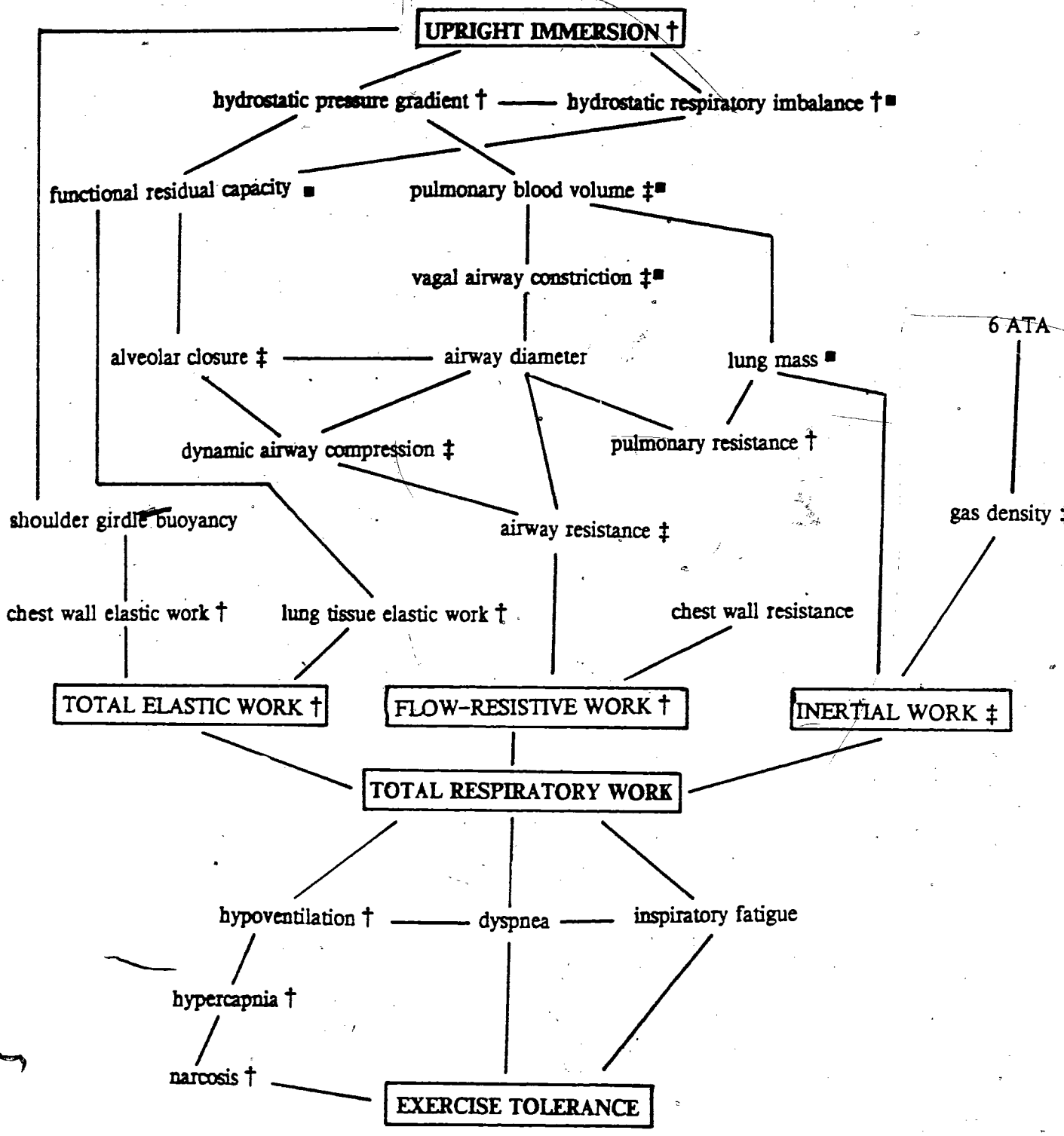


Figure 6.1: **Physiological model illustrating mechanisms through which upright immersion may induce ventilatory exercise intolerance.**

With pressure transducers mounted on the body, the latter system could alter delivery pressure to match diver orientation. For development of such a system a three dimensional analysis of lung *centroid* would be required.

6.3.2

Current works have looked only at the thermoneutral diver, unimpeded by his environment or equipment. Future investigations may be directed towards respiratory mechanical modifications attending changes in water temperature, air density and the use of diving wet and dry suits.

6.3.3

The dynamic contributions to flow-resistive respiratory work provided by the chest wall were not studied due to technical difficulties, and debate surrounding the validity of the available techniques. Its role during exercise has been approximated at about 25% of the total flow-resistive respiratory work (Goldman *et al.* 1976), however, chest wall dynamics have not been analysed during immersion. Work is needed in this area.

6.3.4

Residual volumes were not measured. Without knowledge of residual volume, total lung and functional residual capacities remain unknown. It is possible that airway closure with immersion may invalidate gas dilution techniques, as such methods measure only the accessible residual volume. To overcome this, immersion plethysmography may be required. Plethysmography would also enable the measurement of airway resistance, not obtainable using current pulmonary resistance methods.

6.3.5

To facilitate rapid validation of respiratory relaxation during static pressure-volume manoeuvres, it is suggested a computerised system be developed, for monitoring inspiratory muscle electromyograms. Pressure readings could then be taken when integrated EMG signals were below a predetermined threshold.

APPENDIX ONE:

UNITS OF MEASUREMENT

Units used in this thesis are those recommended by the Eleventh General Conference on Weight and Measures (1960), the *Système Internationale d'Unités* (*SI* units).

Units of Measurement.

MEASUREMENT	ABBREVIATION	UNITS
Area	A	square metre (m^2)
Compliance	C	litre.kilopascal^{-1} ($l.kPa^{-1}$)
Force	F	Newtons (N)
Frequency	f	hertz (Hz)
Length	l	metre (m)
Mass	m	kilogram (kg)
Power	W	Watt (W)
Pressure	P	pascal (Pa)
Resistance	R	kilopascal.litre^{-1}.second ($kPa.l^{-1}.s$)
Specific compliance	sC	kilopascal (kPa)
Time	t	second (s)
Volume	V	cubic metre (m^3)
Work	W	Joule (j)

The *SI* unit of volume is not used with reference to lung volumes, instead the litre (l) is used ($1l=1dm^3=10^{-3}m^3$). Respiratory and cardiac frequencies are expressed as breaths or beats per minute. Carbon dioxide tension is recorded in millimetres of mercury (mm Hg).

APPENDIX TWO:

MAKING INTRAOESOPHAGEAL PRESSURE PROBES.

Indirect measurement of topical pleural surface pressure via the oesophagus may be traced to Luciani (1878), but it was Buytendijk (1949) who introduced the use of oesophageal balloons. Early balloons were made in research laboratories (so called "hand-dipped" balloons) by dipping suitably shaped glass rods into a latex bath, then curing the latex in an oven (Mead *et al.* 1955). Once dry the balloons were removed and attached to catheters. Today one may purchase balloons and even balloon catheters systems. Intraoesophageal probes used in this thesis consisted of manufactured balloons (A & E Medical Corporation, New Jersey, U.S.A.) fitted to catheters in the laboratory.

Adult oesophageal balloons (latex rubber, 15cm long, internal diameter 1.825cm, wall thickness 8.47×10^{-3}mm) were cut to 10cm in length. Ideally balloons should be tapered at both ends, this was not possible with the current balloon length. Catheters (teflon, internal diameter 1.35mm, wall thickness 0.3mm) were cut to one of two lengths. Trials in air used 1 metre catheters, immersion trials used 50cm catheters. A 17 gauge hypodermic needle was machined smooth and firmly sealed into one end of each catheter (epoxy®). The other end was pierced laterally with a small hypodermic needle to provide a spiral sequence of holes between the tip and a point 9cm above the tip. Rough edges were carefully removed (Figure A4.1).

Balloons were secured and sealed to catheters using epoxy® resin. The tip of the pierced end was plugged and a fine layer of epoxy spread around the outer surface. This provided a means of securing the balloon tip permanently to the catheter. Balloons were gently placed over the catheter. Inner surfaces of the balloon mouth were also sealed with epoxy as was the outer catheter surface below this point. Balloons were then secured to the catheter with redundant folds also being coated with an epoxy film. Both ends were immediately secured with dark cotton thread. A final coat of epoxy covered the cotton and balloon edges to ensure sealage and smooth surfaces to avoid discomfort during insertion. After curing, the balloons were pressure tested underwater, dusted with talcum powder and stored in a refrigerator.

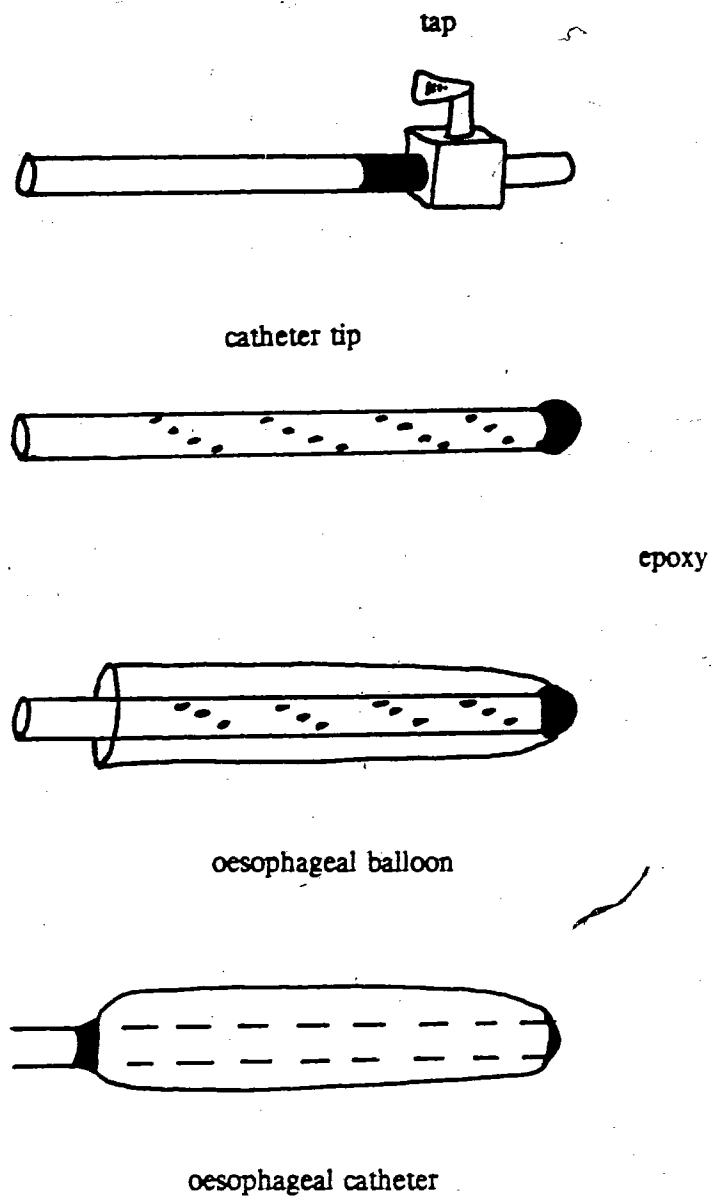


Figure A4.1: Stages in making intraoesophageal pressure probes.

APPENDIX THREE:

SUMMARY TABLES FOR MANOVA AND POST HOC MULTIPLE COMPARISON TESTS.

Title: **Chapter 3: MANOVA summary table.** Elastic work.

Source	mean square	F	df	p
total	8.878	407.88	4/36	0.001*
lung	1.753	151.75	4/36	0.001*
chest	2.960	100.92	4/36	0.001*
muscle	1.206	47.23	4/36	0.001*

*= significant at the 0.05 level.

Title: **Chapter 3: MANOVA summary table.** Lung volumes.

Source	mean square	F	df	p
VC	0.726	8.55	4/32	0.001*
IC	3.355	19.95	4/32	0.001*
V_T	0.144	1.08	4/32	0.383
ERV	6.598	29.55	4/32	0.001*
IRV	3.263	12.61	4/32	0.001*

*= significant at the 0.05 level.

Title: **CHAPTER 3: Total respiratory elastic work.**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	9.24*	7.77*	0.45	8.78*
Mouth		——	8.62*	15.90*	25.14*
P_{LC}-.98			——	7.29*	16.52*
P_{LC}				——	9.23*
P_{LC}+.98					——

Tukey's critical range = 1.90.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 3 : Lung tissue elastic work.**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	4.25*	2.52*	.70	6.73*
Mouth		——	1.73*	4.95*	10.98*
P_{LC}-.98			——	3.22*	9.25*
P_{LC}				——	6.03*
P_{LC}+.98					——

Tukey's critical range = 1.38.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 3: Chest wall elastic work.**

	Air	Mouth	$P_{LC^-.98}$	P_{LC}	$P_{LC^+.98}$
Air	——	11.88*	5.19*	1.08	1.74
Mouth		——	6.69*	10.80*	13.62*
$P_{LC^-.98}$			——	4.11*	6.93*
P_{LC}				——	2.82*
$P_{LC^+.98}$					——

Tukey's critical range = 2.20.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 3: Inspiratory muscle work.**

	Air	Mouth	$P_{LC^-.98}$	P_{LC}	$P_{LC^+.98}$
Air	——	7.47*	2.10*	.43	1.02
Mouth		——	5.37*	7.92*	8.49*
$P_{LC^-.98}$			——	2.53*	3.12*
P_{LC}				——	.59*
$P_{LC^+.98}$					——

Tukey's critical range = 2.06.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 3: Vital capacity.**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	2.55	0.63	1.76	3.41
Mouth		——	3.18	4.31*	5.96*
P_{LC}-.98			——	1.13	1.65
P_{LC}				——	2.78
P_{LC}+.98					——

Tukey's critical range = 3.56.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 3: Inspiratory capacity.**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	8.60*	6.28*	2.11	5.44*
Mouth		——	2.38	6.50*	14.04*
P_{LC}-.98			——	4.18	7.54*
P_{LC}				——	11.72*
P_{LC}+.98					——

Tukey's critical range = 5.03.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 3: Expiratory reserve volume.**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	10.78*	5.57	0.22	9.90*
Mouth		——	5.21	11.00*	20.68*
P_{LC}-.98			——	5.79*	9.68*
P_{LC}				——	15.47*
P_{LC}+.98					——

Tukey's critical range = 5.79.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 3: Inspiratory reserve volume.**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	5.73	5.58	0.81	7.55*
Mouth		——	0.15	4.92	13.28*
P_{LC}-.98			——	4.77	8.36*
P_{LC}				——	13.13*
P_{LC}+.98					——

Tukey's critical range = 6.24.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **Chapter 4: MANOVA summary table.** Flow-resistive work.

