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MOSBERG, ARNOLD THEODORE
AN IN VIVO STUDY OF THE VELOCITY PATTERNS IN
THE CANINE UPPER AIRWAYS.
THE OHIO STATE UNIVERSITY, PH.D., 1978

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AN IN VIVO STUDY OF THE
VELOCITY PATTERNS IN THE CANINE UPPER AIRWAYS

DISSERTATION

Presented in Partial Fulfillment of the Requirements for
the Degree Doctor of Philosophy in the Graduate
School of The Ohio State University

by
Arnold Theodore Mosberg, B.A.A.E., M.S.

* * * * *

The Ohio State University
1978

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ACKNOWLEDGMENTS

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# TABLE OF CONTENTS

<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>ACKNOWLEDGMENTS</td>
<td>ii</td>
</tr>
<tr>
<td>CURRICULUM VITA</td>
<td>iv</td>
</tr>
<tr>
<td>LIST OF TABLES</td>
<td>viii</td>
</tr>
<tr>
<td>LIST OF FIGURES</td>
<td>ix</td>
</tr>
<tr>
<td>NOMENCLATURE</td>
<td>xvii</td>
</tr>
<tr>
<td>CHAPTER I - INTRODUCTION</td>
<td></td>
</tr>
<tr>
<td>1.1 Background</td>
<td>1</td>
</tr>
<tr>
<td>1.2 Review of Previous Work</td>
<td>7</td>
</tr>
<tr>
<td>1.3 Objectives of Present Work</td>
<td>8</td>
</tr>
<tr>
<td>CHAPTER II - ANATOMY AND PHYSIOLOGY OF THE LUNG</td>
<td></td>
</tr>
<tr>
<td>2.1 Physiology of Respiration</td>
<td>11</td>
</tr>
<tr>
<td>2.2 Anatomy of the Canine Lung</td>
<td>23</td>
</tr>
<tr>
<td>CHAPTER III - TECHNICAL ASPECTS OF AIRWAY MEASUREMENT</td>
<td></td>
</tr>
<tr>
<td>3.1 Fundamental Parameters</td>
<td>32</td>
</tr>
<tr>
<td>3.2 Anemometry</td>
<td>32</td>
</tr>
<tr>
<td>3.3 Pneumotachometry</td>
<td>43</td>
</tr>
<tr>
<td>3.4 Acoustic Recording</td>
<td>47</td>
</tr>
<tr>
<td>3.5 Bronchoscopy</td>
<td>49</td>
</tr>
<tr>
<td>3.6 Data Storage</td>
<td>50</td>
</tr>
<tr>
<td>CHAPTER IV - THE EXPERIMENT</td>
<td></td>
</tr>
<tr>
<td>4.1 Introduction</td>
<td>53</td>
</tr>
<tr>
<td>4.2 Animal Model</td>
<td>53</td>
</tr>
<tr>
<td>4.3 Animal Preparation</td>
<td>55</td>
</tr>
<tr>
<td>4.4 Surgical Procedure - Trachea</td>
<td>56</td>
</tr>
<tr>
<td>4.5 Surgical Procedure - Upper Airways</td>
<td>59</td>
</tr>
<tr>
<td>4.6 Fluid Mechanics Parameters</td>
<td>61</td>
</tr>
<tr>
<td>4.7 Frequency Analysis</td>
<td>72</td>
</tr>
</tbody>
</table>
# TABLE OF CONTENTS (con't)

<table>
<thead>
<tr>
<th>CHAPTER V - RESULTS</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>5.1 Introduction.</td>
<td>77</td>
</tr>
<tr>
<td>5.2 Tracheal Velocity Profiles.</td>
<td>78</td>
</tr>
<tr>
<td>5.3 Velocity Spectra-Trachea.</td>
<td>81</td>
</tr>
<tr>
<td>5.4 Upper Airways.</td>
<td>82</td>
</tr>
<tr>
<td>5.5 Lung Sounds</td>
<td>82</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>CHAPTER VI - DISCUSSION OF RESULTS</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>6.1 Introduction.</td>
<td>158</td>
</tr>
<tr>
<td>6.2 Tracheal Flow - Inspiration</td>
<td>159</td>
</tr>
<tr>
<td>6.3 Tracheal Flow - Expiration</td>
<td>169</td>
</tr>
<tr>
<td>6.4 Upper Airways Flow</td>
<td>175</td>
</tr>
<tr>
<td>6.5 Tracheal Sounds</td>
<td>185</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>CHAPTER VII - CONCLUSION AND REMARKS</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>7.1 Introduction.</td>
<td>189</td>
</tr>
<tr>
<td>7.2 Conclusions</td>
<td>190</td>
</tr>
<tr>
<td>7.3 Limitations of Study</td>
<td>192</td>
</tr>
<tr>
<td>7.4 Future Work</td>
<td>194</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>BIBLIOGRAPHY</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>195</td>
</tr>
</tbody>
</table>
**LIST OF TABLES**

<table>
<thead>
<tr>
<th>Table</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.1</td>
<td>Respiratory work at rest, during exercise, and during maximal voluntary ventilation</td>
<td>24</td>
</tr>
<tr>
<td>5.1</td>
<td>Summary of Experiments</td>
<td>84</td>
</tr>
<tr>
<td>6.1</td>
<td>A tabulation of retrograde experiments describing peak velocities, Reynolds numbers and turbulence intensities at each generation.</td>
<td>179</td>
</tr>
</tbody>
</table>
## LIST OF FIGURES

<table>
<thead>
<tr>
<th>Figure</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.1 Volume flow and transpulmonary pressure as a function of time</td>
<td>15</td>
</tr>
<tr>
<td>2.2 Subdivisions of the lung volume</td>
<td>16</td>
</tr>
<tr>
<td>(Reprinted with permission of Physical Therapy, 48:478-494, 1968)</td>
<td></td>
</tr>
<tr>
<td>2.3 A simple mechanical model of the lung</td>
<td>20</td>
</tr>
<tr>
<td>(Reprinted with permission of Physical Therapy, 48:478-494, 1968)</td>
<td></td>
</tr>
<tr>
<td>2.4 Volume, flow and pressure waveforms from a simple lung model</td>
<td>21</td>
</tr>
<tr>
<td>(Reprinted with permission of Physical Therapy, 48:478-494, 1968)</td>
<td></td>
</tr>
<tr>
<td>2.5 Dorsal view of the teased lungs of the dog</td>
<td>31</td>
</tr>
<tr>
<td>3.1 Basic schematic of a constant temperature anemometer system</td>
<td>36</td>
</tr>
<tr>
<td>3.2 Temperature-flow characteristics of a respiratory gas heat exchanger</td>
<td>42</td>
</tr>
<tr>
<td>3.3 Hot-wire anemometer probe advancing mechanism</td>
<td>44</td>
</tr>
<tr>
<td>3.4 Photography by bronchoscopy documenting the position of the retrograde probe in the airways</td>
<td>51</td>
</tr>
<tr>
<td>4.1 Schematic representation of retrograde catheterization technique</td>
<td>62</td>
</tr>
<tr>
<td>4.2 Characteristic velocity waveform with the six time measurement points identified</td>
<td>69</td>
</tr>
<tr>
<td>Figure</td>
<td>Page</td>
</tr>
<tr>
<td>--------</td>
<td>------</td>
</tr>
<tr>
<td>4.3 Normalized velocity profiles in canine trachea at 25% of inspiration illustrating the data presentation format</td>
<td>70</td>
</tr>
<tr>
<td>4.4 Time-dependent frequency of real time waveform</td>
<td>74</td>
</tr>
<tr>
<td>5.1 Normalized velocity profiles measured in canine trachea at 25% of inspiration, exp. 12, U=100 cm/sec, Rn=1111</td>
<td>87</td>
</tr>
<tr>
<td>5.2 Normalized velocity profiles measured in canine trachea at 50% of inspiration, exp. 12, U=200 cm/sec, Rn=2222</td>
<td>88</td>
</tr>
<tr>
<td>5.3 Normalized velocity profiles measured in canine trachea at 75% of inspiration, exp. 12, U=100 cm/sec, Rn=1111</td>
<td>89</td>
</tr>
<tr>
<td>5.4 Normalized velocity profiles measured in canine trachea at 25% of expiration, exp. 12, U=512 cm/sec, Rn=5688</td>
<td>90</td>
</tr>
<tr>
<td>5.5 Normalized velocity profiles measured in canine trachea at 50% of expiration, exp. 12, U=200 cm/sec, Rn=2222</td>
<td>91</td>
</tr>
<tr>
<td>5.6 Normalized velocity profiles measured in canine trachea at 75% of expiration, exp. 12, U=70 cm/sec, Rn=777</td>
<td>92</td>
</tr>
<tr>
<td>5.7 Normalized velocity profiles measured in canine trachea at 25% of inspiration, exp. 13, U=280 cm/sec, Rn=2379</td>
<td>93</td>
</tr>
<tr>
<td>5.8 Normalized velocity profiles measured in canine trachea at 50% of inspiration, exp. 13, U=240 cm/sec, Rn=2039</td>
<td>94</td>
</tr>
<tr>
<td>5.9 Normalized velocity profiles measured in canine trachea at 75% of inspiration, exp. 13, U=172 cm/sec, Rn=1461</td>
<td>95</td>
</tr>