Source	mean square	F	df	p
total	0.383	7.87	4/32	0.001*
inspired	0.052	10.86	4/32	0.001*
expired	0.169	4.52	4/32	0.005*

*= significant at the 0.05 level.

Title: **Chapter 4: MANOVA summary table.** Flow-resistance ($0.5l.s^{-1}$).

Source	mean square	F	df	p
total	0.047	6.48	3/24	0.002*
inspired	0.029	4.41	3/24	0.013*
expired	0.078	4.14	3/24	0.017*

*= significant at the 0.05 level.

Title: **Chapter 4: MANOVA summary table.** Respiratory timing.

Source	mean square	F	df	p
V_T/T_I	0.196	2.56	4/32	0.058
T_I/T_{TOT}	0.004	2.86	4/20	0.051

*= significant at the 0.05 level.

Title: **CHAPTER 4: Total flow-resistive work.**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	4.92*	2.81*	1.98	1.13
Mouth		——	2.11	2.94*	3.79*
P_{LC}-.98			——	.83	1.68
P_{LC}				——	.98
P_{LC}+.98					——

Tukey's critical range = 2.71.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 4: Inspiratory flow-resistive work.**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	1.93*	.84	.89*	.83
Mouth		——	1.09*	1.04*	1.10*
P_{LC}-.98			——	.05	.01
P_{LC}				——	.06
P_{LC}+.98					——

Tukey's critical range = 0.85.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 4: Expiratory flow–resistive work.**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	3.02*	2.0	1.14	.34
Mouth		——	1.02	1.88	2.68*
P_{LC}-.98			——	.86	1.66
P_{LC}				——	.8
P_{LC}+.98					——

Tukey's critical range = 2.82.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 4: Total pulmonary resistance (0.5 $l.s^{-1}$ flow rate)**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	2.29*	1.34*	1.15*	.30
Mouth		——	.95	1.0	1.99*
P_{LC}-.98			——	.19	1.04*
P_{LC}				——	.85
P_{LC}+.98					——

Tukey's critical range = 1.03.
* = significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 4: Inspiratory pulmonary resistance (0.5 l.s^{-1}).**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	1.42*	1.12*	.98*	.47
Mouth		——	.30	.44	.95*
P_{LC}-.98			——	.14	.65
P_{LC}				——	.51
P_{LC}+.98					——

Tukey's critical range = 0.84.
*= significant difference.
Statistical power of MANOVA >0.99.

Title: **CHAPTER 4: Expiratory pulmonary resistance (0.5 l.s^{-1}).**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	3.16*	1.90*	1.15	.42
Mouth		——	1.26	2.01*	2.74*
P_{LC}-.98			——	.75	1.48*
P_{LC}				——	.73
P_{LC}+.98					——

Tukey's critical range = 1.45.
*= significant difference.
Statistical power of MANOVA = 0.97.

Title: **CHAPTER 4: Tidal volume / inspiratory duration.**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	3.35	1.87	1.24	2.86
Mouth		——	1.48	2.11	.49
P_{LC}-.98			——	.63	.99
P_{LC}				——	1.62
P_{LC}+.98					——

Tukey's critical range = 3.38.
** = significant difference.
Statistical power of MANOVA = 0.47

Title: **CHAPTER 4: Inspiratory duration / total cycle duration.**

	Air	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Air	——	.17	.15	.12	.17
Mouth		——	.02	.29	.34
P_{LC}-.98			——	.27	.32
P_{LC}				——	.05
P_{LC}+.98					——

Tukey's critical range = 0.47.
**= significant difference.
Statistical power of MANOVA = 0.70.

Title: **Chapter 5: MANOVA summary table.** Physical power.

Source	mean square	F	df	p
Comfort	0.306	0.68	27/237	0.889
Heart rate	63.761	1.07	27/237	0.376
$\dot{V}_I$	239.445	8.31	27/237	0.000*
$P_{ET}CO_2$	110.173	6.54	27/237	0.000*
Airway pressure	2076.420	3.80	27/237	0.000*

*= significant at the 0.05 level.

The following tables are summaries of the Hotelling's T analysis for correlated samples. The numbers within each table represent T values for the corresponding comparisons. The critical value is the level of T required for significance at the 0.05 level.

Title: **Chapter 5:** Summary of Hotellings analysis for comparisons of paired data. Two comparisons are presented:
(1) Comparison of means at 1 and 6 ATA, with data summed across time and air supply pressure.
(2) Comparison of means at mouth and P_{LC} air supply pressures, with data summed across time and absolute pressure.

Variable	1 ATA versus 6 ATA	Mouth pressure versus P_{LC}
Comfort	3.73*	6.93*
Heart rate	7.45*	3.48*
$P_{ET}CO_2$	13.22*	2.22*
$\dot{V}_I$	9.14*	5.15*
Airway pressure	15.20*	7.18*

Critical value = 2.00.
*= significant difference at the 0.05 level.

Title: **CHAPTER 5: Respiratory comfort at rest.**

	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Mouth	——	2.39*	2.39*	.89
P_{LC}-.98		——	0.0	1.43
P_{LC}			——	1.43
P_{LC}+.98				——

Critical value = 2.365.
* = significant difference.

Title: **CHAPTER 5: Respiratory comfort exercising at 279.6 W.**

	Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
Mouth	——	3.74*	3.86*	1.53
P_{LC}-.98		——	.55	1.82
P_{LC}			——	1.67
P_{LC}+.98				——

Critical value = 2.365.
* = significant difference.

Title: **CHAPTER 5: Respiratory comfort: exercise versus rest.**

Mouth	P_{LC}-.98	P_{LC}	P_{LC}+.98
3.75*	2.40*	3.86*	2.55*

Critical value = 2.365.
* = significant difference.

Title: **Chapter 5: Respiratory comfort at 1 ATA.**

Time	1	2	3	4	5	6	7	8	9	10
Mouth P_{LC} diff.	3.00*	1.63	1.63	1.31	0.69	1.31	0.90	2.25*	2.33*	2.00

Critical value = 2.20.
* = significant difference.

Title: **Chapter 5: Respiratory comfort at 6 ATA.**

Time	1	2	3	4	5	6	7	8	9	10
Mouth P_{LC} diff.	0.56	0.69	2.24*	1.34	1.63	3.21*	2.33*	2.97*	2.00	1.37

Critical value = 2.20.
* = significant difference.

Title: **Chapter 5: Minute ventilation: 1 ATA.**

Time	1	2	3	4	5	6	7	8	9	10
Mouth P_{LC} diff.	2.23*	0.86	0.42	1.22	1.59	0.67	0.40	0.84	0.34	0.85

Critical value = 2.20.
*= significant difference.

Title: **Chapter 5: Minute ventilation: 6 ATA.**

Time	1	2	3	4	5	6	7	8	9	10
Mouth P_{LC} diff.	0.22	0.80	2.14	2.31*	3.27*	2.57*	4.86*	4.05*	3.67*	2.99*

Critical value = 2.20.
* = significant difference.

Title: **Chapter 5: End-tidal carbon dioxide tension: 1 ATA.**

Time	1	2	3	4	5	6	7	8	9	10
Mouth P_{LC} diff.	.06	1.1	2.03	1.73	3.95*	3.61*	2.78*	2.63*	2.94*	2.48*

Critical value = 2.262.
* = significant difference.

Title: **Chapter 5: End-tidal carbon dioxide tension: 6 ATA.**

Time	1	2	3	4	5	6	7	8	9	10
Mouth P_{LC} diff.	.81	.31	.45	.22	.91	0	.06	.05	.14	.02

Critical value = 2.262.
* = significant difference.

Title: **Chapter 5: Heart rate: 1 ATA.**

Time	1	2	3	4	5	6	7	8	9	10
Mouth P_{LC} diff.	1.11	1.75	1.51	1.52	2.02	1.14	.66	.29	.26	.71

Critical value = 2.262.
* = significant difference.

Title: **Chapter 5: Heart rate: 6 ATA.**

Time	1	2	3	4	5	6	7	8	9	10
Mouth P_{LC} diff.	.71	.49	1.65	.57	1.73	1.05	.71	.96	.68	.44

Critical value = 2.262.
* = significant difference.

Title: **Chapter 5: Airway pressure: 1 ATA.**

Time	1	2	3	4	5	6	7	8	9	10
Mouth P_{LC} diff.	5.48*	5.0*	2.60*	2.90*	2.54*	2.67*	4.15*	2.87*	.99	.3

Critical value = 2.262.
* = significant difference.

Title: **Chapter 5: Airway pressure: 6 ATA.**

Time	1	2	3	4	5	6	7	8	9	10
Mouth P_{LC} diff.	2.32*	1.65	1.51	1.86	1.18	1.10	1.26	1.10	3.35*	1.71

Critical value = 2.262.
* = significant difference.

Title: **Chapter 5: MANOVA summary table.** Physical power.
Variables analysed at equivalent minute ventilations at 6 ATA.

Source	mean square	F	df	p
Comfort	1.027	2.22	6/58	0.054
Heart rate	95.452	2.13	6/58	0.063
$P_{ET}CO_2$	9.954	0.35	6/58	0.910
Airway pressure	745.486	0.64*	6/58	0.697

* = significant at the 0.05 level.

Title: **Chapter 5: Respiratory comfort for equivalent ventilations at 6 ATA.**

Level	1	2	3	4	5	6	7
Mouth							
P_{LC}							
diff.	1	2	10*	9*	8	11*	8*

Tukey's critical range = 8.95 and 7.49 beyond level 6.
* = significant difference.
Statistical power of MANOVA = 0.60.

Title: **Chapter 5: Heart rate for equivalent ventilations at 6 ATA.**

Time	1	2	3	4	5	6	7
Mouth							
P_{LC}							
diff.	17	17	91*	69*	100*	97*	90*

Tukey's critical range = 55.7.
* = significant difference.
Statistical power of MANOVA = 0.40.

REFERENCES

1. Agostoni, E. 1964. Action of respiratory muscles. In: Fenn, W.O., & Rahn, H. Handbook of Physiology. Respiration I. American Physiological Society. Washington.

2. Agostoni, E. 1970. Dynamic properties of the relaxed respiratory system. In: Campbell, E.J.M., Agostoni, E., & Newsom Davis, J. The Respiratory Muscles. Mechanics and Neural Control. Lloyd-Luke (Medical Books) Ltd. London. Pp.80-114.

3. Agostoni, E. 1972. Mechanics of the pleural space. Physiol.Rev. 52(1):57-128.

4. Agostoni, E., Campbell, E.J.M., & Freedman, S. 1970. The mechanical work of breathing. In: Campbell, E.J.M., Agostoni, E., & Newsom Davis, J. The Respiratory Muscles. Mechanics and Neural Control. Lloyd-Luke (Medical Books) Ltd. London. Pp.115-137.

5. Agostoni, E., & D'Angelo, E. 1969. Thickness and pressure of the pleural liquid at various heights and with various hydrothoraces. Respir.Physiol. 6:330-342.

6. Agostoni, E., & Fenn, W.O. 1960. Velocity of muscle shortening as a limiting factor in respiratory air flow. J.Appl.Physiol. 15:349-353.

7. Agostoni, E., Gurtner, G., Torri, G., & Rahn, H. 1966. Respiratory mechanics during submersion and negative-pressure breathing. J.Appl.Physiol. 21(1):251-258.

8. Agostoni, E., & Mead, J. 1964. Statics of the respiratory system. In: Fenn, W.O., & Rahn, H. Handbook of Physiology: Respiration I. American Physiological Society, Washingon, D.C. Pp.387-409.

9. Agostoni, E., & Rahn, H. 1960a. Abdominal and thoracic pressures at different lung volumes. J.Appl.Physiol. 15(6):1087-1092.

10. Agostoni, E., & Rahn, H. 1960b. La componente abdominale nella curva pressione-volume del torace. Boll.Soc.Ital.Biol.Sper. 36:1797-1798.

11. Ahlström, H., & Jonson, B. 1974a. Pulmonary mechanics during the first year of life. Scand.J.Resp.Dis. 55:141-154.

12. Ahlström, H., & Jonson, B. 1974b. Pulmonary mechanics in infants. Acta Paediatr.Scand. Suppl. 244:1-21.

13. American College of Chest Physicians - American Thoracic Society. 1975. Pulmonary terms and symbols. Chest 5:583-593.

14. Anthonisen, N.R. 1984. Respiration. In: Shilling, C.W., Carlston, C.B., & Mathias, R.A. The Physician's Guide to Diving Medicine. Plenum Press. New York. Pp.71-85.

15. Anthonisen, N.R., Utz, G., Kryger, M.H., & Urbanetti, J.S. 1976. Exercise tolerance at 4 and 6 ATA. Undersea Biomed.Res. 3(2):95-102.

16. Arborelius, M.Jr., Balldin, U.I., Lilja, B., & Lundgren, C.E.G. 1972a. Regional lung function in man during immersion with the head above water. Aerospace Med. 43(7):701-707.

17. Arborelius, M.Jr., Balldin, U.I., Lilja, B., & Lundgren, C.E.G. 1972b. Hemodynamic changes in man during immersion with the head above water. Aerospace Med. 43(6):592-598.

18. Arborelius, M.Jr., Dahlbäck, G.O., & Data, P-G. Cardiac output and gas exchange during heavy exercise with a positive pressure respiratory protective apparatus. Scand.J.Work Environ.Health. 9:471-477.

19. Askanazi, J., Silverberg, P.A., Foster, R.J., Hyman, A.I., Milic-Emili, J., & Kinney J.M. 1980. Effects of respiratory apparatus on breathing pattern. J.Appl.Physiol. 48:577-580.

20. Asmussen, E., & Nielsen, M. 1960. Alveolo-arterial gas exchange at rest and during work at different O_2 tensions. Acta Physiol.Scand. 50:153-166.

21. Åstrand, P.-O. 1952. Experimental Studies of Physical Working Capacity in Relation to Sex and Age. Munksgaard. Copenhagen.

22. Åstrand, P.-O. 1976. Quantification of exercise capability and evaluation of physical capacity in man. Prog.Cardiovasc.Dis. 19:51-67.

23. Åstrand, P.-O., & Rhyming, I. 1954. A nomogram for calculation of aerobic capacity from pulse rate during submaximal work. J.Appl.Physiol. 7:218-223.