</tbody>
</table>
5.10 Normalized velocity profiles measured in canine trachea at 25% of expiration, exp. 13, U=400 cm/sec, Rn=3398. . . . . . . . . . . 96
5.11 Normalized velocity profiles measured in canine trachea at 50% of expiration, exp. 13, u=150 cm/sec, Rn=1275. . . . . . . . . . . 97
5.12 Normalized velocity profiles measured in canine trachea at 75% of expiration, exp. 13, U=50 cm/sec, Rn=424. . . . . . . . . . . 98
5.13 Normalized velocity profiles measured in canine trachea at 25% of inspiration, exp. 17, U=190 cm/sec, Rn=2359. . . . . . . . . . . 99
5.14 Normalized velocity profiles measured in canine trachea at 50% of inspiration, exp. 17, U=250 cm/sec, Rn=3104. . . . . . . . . . . 100
5.15 Normalized velocity profiles measured in canine trachea at 75% of inspiration, exp. 17, U=173 cm/sec, Rn=2148. . . . . . . . . . . 101
5.16 Normalized velocity profiles measured in canine trachea at 25% of expiration, exp. 17, U=490 cm/sec, Rn=6084. . . . . . . . . . . 102
5.17 Normalized velocity profiles measured in canine trachea at 50% of expiration, exp. 17, U=222 cm/sec, Rn=2756. . . . . . . . . . . 103
5.18 Normalized velocity profiles measured in canine trachea at 75% of expiration, exp. 17, U=142 cm/sec, Rn=1763. . . . . . . . . . . 104
5.19 Normalized velocity profiles measured in canine trachea at 25% of inspiration, exp. 18, U=217 cm/sec, Rn=2478. . . . . . . . . . . 105
5.20 Normalized velocity profiles measured in canine trachea at 50% of inspiration, exp. 18, U=207 cm/sec, Rn=4213. . . . . . . . . . . 106
<table>
<thead>
<tr>
<th>Figure</th>
<th>叙述</th>
</tr>
</thead>
<tbody>
<tr>
<td>5.21</td>
<td>Normalized velocity profiles measured in canine trachea at 75% of inspiration, exp. 18, U=153 cm/sec, Rn=2099.</td>
</tr>
<tr>
<td>5.22</td>
<td>Normalized velocity profiles measured in canine trachea at 25% of expiration, exp. 18, U=561 cm/sec, Rn=7699.</td>
</tr>
<tr>
<td>5.23</td>
<td>Normalized velocity profiles measured in canine trachea at 50% of expiration, exp. 18, U=250 cm/sec, Rn=3431.</td>
</tr>
<tr>
<td>5.24</td>
<td>Normalized velocity profiles measured in canine trachea at 75% of expiration, exp. 18, U=70 cm/sec, Rn=966.</td>
</tr>
<tr>
<td>5.25</td>
<td>Normalized velocity profiles measured in canine trachea at 25% of inspiration, exp. 19, U=200 cm/sec, Rn=2091.</td>
</tr>
<tr>
<td>5.26</td>
<td>Normalized velocity profiles measured in canine trachea at 50% of inspiration, exp. 19, U=270 cm/sec, Rn=2823.</td>
</tr>
<tr>
<td>5.27</td>
<td>Normalized velocity profiles measured in canine trachea at 75% of inspiration, exp. 19, U=171 cm/sec, Rn=1780.</td>
</tr>
<tr>
<td>5.28</td>
<td>Normalized velocity profiles measured in canine trachea at 25% of expiration, exp. 19, U=400 cm/sec, Rn=4183.</td>
</tr>
<tr>
<td>5.29</td>
<td>Normalized velocity profiles measured in canine trachea at 50% of expiration, exp. 19, U=304 cm/sec, Rn=3179.</td>
</tr>
<tr>
<td>5.30</td>
<td>Normalized velocity profiles measured in canine trachea at 75% of expiration, exp. 19, U=120 cm/sec, Rn=1274.</td>
</tr>
<tr>
<td>5.31</td>
<td>Normalized velocity profiles measured in canine trachea at 25% of inspiration, exp. 28, U=68 cm/sec, Rn=535.</td>
</tr>
<tr>
<td>5.32</td>
<td>Normalized velocity profiles measured in canine trachea at 50% of inspiration, exp. 28, U=181 cm/sec, Rn=1735.</td>
</tr>
</tbody>
</table>
5.33 Normalized velocity profiles measured in canine trachea at 75% of inspiration, exp. 28, U=185 cm/sec, Rn=1445. ........................................ 119