24. Attinger, E.O., Herschfus, J.A., & Segal, M.S. 1956. The mechanics of breathing in different body positions. II. In cardiopulmonary disease. J.Clin.Invest. 35:912-920.

25. Attinger, E.O., Monroe, R.G., & Segal, M.S. 1956. The mechanics of breathing in different body positions. I. In normal subjects. J.Clin.Invest. 35:904-911.

26. Attinger, E.O., & Segal, M.S. 1959. Mechanics of breathing I. The physical properties of the lung. Am.Rev.Respir.Dis. 80:38-45.

27. Attinger, E.O., & Segal, M.S. 1959. Mechanics of breathing. II. The respiratory work. Am.Rev.Respir.Dis. 80:46-52.

28. Axen, K., Haas, S.S., Haas, F., Gaudino, D., & Haas, A. 1983. Ventilatory adjustments during sustained mechanical loading in conscious humans. J.Appl.Physiol. 55:1211-1218.

29. Bachofen, H., & Scherrer, M. 1970. Lung tissue resistance in healthy subjects and in patients with lung disease. In: Bouhuys, A. Airway Dynamics. Charles C.Thomas. Springfield, Ill. P.123.

30. Baer, R., Dahlbäck, G.O., & Balldin, U.I. 1986. Pulmonary mechanics and atelectasis during immersion in oxygen breathing subjects. Undersea Biomed.Res. (In press).

31. Ballantine, T.V.N., Proctor, H.J., Broussard, N.D., & Litt, B.D. 1970. The work of breathing: potential for clinical application and the results of studies performed on 100 normal males. Ann.Surg. 171(4):590-594.

32. Balldin, U.I., Lundgren, C.E.G., Lundvall, J., & Melander, S. 1971. Changes in the elimination of 133 Xenon from the anterior tibial muscle in man induced by immersion in water and by shifts in body position. Aerospace Med. 42(5):489-493.

33. Banister, E.W. 1979. The perception of effort: an inductive approach. Eur.J.Applied Physiol. 41:141-150.

34. Bannister, R.G., & Cunningham, D.J.C. 1954. The effects on the respiration and performance during exercise of adding oxygen to the inspired air. J.Physiol. 125:118-137.

35. Banzett, R.B., Lansing, R.W., & Reid, M.B. 1985. Reflex compensation of voluntary inspiration when immersion changes diaphragm length. J.Appl.Physiol. 59(2):611-618.

36. Bartels, H., Dejours, P., Kellogg, R.H., & Mead, J. 1973. Glossary on respiration and gas exchange. J.Appl.Physiol. 34(4):549-558.

37. Basch, S.Von. 1887. Ueber eine funktion des capillardruckes in den lungenalveolen. Weiner Med.Blatter. 15:465-467.

38. Bayliss, L.E., & Robertson, G.W. 1939. The visco-elastic properties of the lung. Q.J.Exp.Physiol. 29:27.

39. Becker, L.J., Crawford, G., Long, G.R., Mayers, I, Schumacker, P.T., Ruygrok, P, Wood, L.D.H., & Breen, P.H. 1983. Gas trapping can underestimate functional residual capacity (FRC) by Boyle's law. Physiologist. 26:A-114.

40. Beckman, E.L., Coburn, K.R., Chambers, R.M., DeForest, R.E., Augerson, W.S., & Benson, V.G. 1961. Physiologic changes observed in human subjects during zero G simulation by immersion in water up to neck level. Aerospace Med. 32:1031–1041.

41. Beers, Y. 1957. Introduction to the Theory of Error. Addison-Wesley Publishing Company. Reading, Massachusetts.

42. Begin, R., Epstein, M., Sackner, M.A., Levinson, R., Dougherty, R., & Duncan, D. 1976. Effects of water immersion to the neck on pulmonary circulation and tissue volume in man. J.Appl.Physiol. 40(3):293–299.

43. Bentley, R.A., Griffin, O.G., Love, R.G., Muir, D.C.F., & Sweetland, K.F. 1973. Acceptable levels for breathing resistance of respiratory apparatus. Arch.Environ.Health 27:273–280.

44. Bernoulli, G. 1911. Zur Mechanik der Atembewegungen. Arch.f.exper.Path.u.Pharmakol. 66:313.

45. Berson, A.S. 1970. Analog-to-digital conversion for medicine. In: Caceres, C.A., & Dreifus, C.S. Clinical Electrocardiography. Academic Press. New York. Pp.21–35.

46. Bishop, B. 1963. Abdominal muscle and diaphragm activities and cavity pressures in pressure breathing. J.Appl.Physiol. 18(1):37–42.

47. Bjurstedt, H., Rosenhamer, G., Hesser, C.M., & Lindborg, B. 1980. Responses to continuous negative-pressure breathing in man at rest and during exercise. J.Appl.Physiol. 48(6):977–981.

48. Bondi, K.R., Young, J.M., Bennett, R.M., & Bradley, M.E. 1976. Closing volumes in man immersed to the neck in water. J.Appl.Physiol. 40(5):736–740.

49. Bondurant, S., Carlson, M., Hawley, R., & Klatte, E. 1960a. Respiratory compression of the esophagus by mediastinal structures. Physiologist. 3:26.

50. Bondurant, S., Hickam, J.B., & Isley, J.K. 1957. Pulmonary and circulatory effects of acute pulmonary vascular engorgement in normal subjects. J.Clin.Invest. 36:59–66.

51. Bondurant, S., Mead, J., & Cook, C.D. 1960b. A re-evaluation of effects of acute central congestion on pulmonary compliance in normal subjects. J.Appl.Physiol. 15(5):875–877.

52. Borelli, G.A. 1680. De Notu Animalium. Volume 1. Bernabo. Rome.

53. Boren, H.G., Kory, R.C., & Syner, J.C. The veterans administration-army cooperative study of pulmonary function. Am.J.Med. 41:96–114.

54. Borg, G.A.V. 1973. Perceived exertion: a note on "history" and methods. Med.Sci.Sports 5(2):90–93.

55. Borst, H.G., Berglund, E., Whittenberger, J.L., Mead, J., McGregor, M., & Collier, C. 1957. The effect of pulmonary vascular pressures on the mechanical properties of the lungs of anesthetized dogs. J.Clin.Invest. 36:1708–1714.

56. Bouhuys, M.D. 1977. Breathing: Physiology, Environment and Lung Disease. Grune & Stratton. New York.

57. Bradley, M.E., Anthonisen, N.R., Vorosmarti, J., & Linaweaver, P.G. 1971. Respiratory and cardiac responses to exercise in subjects breathing helium-oxygen mixtures at pressures from sea level to 19.2 atmospheres. In: Lambertson, C.J. Underwater Physiology. Proceedings of the Fourth Symposium on Underwater Physiology. Academic Press. New York. Pp.325–337.

58. Bradley, M.E., & Leith, D.E. 1978. Ventilatory muscle training and the oxygen cost of sustained hyperpnea. J.Appl.Physiol. 45:885–892.

59. Bradley, M.E., Robertson, C.H.Jr., & Goehring, G.S. 1979. The accurate measurement and fractionation of ventilation during immersion. Undersea Biomed.Res. 6(Suppl.1):22.

60. Briscoe, W.A. 1965. Lung volumes. In: Fenn, W.O., & Rahn, H. Handbook of Physiology, Respiration. Volume 2. American Physiological Society. Washington. Pp.1345–1379.

61. Briscoe, W.A., & Du Bois, A.B. 1958. The relationship between airway resistance, airway conductance and lung volume in subjects of different age and body size. J.Clin.Invest. 37:1279–1285.

62. Brŏzek, J., Henschel, A., & Keys, A. 1949. Effect of submersion in water on the volume of residual air in man. J.Appl.Physiol. 2:240–246.

63. Bryan, A.C., Bentivoglio, L.G., Beerd, F., MacLeish, H., Zidula, A., & Bates, D.V. 1964. Factors affecting regional distribution of ventilation and perfusion in the lung. J.Appl.Physiol. 19:395–402.

64. Buhlmann, A.A. 1963. Respiratory resistance with hyperbaric gas mixtures. In: Lambertsen, C.J., & Greenbaum, L.J. Underwater Physiology. Proceedings of the Second Symposium on Underwater Physiology. National Academy Science – National Research Council Publication. Washington, D.C. Pp.98–107.

65. Buono, M.J. 1983. Effect of central vascular engorgement and immersion on various lung volumes. J.Appl.Physiol. 54(4):1094–1096.

66. Burki, N.K. 1976. Effects of immersion in water and changes in intrathoracic blood volume on lung function in man. Clin.Sci.Mol.Med. 51:303–311.

67. Butler, J., & Arnott, W.M. 1955. The work of pulmonary ventilation at different respiratory levels. Clin.Sci. 4:703–710.

68. Butler, J., Caro, C.G., Alcala, R., & DuBois, A.B. 1960. Physiological factors affecting airway resistance in normal subjects and in patients with obstructive respiratory disease. J.Clin.Invest. 39:584–591.

69. Butler, J., & Smith, B.H. 1957. Pressure–volume relationships of the chest in the completely relaxed anaesthetised patient. Clin.Sci. 16:125–147.

70. Butler, J., White, H.C., & Melville Arnott, W. 1957. The pulmonary compliance in normal subjects. Clin.Sci. 16:709–729.

71. Buytendijk, H.J. 1949. Oesophagusdruk en Longelasticiteit. Thesis. Groningen, Netherlands.

72. Campbell, E.J.M., Westlake, E.K., & Cherniack, R.M. 1959. The oxygen consumption and efficiency of the respiratory muscles of young male subjects. Clin.Sci. 18:55–65.

73. Caro, C.G., Butler, J., & DuBois, A.B. 1960. Some effects of restriction of chest cage expansion on pulmonary function in man: an experimental study. J.Clin.Invest. 39:573–583.

74. Carson, J. 1820. On the elasticity of the lungs. Phil.Trans. 110:29–45.

75. Case, E.M., & Haldane, J.B.S. 1941. Human physiology under high pressure. 1.Effects of nitrogen, carbon dioxide, and cold. J.Hyg. 41:225–249.

76. Castile, R.G., Hyatt, R.E., & Rodarte, J.R. 1980. Determinants of maximal expiratory flow and density dependence in normal humans. J.Appl.Physiol. 49(5):897–904.

77. Cerretelli, P., Sikand, R.S., & Farhi, L.E. 1969. Effect of increased airway resistance on ventilation and gas exchange during exercise. J.Appl.Physiol. 27(5):597–600.

78. Cherniack, R.M. 1959. The oxygen consumption and efficiency of the respiratory muscles in health and emphysema. J.Clin.Invest. 38:494–499.

79. Cherniack, N.S., & Altose, M.D. 1981. Respiratory responses to ventilatory loading. In: Hornbein, T.F. Regulation of breathing. Part II. Marcel Dekker. New York.

80. Cherniack, R.M., Farhi, L.E., Armstrong, B.W., & Proctor, D.F. 1955. A comparison of oesophageal and intrapleural pressure in man. J.Appl.Physiol. 8:203–211.

81. Chiang, S.T. 1971. Distribution of ventilation and frequency-dependence of dynamic lung compliance. Thorax. 26:721–726.

82. Chiang, S.T., Steigbigel, N.H., & Lyons, H.A. 1965. Pulmonary compliance and nonelastic resistance during treadmill exercise. J.Appl.Physiol. 20(6):1194–1198.

83. Choukroun, M.L., Guenard, H., & Varene, P. 1983. Pulmonary capillary blood volume during immersion in water at different temperatures. Undersea Biomed.Res. 10(4):331–342.

84. Christie, R.V. 1953. Dyspnoea in Relation to the Visco-Elastic Properties of the Lung. Proc.R.Soc.Lond. 46:381–386.

85. Christie, R.V., & McIntosh, C.A. 1934. The measurement of the intrapleural pressure in man and its significance. J.Clin.Invest. 13:279–294.

86. Cimsit, M., & Flook, V. 1981. Pulmonary function in divers. In: Bachrach, A.J., & Matzen, M.M. Underwater Physiology VII. Proceedings of the Seventh Symposium on Underwater Physiology. Undersea Medical Society, Inc., Bethesda, Maryland. Pp.249–255.

87. Clark, T.J.H., Freedman, S., Campbell, E.J.M., & Winn, R.R. 1969. The ventilatory capacity of patients with chronic airway obstruction. Clin.Sci. 36:307–316.

88. Clarke,J.R., Jaeger, M.J., Zumrick,J.L., O'Bryan, R., & Spaur,W.H. 1982. Respiratory resistance from 1 to 46 ATA measured with the interrupter technique. J.Appl.Physiol. 52(3):549–555.

89. Clarysse, I., & Demedts, M. 1985. Human esophageal pressures and chest wall configuration in upright and head-down posture. J.Appl.Physiol. 59(2):401–407.

90. Cloetta, M. 1913. Untersuchungen über die Elastizität der Lunge und deren Bedeutung für die Zirkulation. Pflugers Arch. 152:339.

91. Cohen, R., Bell, W.H., Saltzman, H.A., & Kylstra, J.A. 1971. Alveolar-arterial oxygen pressure difference in man immersed up to the neck in water. J.Appl.Physiol. 30(5):720–723.

92. Cook, C.D., Mead, J., Schreiner, G.L., Frank, N.R., & Craig, J.M. 1959. Pulmonary mechanics during induced pulmonary edema in anesthetized dogs. J.Appl.Physiol. 14(2):177–186.

93. Cooper, E.A. 1960. A comparison of the respiratory work done against an external resistance by man and by a sine-wave pump. Q.J.Exp.Physiol. 45:179–191.

94. Cooper, E.A. 1960. Suggested methods of testing and standards of resistance for respiratory protective devices. J.Appl.Physiol. 15:1053–1061.

95. Cotes, J.E. 1979. Lung Function. Assessment and Application in Medicine. Blackwell Scientific Publications, London.

96. Craig, A.B.Jr. 1960. Effects of position on expiratory reserve volume of the lungs. J.Appl.Physiol. 15(1):59–61.

97. Craig, A.B.Jr., & Dvorak, M. 1975. Expiratory reserve volume and vital capacity of the lungs during immersion in water. J.Appl.Physiol. 38(1):5–9.

98. Craig, A.B.Jr., & Ware, D.E. 1967. Effect of immersion in water on vital capacity and residual volume of the lungs. J.Appl.Physiol. 23(4):423–425.

99. Dahlbäck, G.O. 1975. Influence of intrathoracic blood pooling on pulmonary air-trapping during immersion. Undersea Biomed.Res. 2(2):133-140.

100. Dahlbäck, G.O. 1978. Lung mechanics during immersion in water with special reference to pulmonary air trapping. Ph.D. Thesis. Laboratory of Aviation and Naval Physiology, Institute of Physiology and Biophysics, University of Lund, Sweden.

101. Dahlbäck, G.O. 1985. Hydrogen in breathing gas reduces work of breathing. Presented at The Undersea Medical Society Workshop on the Use of Hydrogen as a Component of Breathing Gas Mixtures for Deep Diving held in Wilmington, North Carolina, October 3-5. Pp.1-21.

102. Dahlbäck, G.O., & Balldin, U.I. 1983. Positive-pressure oxygen breathing and pulmonary atelectasis during immersion. Undersea Biomed.Res.10:39-44.

103. Dahlbäck, G.O., & Balldin U.I. 1984. Physiological effects of pressure demand masks during heavy exercise. Am.Ind.Hyg.Assoc.J. 45:177-181.

104. Dahlbäck, G.O., Jonson, B., & Lundgren, C.E.G. 1979. Influence of hydrostatic thorax compression and intra-thoracic blood pooling on dynamic lung mechanics during head-out immersion. Undersea Biomed.Res. 6(Suppl.):23-40.

105. Dahlbäck, G.O., Jönsson, E., & Linér, M.H. 1978. Influence of hydrostatic compression of the chest and intrathoracic blood pooling on static lung mechanics during head-out immersion. Undersea Biomed.Res. 5(1):71-85.

106. Dahlbäck, G.O., & Lundgren, C.E.G. 1972. Pulmonary air-trapping induced by water immersion. Aerospace Med. 43(7):768-774.

107. Daly, W.J., Giammona, S.T., & Ross, J.C. 1965. The pressure-volume relationship of the normal pulmonary capillary bed. J.Clin.Invest. 44(7):1261-1269.

108. Daly, W.J., Giammona, S.T., Ross, J.C., & Feigenbaum, H. 1964. Effects of pulmonary vascular congestion on postural changes in the perfusion and filling of the pulmonary vascular bed. J.Clin.Invest. 43(1):68-76.

109. Daly, W.J., Ross, J.C., & Behnke, R.H. 1963. The effect of changes in the pulmonary vascular bed produced by atropine, pulmonary engorgement, and positive-pressure breathing on diffusing and mechanical properties of the lung. J.Clin.Invest. 42(7):1083-1094.

110. D'Angelo, E., Bonanni, M.V., Michelini, S., & Agostoni, E. 1970. Topography of the pleural surface pressure in rabbits and dogs. Respir.Physiol. 8:204-229.

111. Davis, R.W. 1962. Deep Diving and Submarine Operations. Parts I & II. Siebe Gorman & Co.,Ltd. Chessington, Surrey.

112. Davy, H. 1800. Researches Concerning Nitrous Oxide. Johnson. London.

113. Dempsey, J.A., Hanson, P.A., & Mastenbrook, S.M. 1981. Arterial hypoxemia during heavy exercise in highly trained runners. Fed.Proc. 40:932.

114. Denison, D.M. 1983. The graver side of gravity. Br.J.Dis.Chest 77:105-112.

115. Denison, D.M., Wagner, P.D., Kingaby, G.L., & West, J.B. 1972. Cardiorespiratory responses to exercise in air and underwater. J.Appl.Physiol. 33(4):426-430.

116. Derenne, J.-Ph., Couture, J., Iscoe, S., Whitelaw, W.A., & Milic-Emili, J. 1976. Occlusion pressures in men rebreathing CO_2 under methoxyflurane anesthesia. J.Appl.Physiol. 40:805-814.

117. Derenne, J.-Ph., Macklem, P.T., & Roussos, Ch. 1978. The respiratory muscles: mechanics, control, and pathophysiology. Am.Rev.Respir.Dis. 118:119-133.

118. Derenne, J.-Ph., Macklem, P.T., & Roussos, Ch. 1978. The respiratory muscles: mechanics, control and pathophysiology. Part III. Am.Rev.Respir.Dis. 118:581-601.

119. Dhenin, G. 1978. Aviation Medicine: Physiology and Human Factors. Tri-Med Books. London.

120. Dixon, W.J. 1983. BMDP Statistical Software. University of California Press. Berkeley.

121. Doell, D., Zutter, M., & Anthonisen, N.R. 1973. Ventilatory responses to hypercapnia and hypoxia at 1 and 4 ATA. Resp.Physiol. 18:338-346.

122. Donders, F.C. 1853. Beitrage zum mechanismus der respiration und circulation im gesunden und kranken zustande. Z.Rat.Med. 3:287-319.

123. Donleben, P.G. 1959. Studies on Lung Mechanics. PhD. Thesis. University of Groningen, Netherlands..

124. Dosman, J., Bode, F., Urbanetti, J., Antic, R., Martin, R., & Macklem, P.T. 1975. Role of inertia in the measurement of dynamic compliance. J.Appl.Physiol. 38(1):64-69.

125. Douma, J.H., & Ammerlaan, F.I.A. 1970. Integral method for the calculation of the elastance, resistance and inertance of the human lungs. In: Van Eijnsbergen, B. Progress Report. TNO.NoPR2. Institute of Medical Physics. Utrecht. Pp.81-86.

126. Dressendorfer, R.H., Morlock, J.F., Baker, D.G., & Hong, S.K. 1976. Effects of head-out water immersion on cardiorespiratory responses to maximal cycling exercise. Undersea Biomed.Res. 3(3):177-187.

127. Dressendorfer, R.H.,Wade, C.,& Bernauer, 1977. Combined effects of breathing resistance on aerobic work tolerance. J.Appl.Physiol. 42:444-448.

128. Dreyfuss, D., Basset, G., Soler, P., & Saumon, G. 1985. Intermittent positive-pressure hyperventilation with high inflation pressures produces pulmonary microvascular injury in rats. Am.Rev.Respir.Dis. 132:880-884.

129. DuBois, A.B. 1953. Progress Report. DA-18-108-CML-2212.

130. DuBois, A.B. 1964. Resistance to breathing. In: Fenn, W.O., & Rahn, H. Handbook of Physiology. Respiration I. American Physiological Society. Washington, D.C. Pp.451-462.

131. DuBois, A.B., Botelho, S.Y., Bedell, G.N., Marshall, R., & Comroe, J.H. 1956. A rapid plethysmographic method for measuring thoracic gas volume: a comparison with the nitrogen washout methods for measuring functional redidual capacity in normal subjects. J.Clin.Invest. 35:322-326.

132. DuBois, A.B., Botelho, S.Y., & Comroe, J.H.Jr. 1956. A new method for measuring airway resistance in man using a body plethysmograph: values in normal subjects and in patients with respiratory disease. J.Clin.Invest. 35:327-335.

133. DuBois, A.B., Brody, A.W., Lewis, D.H., & Burgess, B.F.Jr. 1956. Oscillation mechanics of lungs and chest in man. J.Appl.Physiol. 8:587-594.

134. D'Urzo, A.D., Chapman, K.R., & Rebuck, A.S. 1985. Effect of elastic loading on ventilatory pattern during progressive exercise. J.Appl.Physiol. 59(1):34-38.

135. Dwyer, J. 1983. Estimation of oxygen uptake from heart rate response to undersea work. Undersea Biomed.Res. 10(2):77-87.

136. Dwyer, J., Saltzman, H.A., & O'Bryan, R. 1977. Maximal physical-work capacity of man at 43.4 ATA. Undersea Biomed.Res. 4(4):359-372.

137. Eagan, C.J., & Plese, E.R. 1969. Cardiorespiratory and metabolic effects of work during hypo- and hyperbaria. Fed.Proc. 28:593.

138. Echt, M., Lange, L., & Gauer, O. 1974. Changes of peripheral venous tone and central transmural venous pressure during immersion in a thermo-neutral bath. Pflugers Arch. 352:211-217.

139. Ehrner, L. 1960. Lung compliance and respiratory resistance, determined from time-marked esophageal pressure-tidal volume curves and their relation to some other tests of lung function. Acta Med.Scand. 167:Suppl.353:1-220.

140. Ehrner, L, & Nisell, O. 1959. Variability of some characteristics of respiratory mechanics in normal adults. Acta Med.Scand. 164(2):95-103.

141. Ekert, F. 1956. Rontgenkymographische untersuchungen der pwntralen kreislauforgane warend therapeutischer Bader und bei hydrostatischer druckerhohung. Arch.Physik.Therapie. 8:66.

142. Epstein, M. 1976. Cardiovascular and renal effects of head-out water immersion in man. Application of the model in the assessment of volume homeostasis. Circ.Res. 39(5):619-628.

143. Ernsting. J. 1958. The compliance of the human lungs during positive and negative pressure breathing. J.Physiol. 144:14P-16P.

144. Evans, C.L., & Hill, A.V. 1914. The relation of length to tension development and heat production on contraction in muscles. J.Appl.Physiol. 49:1-16.

145. Evans, D.E. 1984. Cardiovascular effects. In: Shilling, C.W., Carlston, C.B., & Mathias, R.A. The Physician's Guide to Diving Medicine. Plenum Press. New York. Pp.99-109.

146. Fagraeus, L. 1974. Cardiorespiratory and metabolic functions during exercise in the hyperbaric environment. Acta Physiol.Scand. Suppl.414:1-40.

147. Fagraeus, L. 1981. Current concepts of dyspnea and ventilatory limitations to exercise at depth. In: Bachrach, A.J., & Matzen, M.M. Underwater Physiology VII. Proceedings of the Seventh Symposium on Underwater Physiology. Undersea Medical Society, Inc., Bethesda, Maryland. Pp.141-149.

148. Fagraeus, L.,& Hesser,C.M. 1970. Ventilatory response to CO_2 in hyperbaric environments. Acta Physiol.Scand. 80:19A-20A.

149. Fagraeus, L., & Linnarsson, D. 1973. Maximal voluntary and exercise ventilation at high ambient air pressures. Forsvarsmedicin. 9:275-278.

150. Farhi, L.E., & Linnarsson, D. 1977. Cardiopulmonary readjustments during graded immersion in water at 35° C. Respir.Physiol. 30:35-50.

151. Farkas, G.A., Decramer, M., Rochester, D.F., & De Troyer, A. 1985. Contractile properties of intercostal muscles and their functional significance. J.Appl.Physiol. 59(2):528-535.

152. Fenn, W.O. 1951. Mechanics of respiration. Am.J.Med. 10:77-90.

153. Fenn, W.O., Otis, A.B., Rahn, H., Chadwick, L.E., & Hegnauer, A.H. 1947. Displacement of blood from the lungs by pressure breathing. Am.J.Physiol. 151:258-269.

154. Ferris, B.G.Jr., Mead, J., & Opie, L.H. 1964. Partitioning of respiratory flow resistance in man. J.Appl.Physiol. 19(4):653-658.

155. Ferris, B.G.Jr., & Pollard, D.S. 1960. Effect of deep and quiet breathing on pulmonary compliance in man. J.Clin.Invest. 39:143-149.

156. Field, S., Sanci, S., & Grassino, A. 1984. Respiratory muscle oxygen consumption estimated by the diaphragm pressure-time index. J.Appl.Physiol. 57(1):44-51.

157. Finucane, K.E, Egan, B.A., & Dawson, S.V. 1972. Linearity and frequency response of pneumotachographs. J.Appl.Physiol. 32(1):121-126.

158. Florio, J.T., Morrison, J.B., & Butt, W.S. 1979. Breathing pattern and ventilatory response to carbon dioxide in divers. J.Appl.Physiol. 46:1076-1080.

159. Flynn, E.T.Jr., Camporesi, E.M., & Nunneley, S.A. 1975. Cadiopulmonary responses to pressure breathing during immersion in water. In: Lanphier, E.H., & Rahn, H. Man Water Pressure. Volume II. State University of New York. Buffalo, New York. Pp.79-94.

160. Frank, N.R. 1959. Influence of acute pulmonary vascular congestion on recoiling force of excised cats' lung. J.Appl.Physiol. 14(6):905-908.

161. Fritts, H.W.Jr., Filler, J., Fishman, A.P., Cournand, A. 1959. The efficiency of ventilation during voluntary hyperpnea: studies in normal subjects and in dyspneic patients with either chronic pulmonary emphysema or obesity. J.Clin.Invest. 38:1339-1348.

162. Fry, D.L. 1960. Physiologic recording by modern instruments with particular reference to pressure recording. Physiol.Rev. 40:753-788.

163. Fry, D.L., Ebert, R.V., Stead, W.W., & Brown, C.C. 1954. The mechanics of pulmonary ventilation in normal subjects and in patients with emphysema. Am.J.Med. 16:80-97.

164. Fry, D.L., & Hyatt, R.E. 1960. Pulmonary mechanics: a unified analysis of the relationship between pressure, volume and gasflow in the lungs of normal and diseased human subjects. Am.J.Med. 29:672-689.

165. Fry, D.L., Stead, W.W., Ebert, R.V., Lubin, R.I., & Wells, H.S. 1952. The measurement of intraesophageal pressure and its relationship to intrathoracic pressure. J.Lab.Clin.Med. 40:664-673.

166. Gauer, O.H. 1975. Recent advances in the physiology of whole body immersion. Acta Astron. 2:31-39.

167. Gautier, H., Peslin, R., Grassino, A., Milic-Emili, J., Hannhart, B, Powell, E., Miserocchi, G. Bonora, M. & Fischer, J.T. 1982. Mechanical properties of the lungs during acclimatization to altitude. J.Appl.Physiol. 52(6):1407-1415.

168. Gehring, R.E. 1978. Basic Behavioural Statistics. Houghton Mifflin. Boston.

169. Gelfand, R. 1984. Invited Review. Concepts of ventilatory and respiratory gas homeostasis in simulated undersea exposure. In: Bachrach, A.J., & Matzen, M.M. Underwater Physiology VIII. Proceedings of the Eighth Symposium on Underwater Physiology. Undersea Medical Society. Bethesda, Maryland. Pp.515-533

170. Gelfand, R., Lambertsen, C.J., & Peterson, R.E. 1980. Human respiratory control at high ambient pressures and inspired gas densities. J.Appl.Physiol. 48(3):528-539.

171. Giannelli, S.Jr., Ayres, S.M., & Buehler, M.E. 1967. Effect of pulmonary blood flow upon lung mechanics. J.Clin.Invest. 46(10):1625-1642.

172. Gibney, R.T.N., Wilson, R.S., & Pontoppidan, H. 1982. Comparison of work of breathing on high gas flow and demand valve continuous positive airway pressure systems. Chest 82:6692-695.

173. Gibson, G.J., & Pride, N.B. 1976. Lung Distensibility. The static pressure-volume curve of the lungs and its use in clinical assessment. Br.J.Dis.Chest. 70:143-184.

174. Gilbert, R., Auchincloss, J.H.Jr., Brodsky, J., & Boden, W. 1972. Changes in tidal volume, frequency, and ventilation induced by their measurement. J.Appl.Physiol. 33:252-254.

175. Giuntini, C., Maseri, A., & Bianchi, R. 1966. Pulmonary vascular distensibility and lung compliance as modified by dextran infusion and subsequent atropine injection in normal subjects. J.Clin.Invest. 45(11):1770-1789.

176. Glaister, D.H., Schroter, R.C., Sudlow, M.F., & Milic-Emili, J. 1973. Bulk elastic properties of excised lungs and the effect of a transpulmonary pressure gradient. Respir.Physiol. 17:347-364.

177. Glauser, S.C., Glauser, E.M., & Rusy, B.F. 1967. Gas density and the work of breathing. Respir.Physiol. 2:344-350.

178. Glazier, J.B., Hughes, J.M.B., Maloney, J.E., & West, J.B. 1967. Vertical gradient of alveolar size in lungs of dogs frozen intact. J.Appl.Physiol. 23:694-705.

179. Goldberg, H.S., Mitzner, W., Adams, K., Menkes, H., Lichtenstein, S., & Permutt, S. 1975. Effect of intrathoracic pressure on pressure-volume characteristics of the lung in man. J.Appl.Physiol. 38(3):411-417.

180. Goldman, M.D., Grimby, G., & Mead, J. 1976. Mechanical work of breathing derived from rib cage and abdominal V-P partitioning. J.Appl.Physiol. 41(5):752-763.

181. Goldman, M.D., Grassino, A., Mead, J., & Sears, T.A. 1978. Mechanics of the human diaphragm during voluntary contraction: dynamics. J.Appl.Physiol. 44:840-848.

182. Goldman, M.D., & Mead, J. 1973. Mechanical interaction between the diaphragm and rib cage. J.Appl.Physiol. 35(2):197-204.

183. Granath, A., Horie, E., & Linderholm, H. 1959. Compliance and resistance of the lungs in the sitting and supine positions at rest and during work. Scand.J.Clin.Lab.Invest. 11:226-234.

184. Gray, H. 1977. Anatomy. Descriptive and Surgical. Bounty Books, New York.

185. Green, M. 1965. The effects of changes in gas viscosity on measurement of gas flow rates and volumes by the pneumotachograph. Proc.Physiol.Soc. 181(2):19P-20P.

186. Greene, R., Hughes, J.M.B., Sudlow, M.F., & Milic-Emili, J. 1974. Regional lung volumes during water immersion to the xiphoid in seated man. J.Appl.Physiol. 36(6):734-736.

187. Grimby, G., Goldman, M. & Mead, J. 1976. Respiratory muscle action inferred from rib cage and abdominal V-P partitioning. J.Appl.Physiol. 41(5):739-751.

188. Grismer, J.T., & Goodwin, M.H. 1983. Field study of ventilation in volunteer scuba divers during head-out immersion. Undersea Biomed.Res. 10(4):311-320.

189. Guyatt, A.R., Newman, F., Cinkotai, F.F., Palmer, J.I., & Thomson, M.L. 1965. Pulmonary diffusing capacity in man during immersion in water. J.Appl.Physiol. 20(5):878-881.

190. Guyton, A.C. 1963. Venous return. In: Handbook of Physiology. Circulation. American Physiological Society. Washington, D.C.

191. Hajduczok, Z., Hickey, D.D., & Lundgren, C.E.G. 1984. Cardiac output adjustment to sudden immersion determined by impedance cardiography. In: Bachrach, A.J., & Matzen, M.M. Underwater Physiology VIII. Proceedings of the Eighth Symposium on Underwater Physiology. Undersea Medical Society. Bethesda, Maryland. Pp.295-300.

192. Haldane, J.S. 1907. Deep water diving. Report of a committee appointed by the Lords Commissioners of the Admiralty. Parliamentary Paper. CN1549. Pp.1-84.

193. Hamilton, W.F., & Mayo, J.P. 1944. Changes in the vital capacity when the body is immersed in water. Am.J.Physiol. 141:51-53.

194. Hartshorne, H. 1847. Water versus Hydrotherapy or an Essay on Water and its True Relations to Medicine. Lloyd P.Smith Press. Philadelphia.

195. Hedenstierna, G., & McCarthy, G. 1975. Mechanics of breathing, gas distribution and functional residual capacity at different frequencies of respiration during spontaneous and artificial ventilation. Br.J.Anaest. 47:706-712.

196. Henning, R.J., Shubin, H., & Weil, M.H. 1977. The measurement of the work of breathing for the clinical assessment of ventilator dependence. Crit.Care Med. 5:264-268.

197. Henry, F.M. 1949. Errors in measurement. In: Scott, G. Research Methods in Health and Physical Education. Washington, D.C.

198. Henry, W.C., West, G.A., & Wilson, R.S. 1983. A comparison of the oxygen cost of breathing between a continuous-flow CPAP system and a demand-flow CPAP system. Resp.Care 28:1273-1281..

199. Hesser, C.M., Adolfson, J., & Fagraeus, L. 1971. Role of CO_2 in compressed-air narcosis. Aerospace Med. 42:163-168.

200. Hesser, C.M., Fagraeus, L, & Adolfson, M. 1978. Roles of N_2, O_2, and CO_2 in compressed-air narcosis. Undersea Biomed.Res. 5(4):391-400.

201. Hesser, C.M., Fagraeus, L., & Linnarsson, D. 1968. Cardio-respiratory responses to exercise in hyperbaric environment. Report from the Laboratory of Aviation & Naval Medicine. Karolinska Institutet, Stockholm. Pp.1-11.

202. Hesser, C.M., & Lind, F. 1981. Effects of exercise and hyperbaric air on ventilation and central inspiratory activity. In: Bachrach, A.J. & Matzen, M.M. Underwater Physiology VII. Proceedings of the Seventh Symposium on Underwater Physiology. Undersea Medical Society, Inc., Bethesda, Maryland.

203. Hesser, C.M., Linnarsson, D., & Fagraeus, L. 1981. Pulmonary mechanics and work of breathing at maximal ventilation and raised air pressure. J.Appl.Physiol. 50(4):747-753.

204. Hickey, D.D., Lundgren, C.E.G., & Pasche, A.J. 1983. Influence of exercise on maximal voluntary ventilation and forced expiratory flow at depth. Undersea Biomed.Res. 10(3):241-254.

205. Hirschfelder, J.O., Curtiss, C.F., & Bird, R.B. 1954. The transport properties of dense gases. In: Molecular Theory of Gases and Liquids. John Wiley. New York. Pp.611-667.

206. Hoff, E.C. 1948. A Bibliographical Sourcebook of Compressed Air, Diving and Submarine Medicine. Research Division, Project X-427, Bureau of Medicine and Surgery, Navy Department, Washington, D.C. Pp.1-328.

207. Hoff, E.C., & Greenbaum, L.J.Jr. 1954. A Bibliographical Sourcebook of Compressed Air, Diving and Submarine Medicine. Volume II. Office of Naval Research & Bureau of Medicine and Surgery, Department of the Navy, Washington, D.C. Pp.71-298.

208. Hogg, J.C., Macklem, P.T., & Thurlbeck, W.M. 1968. Site and nature of airway obstruction in chronic obstructive lung disease. N.Eng.J.Med. 278:1355-1360.

209. Hogg, J.C., & Nepszy, S. 1969. Regional lung volume and pleural pressure gradient estimated from lung density in dogs. J.Appl.Physiol. 27:198-203.

210. Holloszy, J.O., & Booth, F.W. 1976. Biochemical adaptations to endurance exercise in muscle. Am.Rev.Physiol. 38:273-291.

211. Holmgrer., A. 1967. Cardiorespiratory determinants of cardiovascular fitness. Can.Med.Ass.J. 96:697-705.

212. Holmgren, A., Herzog, P., & Astrom, H. 1973. Work of breathing during exercise in healthy young men and women. Scand.J.Clin.Lab.Invest. 31(2):165-174.

213. Hong,S.K., Cerretelli, P., Cruz, J.C., & Rahn, H. 1969. Mechanics of respiration during submersion in water. J.Appl.Physiol. 27:535-538.

214. Hong, S.K., Ting, E.Y., & Rahn, H. 1960. Lung volumes at different depths of submersion. J.Appl.Physiol. 15(4):550-553.

215. Hood, W.B., Murray, R.H., Urschel, C.W., Bowers, J.A., & Goldman, J.K. 1968. Circulatory effects of water immersion upon human subjects. Aerospace Med. 39:579-584.

216. Hubay, C.A., Waltz, R.C., Brecher, G.A., Praglin, J., & Hingson, R.A. 1954. Circulatory dynamics of venous return during positive-negative pressure respiration. Anaesthesia 15(5):445-461.

217. Huck, S.W., Cormier, W.H., & Bounds, W.G. 1974. Reading Statistics and Research. Harper and Row. New York.

218. Hughes, J.M.B., Glazier, J.B., Maloney, J.E., & West, J.B. 1968. Effect of lung volume on the distribution of pulmonary blood flow in man. Respir.Physiol. 4:58-72.

219. Hunt, N.C. 1967. Positive pressure breathing during water immersion. Aerospace Med. 38:731-735.

220. Hutchinson, J. 1846. On the capacity of the lungs, and on the respiratory functions, with a view of detecting disease by the spirometer. Med.Chir.Trans. (Series 2) 11:137-252.

221. Hutchinson, J. 1849-1852. Thorax. In: Todd Cyclopaedia of Anatomy and Physiology. Volume 4, Part 2, p.1016. London.

222. Hyatt, R.E., Schilder, D.P., & Fry, D.L. 1958. Relationship between maximum expiratory flow and degree of lung inflation. J.Appl.Physiol. 13:331-336.

223. Hyatt, R.E., & Wilcox, R.E. 1961. Extrathoracic airway resistance in man. J.Appl.Physiol. 16:326-330.

224. Ingram, R.H.Jr., O'Cain, C.F., & Fridy, W.W.Jr. 1974. Simultaneous quasi-static lung pressure-volume curves and "closing volume" measurements. J.Appl.Physiol. 36(2):135-141.

225. Ishi, M., Matsumoto, N., Fuyuki, T., Hida, W., Ichinose, M., Inoue, H., & Takishima, T. 1985. Effects of hemodynamic edema formation on peripheral versus central airway mechanics. J.Appl.Physiol. 59(5):1578-1584.

226. Jaeger, M.J., & Matthys, H. 1970. The pressure flow characteristics of the human airways. In: Bouhuys, A. Airway Dynamics; Physiology and Pharmacology. Charles C.Thomas. Springfield. Pp.21-32.

227. Jaeger, M.J., & Otis, A.B. 1964. Effects of compressibility of alveolar gas on dynamics and work of breathing. J.Appl.Physiol. 19(1):83-91.

228. Jansson, L, & Jonson, B. 1975. A method for studies of pulmonary mechanics at rest and exercise. Department of Clinical Physiology, University of Lund, Sweden. S-221 85. Pp.1-17.

229. Jaquet, A. 1908. Zur Mechanik der Atembewegungen. Arch.f.exper.Path.u.Pharmakol. Suppl. 59:309-316.

230. Jarrett, A.S. 1965. Effect of immersion on intrapulmonary pressure. J.Appl.Physiol. 20(6):1261-1266.

231. Jarrett, A.S. 1966. Alveolar carbon dioxide tension at increased ambient pressures. J.Appl.Physiol. 21(1):158-162.

232. Johnson, B.L., Stromme, S.B., Adamczyk, J.W., & Tennoe, K.O. 1977. Comparison of oxygen uptake and heart rate during exercises on land and in water. Phys.Ther. 57(3):273-278.

233. Johnson, L.F., Lin, Y.C., & Hong, S.K. 1975. Gastroesophageal dynamics during immersion in water to the neck. J.Appl.Physiol. 38(3):449-454.

234. Johnson, L.F., & Mead, J. 1963. Volume-pressure relationships during pressure breathing and voluntary relaxation. J.Appl.Physiol. 18:505-508.

235. Jones, J.G., Lemen, R., & Graf, P.D. 1978. Changes in airway calibre following pulmonary venous congestion. Br.J.Anaest. 50:743-752.

236. Jones, N.L. 1984. Dyspnea in exercise. Med.Sci.Sports Exerc. 16(1):14-19.

237. Jones, N.L., Robertson, D.G., & Kane, J.W. 1979. Difference between end-tidal and arterial PCO_2 in exercise. J.Appl.Physiol. 47(5):954-960.

238. Jonson, B. 1969. A method for determination of pulmonary elastic recoil and resistance at a regulated flow rate. Scand.J.Clin.Lab.Invest. 24:115-125.

239. Jonson, B. 1970. Pulmonary resistance and elastic recoil as determinants of maximum efficiency of ventilation. Bull.Physiol-path.Resp. 6:865-874.

240. Jonson, B. 1971. Pulmonary mechanics as a factor limiting the capacity for work in disease. Scand.J.Resp.Dis. 77:94-97.

241. Jonson, B. 1971. The work of breathing: pulmonary mechanics at exercise. Scand.J.Resp.Dis. 77:14-18.

242. Jonson, B., & Olsson, L.G. 1971. Measurement of the work of breathing to overcome pulmonary viscous resistance. Scand.J.Clin.Lab.Invest. 28:135-140.

243. Keppel, G. 1973. Design and Analysis: a Researcher's Handbook. Prentice-Hall. New Jersey.

244. Kilburn, K.H., & Sieker, H.O. 1960. Hemodynamic effects of continuous positive and negative pressure breathing in normal man. Circ.Res. 8:660-669.

245. Killian, K.J., & Campbell, E.J.M. 1983. Dyspnea and exercise. Annu.Rev.Physiol. 45:465-479.

246. Killian, K.J., & Jones, N.L. 1984. The use of exercise testing and other methods in the investigation of dyspnea. Clin.Chest Med. 5(1):99-108.

247. King, A.L., & Lawton, R.W. 1950. Elasticity of body tissues. U.Physics. 2:303.

248. Knowles, J.H. Hong, S.K., & Rahn, H. 1960. Possible errors using the esophageal balloon in the determination of pressure-volume characteristics of the lung and thoracic cage in various postures. Wright Air Development Division. Technical report 60-1.

249. Krasney,J.A., Hajduczok,G., Akiba,C., McDonald, B.W., Pendergast, D.R., & Hong, S.K.1984. Cardiovascular and renal responses to head-out water immersion in canine model. Undersea Biomed.Res. 11(2):169-183.

250. Kroetz, C., & Wachter, R. 1933. Uber das Minutenvolumen des Herzens in verschiedenen Baderarten. Klin.Wochschr. 12:1517-1520.

251. Kuckelt, W., Scharfenberg, J., Mrochen, H., Dauberschmidt, R., Petrakov, G., Kassil, W., & Meyer, M. 1981. Effect of PEEP on gas exchange, pulmonary mechanics, and hemodynamics in adult respiratory distress syndrome (ARDS). Intensive Care Med. 7:177-185.

252. Kuo, S.S. 1972. Computer Applications of Numerical Methods. Addison Wesley. Reading, Massachusetts.

253. Kurss, D.I., Lundgren, C.E.G., & Pasche, A.J. 1981. Effect of water temperature and vital capacity in head-out immersion. In: Bachrach, A.J., & Matzen, M.M. Underwater Physiology VII. Proceedings ofthe Seventh Symposium on Underwater Physiology. Undersea Medical Society. Bethesda, Maryland. Pp.297-301.

254. Labadie, H., & Van Eenige, M.J. 1969. Analysis of static volume-pressure curves in obstructive lung disease. In: Orie, N.G.M., & Van Der Lende, R. 1969. Bronchitis III: Proceedings of the Third International Symposium on Bronchitis. Groningen, Netherlands. Royal Van Gorcum, Netherlands. Pp.241-256.

255. Lally, D.A., Moore, T.O., & Hong, S.K. 1971. Cardio-respiratory responses to exercise in air and water at 1 and 2 ATA. Department of Physiology, University of Hawaii School of Medicine. UHI-med-71-01. Pp.1-22.

256. Lally, D.A., Zechman, F.W., & Tracy, R.A. 1974. Ventilatory responses to exercise in divers and non-divers. Respir.Physiol. 20:117-129.

257. Lambertsen, C.J. 1976. Collaborative investigation of limits of human tolerance to pressurization with helium, neon and nitrogen, simulation of density equivalent to helium-oxygen respiration at depth to 2000, 3000, 4000 and 5000 feet of sea water. In: Lambertsen, C.J. Underwater Physiology V. Proceedings of Fifth Symposium on Underwater Physiology. Undersea Medical Society. Bethesda.

258. Lambertsen, C.J., Gelfand, R., Lever, M.J., Bodammer, G., Takano, N., Reed, T.A., Dickson, J.G., & Watson, P.T. 1973. Respiration and gas exchange during a 14-day continuous exposure to 5.2% O_2 in N_2 at pressure equivalent to 100 FSW (4 ata). Aerospace Med. 44:844-849.

259. Lambertsen, C.J., Hall, P., Wollman, H., & Goodman, M.W. 1963. Quantitative interactions of increased PO_2 and PCO_2 upon respiration in man. Ann.N.Y.Acad.Sci. 109:731-741.

260. Lange, L., Lange, S., Echt, M., & Gauer, O.H. 1974. Heart volume in relation to body posture and immersion in a thermo-neutral bath. Pflugers Arch. 352:219-226.

261. Lanphier, E.H. 1963. Influence of increased ambient pressure upon alveolar ventilation. In: Lambertsen, C.J., & Greenbaum, L.J. Underwater Physiology. Proceedings of the Second Symposium on Underwater Physiology. National Academy of Science, Washington,D.C. Pp.124-133.

262. Lanphier, E.H., & Camporesi, E.M. 1982. Respiration and exercise. In:Bennett, P.B., & Elliott, D.H. The Physiology and Medicine of Diving. Bailliere Tindall.London. Pp.99-156.

263. Lapedes, D.N. 1978. Dictionary of Physics and Mathematics. McGraw-Hill. New York.

264. Larson, H.E. 1959. A History of Self-Contained Diving and Underwater Swimming. NAS/NRC Publication 469. Washington, D.C.

265. Lawton, R.W., & King, A.L. 1949. Elasticity of the rat lung. Fed.Proc. 8:92.

266. Leaver, D.G., & Pride, N.B. 1971. Flow-volume curves and expiratory pressures during exercise in patients with chronic airways obstruction. Scand.J.Resp.Dis. Suppl.77:23-27.

267. Ledsome, R.R., & Linden, R.J. 1964. A reflex increase in heart rate from distension of the pulmonary vein-atrial junctions. J.Physiol. London. 170:456-473.

268. Lemen, R., Benson, M., & Jones, J.G. 1974. Absolute pressure measurements with hand-dipped and manufactured esophageal balloons. J.Appl.Physiol. 37(4):600-603.

269. LeRoy, N.B., & Guerrant, J.L. 1965. Breathing mechanics in asthma. Ann.Intern.Med. 63:572-582.

270. Levine, O.R., Mellins, R.B., & Fishman, A.P. 1965. Quantitative assessment of pulmonary edema. Circ.Res. 17:414-426.

271. Levison, H., & Cherniack, R.M. 1968. Ventilatory cost of exercise in chronic obstructive pulmonary disease. J.Appl.Physiol. 25:21-27.

272. Lin, Y.C. 1984. Circulatory functions during immersion and breath-hold dives in humans. Undersea Biomed.Res. 11(2):123-138.

273. Linnarsson, D., & Fagraeus, L. 1976. Maximal work performance in hyperbaric air. In: Lambertsen, C.J. Underwater Physiology V. Proceedings of the Fifth Symposium on Underwater Physiology. FASEB. Bethesda, Maryland. Pp.55-60.

274. Linnarsson, D., & Hesser, C.M. 1978. Dissociated ventilatory and central respiratory responses to CO_2 at raised N_2 pressure. J.Appl.Physiol. 45:756-761.

275. Löllgen, H., Nieding, G.v., & Horres, R. 1980. Respiratory and hemodynamic adjustment during head out water immersion. Int.J.Sports Med. 1:25-29.

276. Long, E.C., Hull, W.E., & Gebel, E.L. 1962. Respiratory dynamic resistance. J.Appl.Physiol. 17(4):609-612.

277. Lourenco, R.V., Turino, G.M., Davidson, L.A.G., & Fishman, A.P. 1965. The regulation of ventilation in diffuse pulmonary fibrosis. Am.J.Med. 38:199-216.

278. Lord, G.P., Bond, G.F., & Schaefer, K.E. 1967. Breathing under high ambient pressure. J.Appl.Physiol. 21(6):1833-1838.

279. Luciani, L. 1878. Delle oscillazioni della pressione intratoracica e intraddominale. Arch.Sci.Med. 2:177-224.

280. Ludwig, C. 1847. Beiträge zur Kenntnis des Einflusses der Respirationsbewegungen auf den Blutlauf im Aortensysteme. Arch.Anat.Physiol.Wiss.Med. :242-302.

281. Lundgren, C.E.G. 1984. Respiratory function during simulated wet dives. Undersea Biomed.Res. 11(2):139-147.

282. Lundgren, C.E.G., & Pasche, A.J. 1984. Immersion effects. In: Shilling, C.W., Carlston, C.B., & Mathias, R.A. The Physician's Guide to Diving Medicine. Plenum Press. New York. Pp.86-99.

283. Lundsgaard, C., & Schierbeck, K. 1923. Untersuchungen über die Volumina der Lungen. I-IV. Acta Med.Scand. 58:470.

284. Lynne-Davies, Couture, J., Pengelly, L.D., West, D., Bromage, P.R., & Milic-Emili,J. 1971. Partitioning of immediate ventilatory stability to added elastic loads in cats. J.Appl.Physiol. 30:814-819.

285. Macdonald, J.W., & Pilmanis, A.A. 1981. Carbon dioxide retention with underwater work in the open ocean. In: Bachrach, A.J., & Matzen, M.M. Underwater Physiology VII. Proceedings of the Seventh Symposium on Underwater Physiology. Undersea Medical Society, Inc., Bethesda, Maryland. Pp.197-207.

286. Macklem, P.T. 1978. Respiratory mechanics. Annu.Rev.Physiol. 40:157-184.

287. Macklem, P.T. 1979. Rib cage-diaphragm interaction. Am.Rev.Respir.Dis. 119:27-29.

288. Macklem, P.T. 1980. Respiratory muscles: the vital pump. Chest 78(5):753-758.

289. Macklem, P.T., & Mead, J. 1967. Resistance of central and peripheral airways measured by a retrograde catheter. J.Appl.Physiol. 22:395–401.

290. Maio, D.A., & Farhi, L.E. 1967. Effect of gas density on mechanics of breathing. J.Appl.Physiol. 23(5):687–693.

291. Manco, J.C., & Hyatt, R.E. 1975. Relationship of air trapping to increased lung recoil pressure induced by chest cage restriction. Am.Rev.Respir.Dis. 111:21–26.

292. Margaria, R., Milic–Emili, G., Petit, J.M., & Cavagna, G. 1960. Mechanical work of breathing during muscular exercise. J.Appl.Physiol. 15(3):354–358.

293. Marshall, R. 1957. The physical properties of the lungs in relation to the subdivisions of lung volume. Clin.Sci. 16:507–515.

294. Marshall, R. 1962. Relationships between stimulus and work of breathing at different lung volumes. J.Appl.Physiol. 17(6):917–921.

295. Marshall, R., & DuBois, A.B. 1956. The measurement of the viscous resistance of the lung tissues in normal man. Clin.Sci. 15:161–170.

296. Marshall, R., McIlroy, M.B., & Christie, R.V. 1954. The work of breathing in mitral stenosis. Clin.Sci. 13:137–146..

297. Martin, J.G., Habib, M., & Engel, L.A. 1980. Inspiratory muscle activity during induced hyperinflation. Respir.Physiol. 39:303–313.

298. Martin, J.G., Shore, S., & Engel, L.A. 1982. Effect of continuous positive airway pressure on respiratory mechanics and pattern of breathing in induced asthma. Am.Rev.Respir.Dis. 126:812–817.

299. Maxwell, D.L., Cover, D., & Hughes, J.M.B. 1985. Effect of respiratory apparatus on timing and depth of breathing in man. Respir.Physiol. 61:255–264.

300. McCall, C.B., Hyatt, R.E., Noble, F.W., & Fry, D.L. 1957. Harmonic content of certain respiratory flow phenomena of normal individuals. J.Appl.Physiol. 10(2):215–218.

301. McDermott, M. 1966. Diurnal and weekly cyclical changes in lung airways resistance. J.Physiol. 186:P90–P91.

302. McGregor, M., & Becklake, M.R. 1961. The relationship of oxygen cost of breathing to respiratory mechanical work and respiratory force. J.Clin.Invest. 40:971–980.

303. McIlroy, M.B., & Christie, R.V. 1954. The work of breathing in emphysema. Clin.Sci. 13:147–154.

304. McIlroy, M.B., Mead, J., Selverstone, N.J., & Radford, E.P.Jr. 1955. Measurement of lung tissue viscous resistance using gases of equal kinematic viscosity. J.Appl.Physiol. 7:485–490.

305. McIlroy, M.B., Marshall, R., & Christie, R.V. 1954. The work of breathing in normal subjects. Clin.Sci. 13:127–136.

306. McKenna, W.J., Griffin, P.M., Anthonisen, N.R., & Menkes, H.A. 1973. Oscillation mechanics of the submerged respiratory system. Aerospace Med. 44(3):324–326.

307. McMinn, R.M.H., & Hutchings, R.T. 1977. Color Atlas of Human Anatomy. Year Book Medical Publishers. Chicago.

308. McMinn, R.M.H., & Hutchings, R.T. 1983. Color Atlas of Human Anatomy. Year Book Medical Publishers. Chicago.

309. McWhirter, J., Harrison, P., & Florio, J. 1979. The effect of immersion on $FEV_{1.0}$/FVC. Admiralty Marine Technology Establishment. England. Report 404. Pp.1–17.

310. Mead, J. 1955. Resistance to breathing at increased ambient pressures. In: Goff, L.G. Underwater Physiology I. Proceedings of the First Symposium on Underwater Physiology. National Academy of Science, National Research Council. Publ.377. Pp.112-123.

311. Mead, J. 1956. Measurement of inertia of the lungs at increased ambient pressure. J.Appl.Physiol. 9:208-212.

312. Mead, J. 1961. Mechanical properties of lungs. Physiol.Rev. 41:281-330.

313. Mead, J. 1979. Responses to loaded breathing. Bull.Europ.Physiopathol.Respir. 15:61-71.

314. Mead, J. 1980. Expiratory flow limitation: a physiologist's point of view. Fed.Proc. 39:2771-2775.

315. Mead, J. 1981. Mechanics of chest wall. Operational length compensation - a new respiratory reflex. Jap.J. of Thoracic Dis. 19(10):683.

316. Mead, J., & Agostoni, E. 1964. Dynamics of breathing. In: Fenn, W.O., & Rahn, H. Handbook of Physiology. Respiration I. American Physiological Society. Washington, D.C. Pp.411-427.

317. Mead, J., McIlroy, M.B., Selverstone, N.J., & Kriete, B.C. 1955. Measurement of intraesophageal pressure. J.Appl.Physiol. 7:491-495.

318. Mead, J., & Milic-Emili, J. 1964. Theory and methodology in respiratory mechanics with glossary of symbols. In: Fenn, W.O., & Rahn, H. Handbook of Physiology. Respiration I. American Physiological Society. Washington. Pp.363-376.

319. Mead, J., Turner, J.M., Macklem, P.T., & Little, J.B. 1967. Significance of the relationship between lung recoil and maximum expiratory flow. J.Appl.Physiol. 22:95-108.

320. Mead, J., & Whittenberger, J.L. 1953. Physical properties of human lungs measured during spontaneous respiration. J.Appl.Physiol. 5:779-796.

321. Mead, J., & Whittenberger, J.L. 1954. Evaluation of airway interruption technique as a method for measuring pulmonary air-flow resistance. J.Appl.Physiol. 6:408-416.

322. Milic-Emili, G., & Cajani, F. 1957. La frequenza dei respiri in funzione della ventilazione durante la marcia. Boll.Soc.Ital.Biol.Sper. 33:825-827.

323. Milic-Emili, G., & Petit, J.M. 1960. Mechanical efficiency of breathing. J.Appl.Physiol. 15(3):359-362.

324. Milic-Emili, G., Petit, J.M., & Derouanne, R. 1962. Mechanical work of breathing during exercise in trained and untrained subjects. J.Appl.Physiol. 17(1):43-46.

325. Milic-Emili, J. 1984. Measurement of pressures in respiratory physiology. Respir.Physiol. P412:1-22.

326. Milic-Emili, J., Grassino, A.E., & Whitelaw, W.A. 1981. Measurement and testing of respiratory drive. In: Hornbein, T.F. Regulation of breathing. Part II. Marcel Dekker. New York. Pp.675-743.

327. Milic-Emili, J., Henderson, J.A.M., Dolovich, M.B., Trop, D., & Kaneko, K. 1966. Regional distribution of inspired gas in the lung. J.Appl.Physiol. 21:749-759.

328. Milic-Emili, J., Mazzarelli, M. Derenne, J.-Ph., Whitelaw, W.A., & Couture, J. 1975. A new approach to study control of breathing. Clin.Res. 23:646A.

329. Milic-Emili, J., Mead, J., & Turner, J.M. 1964b. Topography of esophageal pressure as a function of posture in man. J.Appl.Physiol. 19(2):212-216.

330. Milic-Emili, J., Mead, J., Turner, J.M., & Glauser, E.M. 1964a. Improved technique for estimating pleural pressure from oesophageal balloons. J.Appl.Physiol. 19(2):207-211.

331. Milic-Emili, J., & Tyler, J.M. 1963. Relation between work output of respiratory muscles and end-tidal CO_2 tension. J.Appl.Physiol. 18(3):497-504.

332. Miller, J.N., Wangensteen, O.D., & Lanphier, E.H. 1971. Ventilatory limitations on exertion at depth. In: Lambertsen, C.J. Underwater Physiology. Proceedings of Fourth Symposium on Underwater Physiology. Academic Press. New York. Pp.317-323.

333. Miller, J.P., Cunningham, D.J.C., Lloyd, B.B., & Younger, J.M. 1974. The transient respiratory effects in man of sudden changes in alveolar CO_2 in hypoxia and in high oxygen. Respir.Physiol. 20:17-31.

334. Milne, D.M., & Morrison, J.B. 1979. Hydrostatic pressure imbalance with open circuit underwater breathing apparatus. Ergonomics. 22(3):299-306.

335. Minh, V., Dolan, G.F., Linaweaver, P.G., Friedman, P.J., Konopka, R.G., & Brach, B.B. 1977. Diaphragmatic function during immersion. J.Appl.Physiol. 43(2):297-301.

336. Minh,V., Dolan, G.F., Nam, D.D., Linaweaver, P.G., & Harvey, C. 1979. Immersion versus pressure-breathing and diaphragmatic function in the upright position. Respir.Physiol. 36:39-49.

337. Miserocchi, G., Nakamura, T., Mariani, E., & Negrini, D. 1981. Pleural liquid pressure over the interlobar mediastinal and diaphragmatic surfaces of the lung. Respir.Physiol. 46:61-69.

338. Mittman, C., Edelman, N.H., Norris, A.H., & Shock, N.W. 1965. Relationship between chest wall and pulmonary compliance and age. J.Appl.Physiol. 20(6):1211-1216.

339. Mognoni, P., Saibene, F., & Veicsteinas, A. 1982. Ventilatory work during exercise at high altitude. Int.J.Sports Med. 3:33-36.

340. Morrison, J.B. 1973. Oxygen uptake studies of divers when fin swimming with maximum effort at depths of 6-176 feet. Aerospace Med. 44(10):1120-1129.

341. Morrison, J.B. 1983. The influence of breathing system design on the capacity of divers to work at depth. Offshore Goteborg, 83. Svenska Massan Stiftelse, Goteborg, Sweden.

342. Morrison, J.B., & Butt, W.S. 1972. The effect of underwater breathing apparatus and absolute air pressure on divers' ventilatory capacity. Aerospace Med. 43:881-886.

343. Morrison, J.B., Butt, W.S., Florio, J.T., & Mayo, I.C. 1976. Effects of increased O_2 - N_2 pressure and breathing apparatus on respiratory function. Undersea Biomed.Res. 3(3):217-234.

344. Morrison, J.B., & Florio, J.T. 1971. Respiratory function during a simulated saturation dive to 1,500 feet. J.Appl.Physiol. 30(5):724-732.

345. Morrison, J.B., Florio, J.T., & Butt, W.S. 1981. Effects of CO_2 insensitivity and respiratory pattern on respiration in divers. Undersea Biomed.Res. 8:209-217.

346. Morrison, J.B., Morariu, G., & Wood, I. 1986. Performance characteristics of open circuit demand regulators from 0.5 to 50 metres seawater. Annals Physiol.Anthropol. 5(3):169.

347. Morrison, J.B., & Reimers, S.D. 1982. Design principles of underwater breathing apparatus. In: Bennett, P.B., & Elliott, D.H. The Physiology and Medicine of Diving. Bailliere Tindall. London. Pp.55-98.

348. Morrison, J.B., & Wood, I. 1986. Ventilatory responses and perceived respiratory effort of divers working at maximal physical capacity. Annals Physiol.Anthropol. 5(3):169.

349. Müller, E., Vogel, J., Wuthe, H., & Petro, W. 1976. Automated analysis of changes in parameters of lung mechanics with alterations in breathing frequency, lung volume, and physical effort. Z.Erkr.Atmungsorgane 145:274–283.

350. Neergaard, K.v. 1929. Neue Auffassungen uber einen Grundbegriff der Atemmechanik. Ztschr.f.d.ges.exper.Med. 66:373.

351. Neergaard, K.v., & Wirz, K. 1927. Uber eine Methode zur Messung der Lungenelastizitat am lebenden Menschen, insbesondere beim Emphysem. Ztschr.f.klin.Med. 105:35.

352. Nemery, B., Nullens, W., Veriter, C., Brasseur, L., & Frans, A. 1983. Effects of gas density on pulmonary gas exchange of normal man at rest and during exercise. Pflugers Arch. 397:57–61.

353. Niewoehner, D.E., & Kleinerman, J. 1974. Morphologic basis of pulmonary resistance in the human lung and effects of aging. J.Appl.Physiol. 36(4):412–418.

354. Nisell, O. 1960. The respiratory work and pressure during exercise, and their relation to dyspnea. Acta Med.Scand. 166(2):113–119.

355. Nisell, O., & Ehrner, L. 1958. The resistance to breathing determined from time-marked respiratory pressure volume loops. Acta Med.Scand. 156:427–436.

356. Olafsson, S., & Hyatt, R.E. 1969. Ventilatory mechanics and expiratory flow limitation during exercise in normal subjects. J.Clin.Invest. 48:564–573.

357. Olson, D.E., Dart, G.A., & Filley, G.F. 1970. Pressure drop and fluid flow regime of air inspired into the human lung. J.Appl.Physiol. 28:482–494.

358. Opie, L.H., Spalding, J.M.K., & Stott, F.D. 1959. Mechanical properties of the chest during intermittent positive-pressure respiration. Lancet. 1:545–550.

359. Orlov, V.N., Fomin, I.O., Radzevich, A.E., & Leskin, G.S. 1985. Effect of positive pressure breathing on hemodynamics of patients with borderline hypertension during water immersion. Kosm.Biol.Aviakosm.Med. 19:39–41.

360. Ostrove, S.M., & Vaccaro, P. 1982. Effect of immersion on RV in young women: implications for measurement of body density. Int.J.Sports Med. 3:220–223.

361. Otis, A.B. 1964. The work of breathing. In: Fenn, W.O., & Rahn, H. Handbook of Physiology. Respiration I. American Physiological Society. Washington, D.C. Pp.463–476.

362. Otis, A.B., Fenn, W.O., & Rahn, H. 1950. Mechanics of breathing in man. J.Appl.Physiol. 2:592–607.

363. Otis, A.B., McKerrow, C.B., Bartlett, R.A., Mead, J., McIlroy, M.D., Selverstone, N.J., & Radford, E.P.Jr. 1956. Mechanical factors in distribution of pulmonary ventilation. J.Appl.Physiol. 8:427–443.

364. Owen, P.R. 1969. Turbulent flow and particle deposition in the trachea. In: Wolstenholme, G.E.W., & Knight, J. Circulatory and Respiratory Mass Transport. Ciba Foundation Symposium. Little, Brown & Company. Boston. Pp.236–252.

365. Paley, H.W., Leonard, J.J., Eggers, G.W.N.Jr., DeGroot, W.J., & Warren, J.V. 1960. Hemodynamic effects of full phase negative pressure breathing. Circulation. 22:794.

366. Pappenheimer, J.R. 1950. Standardization of definitions and symbols in respiratory physiology. Fed.Proc. 9:602–605.

367. Pardy, R.L., Hussain, S.N.A., & Macklem, P.T. 1984. The ventilatory pump in exercise. Clin.Chest Med. 5(1):35–49.

368. Paton, W.D.M., & Sand, A. 1947. The optimum intrapulmonary pressure in underwater respiration. J.Physiol. 106:119–138.

369. Pedersen, O.F., Thiessen, B., Naeraa, N., Lyager, S., & Hilberg, C. 1984. Factors determining residual volume in normal and asthmatic subjects. Eur.J.Respir.Dis. 65:99–105.

370. Pengelly, L.D., Alderson, A.M., & Milic-Emili, J. 1971. Mechanics of the diaphragm. J.Appl.Physiol. 30:797–805.

371. Permutt, S., & Martini, H.B. 1960. Static pressure–volume characteristics of lungs in normal males. J.Appl.Physiol. 15(5):819–825.

372. Peterson, R.E., & Wright, W.B. 1976. Pulmonary mechanical functions in man breathing dense gas mixtures at high ambient pressures - predictive studies III. In: Lambertsen, C.J. Underwater Physiology V. Proceedings of Fifth Symposium on Underwater Physiology. Undersea Medical Society. Bethesda, Maryland. Pp.67–77.

373. Pierce, J.A., & Ebert, R.V. 1965. Fibrous network of the lung and its change with age. Thorax 20:469–476.

374. Pirnay, F., Marechal, R.,Dujardin, M., Lamay, M., Deroanne, R., & Petit, J.M. 1973. Exercise during hyperocia and hyperbaric oxygenation. Int.Z.Angew.Physiol.Einschl.Arbietsphysiol. 31:259–268.

375. Potter, W.A., Olafsson, S., & Hyatt, R.E. 1971. Ventilatory mechanics and expiratory flow limitation during exercise in patients with obstructive lung disease. J.Clin.Invest. 50:910–919.

376. Prefaut, Ch., Du Bois, F., Roussos, Ch., Amaral-Marques, R., Macklem, P.T., & Ruff, F. 1979. Influence of immersion to the neck in water on airway closure and distribution of perfusion in man. Respir.Physiol. 37:313–323.

377. Prefaut, Ch., Lupi-H, E., & Anthonisen, N.R. 1976. Human lung mechanics during water immersion. J.Appl.Physiol. 40(3):320–323.

378. Prefaut, Ch., Ramonatxo, M., Boyer, R., & Chardon, G. 1978. Human gas exchange during water immersion. Respir.Physiol. 34:307–317.

379. Pride, N.B. 1971. The assessment of airflow obstruction. Role of measurements of airways resistance and of tests of forced expiration. Br.J.Dis.Chest 65:135–169.

380. Proctor, H.J., Fry, J., & Harry, R. 1974. The metabolic cost of increased respiratory work. South.Med.J. 67(9):1047–1051.

381. Pump, K.K. 1966. The circulation in the peripheral parts of the human lung. Diseases of the Chest. 49:119–129.

382. Radford, E.P.Jr. 1964. Static mechanical properties of mammalian lungs. In: Fenn, W.O., & Rahn, H. Handbook of Physiology: Respiration 1. American Physiological Society. Washington, D.C. Pp.429–449.

383. Rahn, H., & Bahnson, H.T. 1953. Effect of unilateral hypoxia on gas exchange and calculated pulmonary blood flow in each lung. J.Appl.Physiol. 6:105–112.

384. Rahn, H., Otis, A.B., Chadwick, L.E., & Fenn, W.O. 1946. The pressure–volume diagram of the thorax and lung. Am.J.Physiol. 146(6):161–178.

385. Reid, M.B., Lanzett, R.B., Feldman, H.A., & Mead, J. 1985. Reflex compensation of spontaneous breathing when immersion changes diaphragm length. J.Appl.Physiol. 58(4):1136–1142.

386. Remmers, J.E. 1976. Analysis of ventilatory responses. Chest 70:134–137.

387. Rennie, D.W., DiPrampero, P., & Cerretelli, P. 1971. Effects of water immersion on cardiac output, heart rate, and stroke volume of man at rest and during exercise. Med.Sport 24:223-228

388. Risch, W.D., Koubenec, H.J., Beckmann, U., Lange, S., & Gauer, O.H. 1978. The effect of graded immersion on heart volume, central venous pressure, pulmonary blood distribution, and heart rate in man. Pflugers Arch. 374:115-118.

389. Risch, W.D., Koubenac, H.J., Gauer, O.H., & Lange, S. 1978. Time course of cardiac distension with rapid immersion in a thermo-neutral bath. Pflugers Arch. 374:119-120.

390. Robertson, C.H.Jr., Engle, C.M., & Bradley, M.E. 1978. Lung volumes in man immersed to the neck: dilution and plethysmographic techniques. J.Appl.Physiol. 44(5):679-682.

391. Rohrer, F. 1915. Der Strömungswiderstand in den menschlichen Atemwegen und der Einfluss der unregelmässigen Verweigung des Bronchialsystems auf den Atmungsverlauf verschiedenen Lungenbezirken. Arch.Ges.Physiol. 162:225-299.

392. Rohrer, F. 1916. Der Zusammenhang der Atemkräfte und ihre Abhängigkeit vom Dehnungszustand der Atmungsorgane. Pflugers Arch. 165:419.

393. Rohrer, F. 1925. Physiologie der Atembewegung. Handbuch der normalen und path.Physiol. 2:70.

394. Romanoff, M. 1910-1911. Experimente über Beziehungen zwischen Atmung und Kreislauf. Arch.f.exper.Path. 64:183.

395. Roussos, C.S., Fixley, M., Gross, D., & Macklem, P.T. 1979. Fatigue of inspiratory muscles and their synergic behaviour. J.Appl.Physiol. 46:897-904.

396. Roussos, C.S., & Macklem, P.T. 1977. Diaphragmatic fatigue in man. J.Appl.Physiol. 43(2):189-197.

397. Roussos, C., & Macklem, P.T. 1982. The respiratory muscles. N.Eng.J.Med. 307(13):786-797.

398. Saltin, B. 1973. Metabolic fundamentals in exercise. Med.Sci.Sports. 5:137-146.

399. Saltin, B., & Åstrand, P.-O. 1967. Maximal oxygen uptake in athletes. J.Appl.Physiol. 23:353-358.

400. Saltin, B., & Rowell, L.V. 1980. Functional adaptation to physical activity and inactivity. Fed.Proc. 39:1506-1513.

401. Salzano, J.V., Camporesi, E.M., Stolp, B.W., & Moon, R.E. 1984. Physiological responses to exercise at 47 and 66 ATA. J.Appl.Physiol. 57(4):1055-1068.

402. Salzano, J.V., Stolp, B.W., Moon, R.E., & Camporesi, E.M. 1981. Exercise at 47 and 66 ATA. In: Bachrach, A.J., & Matzen, M.M. Underwater Physiology VII. Proceedings of the Seventh Symposium on Underwater Physiology. Undersea Medical Society. Bethesda, Maryland. Pp.181-196.

403. Sawka, M.N., Weber, H., & Knowlton, R.G. 1978. The effect of total body submersion on residual lung volume and body density measurements in man. Ergonomics 21(2):89-94.

404. Scharf, S., Bye, P., Pardy, R., & Macklem, P.T. 1984. Dyspnea, fatigue, and second wind. Am.Rev.Respir.Dis. 129:Suppl S88-S89.

405. Schilder, D.P., Roberts, A., & Fry, D.L. 1963. Effect of gas density and viscosity on the maximal expiration flow-volume relationships. J.Clin.Invest. 42:1705-1713.

406. Severinghaus, J.S. 1974. In: Nahas, G. & Schaefer, K.E. Carbon Dioxide and Metabolic Regulations. Springer. New York. Pp.138-143.

407. Shapiro, W., Johnston, C.E., Dameron, R.A.Jr., & Patterson, J.L.Jr. 1964. Maximum ventilatory performance and its limiting factors. J.Appl.Physiol. 19(2):199–203.

408. Sharp, J.T., Griffith, G.T., Bunnell, I.L., & Greene, D.G. 1958. Ventilatory mechanics in pulmonary edema in man. J.Clin.Invest. 37:111–117.

409. Sharp, J.T., Henry, J.P., Sweany, S.K., Meadows, W.R., & Pietras, R.J. 1964a. Total respiratory inertance and its gas and tissue components in normal and obese men. J.Clin.Invest. 43(3):503–509.

410. Sharp, J.T., Henry, J.P., Sweany, S.K., Meadows, W.R., & Pietras, R.J. 1964b. The total work of breathing in normal and obese men. J.Clin.Invest. 43(4):728–739.

411. Shephard, R.J. 1966. The oxygen cost of breathing during vigorous exercise. Q.J.Exp.Physiol. 51:336–350.

412. Shephard, R.J. 1967. The maximum sustained voluntary ventilation in exercise. Clin.Sci. 32:167–176.

413. Sherman, D., Eilender, E., Shefer, A., & Kerem, D. 1980. Ventilatory and occlusion-pressure responses to hypercapnia in divers and non-divers. Undersea Biomed.Res. 7:61–74.

414. Shilling, C.W., Hawkins, J.A., & Hansen, R.A. 1936. The influence of increased barometric pressure on the pulse rate and arterial blood pressure. U.S.Med.Bull. 34(1):39–47.

415. Silverman, L., Lee, G., Plotkin, T., Sawyers, L.A., & Yancey, A.R. 1951. Air flow measurements on human subjects with and without respiratory resistance at several work rates. Arch.Ind.Hyg. 3:461–478.

416. Sladen, A., Laver, M.B., & Pontoppidan, H. 1968. Pulmonary omplications and water retention in prolonged mechanical ventilation. N.Eng.J.Med. 279:448–453.

417. Smith, P.K., Tyson, G.S., Hammon, J.W., Olsen, C.O., Hopkins, R.A., Maier, G.W., Sabiston, D.C., & Rankin, J.S. 1982. Cardiovascular effects of ventilation with positive expiratory airway pressure. Ann.Surg. 195:121–130.

418. Spaur, W.H., Raymond, L.W., Knott, M.M., Crothers, J.C., Braithwaite, W.R., Thalmann, E.D., & Uddin, D.F. 1977. Dyspnea in divers at 49.5 ATA: mechanical, not chemical in origin. Undersea Biomed.Res. 4(2):183–198.

419. Staats, B.A., Wilson, T.A., Lai-Fook, S.J., Rodarte, J.R., & Hyatt, R.E. 1980. Viscosity and density dependence during maximal flow in man. J.Appl.Physiol. 48(2):313–319.

420. Staub, N.C. 1963. The interdependence of pulmonary structure and function. Anesthesiology 24:831–854.

421. Stead, W.W., Fry, D.L., & Ebert, R.V. 1952. The elastic properties of the lung in normal men and in patients with chronic pulmonary emphysema. J.Lab.Clin.Med. 40:674–681.

422. Stenger, S.H., Frayser, R., & Ross, J.C. 1965. Alterations in pulmonary diffusing capacity and pulmonary capillary blood volume with negative pressure breathing. J.Clin.Invest. 44(10):1623–1630.

423. Stenberg, J., Ekblom, B., & Messin, R. 1966. Hemodynamic response to work at simulated altitude, 4000m. J.Appl.Physiol. 21:1589–1594.

424. Sterk, W. 1970. Diver and underwater breathing apparatus. Nederl.Milit.Geneesk.T. 23:322–356.

425. Sterk, W. 1973. Respiratory mechanics of diver and diving apparatus. PhD Thesis. University of Utrecht (Holland). Drukkerij Elinkwijk, Utrecht.

426. Stigler, R. 1911. Die Kraft unserer Inspirationmuskulatur. Pflugers Arch. 139:234–254.

427. Stolp, B.W., Moon, R.E., Camporesi, E.M., & Salzano, J.V. 1981. Frequency dependence of MVV during deep diving. Undersea Biomed.Res. Suppl(Abstract). 8(1):24.

428. Strohl, K.P., Mead, J., Banzett, R.B., Loring, S.H., & Kosch, P.C. 1981. Regional differences in abdominal muscle activity during various maneuvers in humans. J.Appl.Physiol. 51(6):1471-1476.

429. Stubbing, D.G., Pengelly, L.D., Morse, J.L.C., & Jones, N.L. 1980a. Pulmonary mechanics during exercise in normal males. J.Appl.Physiol. 49(3):506-510.

430. Stubbing, D.G., Pengelly, L.D., Morse, J.L.C., & Jones, N.L. 1980b. Pulmonary mechanics during exercise in subjects with chronic airflow obstruction. J.Appl.Physiol. 49(3):511-515.

431. Sutherland, P.W., Katsura, T., & Milic-Emili, J. 1968. Previous volume history of the lung and regional distribution of gas. J.Appl.Physiol. 25(5):566-574.

432. Taunton, J.E., Banister, E.W., Patrick, T.R., Oforsagd, P., & Duncan, W.R. 1970. Physical work capacity in hyperbaric environments and conditions of hyperoxia. J.Appl.Physiol. 28(4):421-427.

433. Tenney, S.M. 1959. Fluid volume redistribution and thoracic volume changes during recumbency. J.Appl.Physiol. 14:129-132.

434. Thalmann, E.D., Sponholtz, D.K., & Lundgren, C.E.G. 1978. Chamber-based system for physiological monitoring of submerged exercising subjects. Undersea Biomed.Res. 5:293-300.

435. Thalmann, E.D., Sponholtz, D.K., & Lundgren, C.E.G. 1979. Effects of immersion and static lung loading on submerged exercise at depth. Undersea Biomed.Res. 6(3):259-290.

436. Thompson, L.J., & McCally, M. 1966. Role of transpharyngeal pressure gradients in determining intrapulmonary pressure during immersion. Aerospace Med. 37:304.

437. Thompson, L.J., & McCally, M. 1967. Role of transpharyngeal pressure gradients in determining intrapulmonary pressure during immersion. Aerospace Med. 38:931-935.

438. Timson, B.F., & Coffman, J.L. 1984. Body composition by hydrostatic weighing at total lung capacity and residual volume. Med.Sci.Sports Exerc. 16(4):411-414.

439. Ting, E.Y., Hong, S.K., & Rahn, H. 1960a. Lung volumes, lung compliance and airway resistance during negative-pressure breathing. J.Appl.Physiol. 15(4):554-556.

440. Ting, E.Y., Hong, S.K., & Rahn, H. 1960b. Cardiovascular responses of man during negative-pressure breathing. J.Appl.Physiol. 15(4):557-560.

441. Toung, T.J.K., Saharia, P., Mitzner, W.A., Permutt, S., & Cameron, J.L. 1978. The beneficial and harmful effects of positive end expiratory pressure. Surg.Gynecol.Obstet. 147:518-524.

442. Turner, J.M., Mead, J., & Wohl, M.E. 1968. Elasticity of human lungs in relation to age. J.Appl.Physiol. 25:664-671.

443. Uhl, R.R., Van Dyke, C., Cook, R.B., Horst, R.A., & Merz, J.M. 1972. Effects of externally imposed mechanical resistance on breathing dense gas at exercise: mechanics of breathing. Aerospace Med. 43(8):836-841.

444. Vail, E.G. 1973. Pulmonary insufficiency with airway flutter, closure, and collapse. Aerospace Med. 44(6):649-662.

445. Van Liew, H.D. 1982. A review of the consequences of breathing dense gas. Proceedings of the First Annual Canadian Ocean Technology Congress, Toronto, Ontario. Pp.133-147.

446. Van Liew, H.D. 1983. Mechanical and physical factors in lung function during work in dense environments. Undersea Biomed.Res. 10(3):255-264.

447. Varene, P., Vieillefond, H., Lemaire, C., & Saumon, G. 1974. Expiratory flow volume curves and ventilatory limitation of muscular exercise at depth. Aerospace Med. 45(2):161-166.

448. Velasquez, T., & Farhi, L.E. 1964. Effect of negative-pressure breathing on lung mechanics and venous admixture. J.Appl.Physiol. 19(4):665-671.

449. Venus, B., & Jacobs, H.K. 1984. Alterations in regional myocardial blood flows during different levels of positive end-expiratory pressure. Crit.Care Med. 12:96-101.

450. Von Basch, S. 1887. Ueber eine Funktion des Capillardruckes in den Lungenalveolen. Wiener Med.Blatter 15:465.

451. Vorosmarti, J.Jr. 1979. Influence of increased gas density and external resistance on maximum expiratory flow. Undersea Biomed.Res. 6(4):339-346.

452. Vorosmarti, J., Bradley, M.E., & Anthonisen, N.R. 1975. The effects of increased gas density on pulmonary mechanics. Undersea Biomed.Res. 2(1):1-10.

453. Vorosmarti, J.Jr., & Lanphier, E.H. 1976. Effects of external resistance on maximum expiratory flow at increased gas density. In: Lambertsen, C.J. Underwater Physiology V. Proceedings of the Fifth Symposium on Underwater Physiology. FASEB, Bethesda, Maryland. Pp.101-107.

454. Weibel, E.R. 1963. Morphometry of the Human Lung. Springer. Berlin.

455. Weltman, A., & Katch, V. 1981. Comparison of hydrostatic weighing at residual volume and total lung capacity. Med.Sci.Sports Exerc. 13(3):210-213.

456. West, B., Dollery, C.T., & Heard, B.E. 1965. Increased pulmonary vascular resistance in the dependent zone of the isolated dog lung caused by perivascular edema. Circ.Res. 17:191-206.

457. West, J.B. 1965. Ventilation/Blood Flow and Gas Exchange. Blackwell. Oxford.

458. Whipp, B.J. 1981. Mechanics of ventilatory control during exercise. In: Lanphier, E.H. Respiratory Control and Loss of Consciousness in Diving. Undersea Medical Society. Bethesda, Maryland.

459. Wilhelmsen, L., & Varnauskas, E. 1967. Effects of acute plasma expansion on the mechanics of breathing. Clin.Sci. 33:29-38.

460. Wirz, K. 1923. Das Verhalten des Druckes im Pleuraraum bei der Atmung und die Ursachen seiner Veranderlichkeit. Arch.Ges.Physiol. 199:1-56.

461. Wood, L.D.H., & Bryan, A.C. 1971. Mechanical limitations of exercise ventilation at increased ambient pressure. In: Lambertsen, C.J. Underwater Physiology. Proceedings of the Fourth Symposium on Underwater Physiology. Academic Press. New York. Pp.307-316.

462. Wood, T.E., McLeod, P., Anthonisen, N.R., & Macklem, P.T. 1971. Mechanics of breathing in mitral stenosis. Am.Rev.Respir.Dis. 104:52-60.

463. Wood, W.B. 1963. Ventilatory dynamics under hyperbaric states. In: Lambertsen, C.J., & Greenbaum, L.J. Underwater Physiology II. Proceedings of the Second Symposium on Underwater Physiology. National Academy of Science National Research Council. Publ.1181. Washington, D.C. Pp.108-123.

464. Wyndham, C.H., Strydom, N.B., van Rensburg, A.J., & Rogers, G.G. 1970. Effects on maximal oxygen intake of acute changes in altitude in a deep mine. J.Appl.Physiol. 29:552-555.

465. Yamashiro, S.M., Daubenspeck, J.A., Lauritsen, T.N., & Grodins, F.S. 1975. Total work rate of breathing optimization in CO2 inhalatio. and exercise. J.Appl.Physiol. 38(4):702-709.

466. Zidulka, A., Despas, P.J., Milic-Emili, J., & Anthonisen, N.R. 1973. Pulmonary function with acute loss of excess lung water by hemodialysis in patients with chronic uremia. Am.J.Med. 55:134-141.

467. Zocche, G.P., Fritts, H.W.Jr., & Cournand, A. 1960. Fraction of maximum breathing capacity available for prolonged hyperventilation. J.Appl.Physiol. 15:1073-1074.

468. Zocchi, L, Luijendijk, S.C.M., Zin, W.A., Rossi, A., & Milic-Emili, J. 1984. Effect of resistive loading on inspiratory work output in anesthetized cats. J.Appl.Physiol. 57(3):839-849.