5.34 Normalized velocity profiles measured in canine trachea at 25% of expiration, exp. 28, U=429 cm/sec, Rn=4424. ........................................ 120

5.35 Normalized velocity profiles measured in canine trachea at 50% of expiration, exp. 28, U=238 cm/sec, Rn=2454. ........................................ 121

5.36 Normalized velocity profiles measured in canine trachea at 75% of expiration, exp. 28, U=95 cm/sec, Rn=975. ........................................ 122

5.37 Time-dependent spectral display of velocity waveform measured at trachea centerline location 3 cm from the thyroid cartilage, exp. 12 ........................................ 123

5.38 Time-dependent spectral display of velocity waveform measured at trachea centerline location 7 cm from the thyroid cartilage, exp. 12 ........................................ 124

5.39 Time-dependent spectral display of velocity waveform measured at trachea centerline location 13 cm from the thyroid cartilage, exp. 12 ........................................ 125

5.40 Time-dependent spectral display of velocity waveform measured at trachea centerline location 2 cm from the thyroid cartilage, exp. 13 ........................................ 126

5.41 Time-dependent spectral display of velocity waveform measured at trachea centerline location 4 cm from the thyroid cartilage, exp. 13 ........................................ 127

5.42 Time-dependent spectral display of velocity waveform measured at trachea centerline location 6 cm from the thyroid cartilage, exp. 13 ........................................ 128
<table>
<thead>
<tr>
<th>Figure</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>5.43</td>
<td>129</td>
</tr>
<tr>
<td>5.44</td>
<td>130</td>
</tr>
<tr>
<td>5.45</td>
<td>131</td>
</tr>
<tr>
<td>5.46</td>
<td>132</td>
</tr>
<tr>
<td>5.47</td>
<td>133</td>
</tr>
<tr>
<td>5.48</td>
<td>134</td>
</tr>
<tr>
<td>5.49</td>
<td>135</td>
</tr>
<tr>
<td>5.50</td>
<td>136</td>
</tr>
<tr>
<td>5.51</td>
<td>137</td>
</tr>
</tbody>
</table>
5.52 Time-dependent spectral display of velocity waveform measured in the 0th generation by retrograde catheter in exp. 29. 138

5.53 Time-dependent spectral display of velocity waveform measured in the 1st generation by retrograde catheter in exp. 29. 139

5.54 Time-dependent spectral display of velocity waveform measured in the 2nd generation by retrograde catheter in exp. 29. 140

5.55 Time-dependent spectral display of velocity waveform measured in the 4th generation by retrograde catheter in exp. 29. 141

5.56 Composite of real time velocity waveforms recorded in the upper airways during exp. 29. 142

5.57 Time-dependent spectral display of velocity waveform measured in the 0th generation by retrograde catheter in exp. 30. 143

5.58 Time-dependent spectral display of velocity waveform measured in the 1st generation by retrograde catheter in exp. 30. 144

5.59 Time-dependent spectral display of velocity waveform measured in the 2nd generation by retrograde catheter in exp. 30. 145

5.60 Time-dependent spectral display of velocity waveform measured in the 3rd generation by retrograde catheter in exp. 30. 146

5.61 Composite of real time velocity waveforms recorded in the upper airways during exp. 30. 147

5.62 Time-dependent spectral display of sound waveforms recorded during exp. 12 at a location 2 cm from the thyroid cartilage. 148

5.63 Time-dependent spectral display of sound waveforms recorded during exp. 12 at a location 7 cm from the thyroid cartilage. 149
<table>
<thead>
<tr>
<th>Figure</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>5.64</td>
<td>Time-dependent spectral display of sound waveforms recorded during exp. 12 at a location 3 cm from the thyroid cartilage.</td>
</tr>
<tr>
<td>5.65</td>
<td>Time-dependent spectral display of sound waveforms recorded during exp. 13 at a location 2 cm from the thyroid cartilage.</td>
</tr>
<tr>
<td>5.66</td>
<td>Time-dependent spectral display of sound waveforms recorded during exp. 13 at a location 6 cm from the thyroid cartilage.</td>
</tr>
<tr>
<td>5.67</td>
<td>Time-dependent spectral display of sound waveforms recorded during exp. 13 at a location 10 cm from the thyroid cartilage.</td>
</tr>
<tr>
<td>5.68</td>
<td>Time-dependent spectral display of sound waveforms recorded during exp. 28 at a location 2 cm from the thyroid cartilage.</td>
</tr>
<tr>
<td>5.69</td>
<td>Time-dependent spectral display of sound waveforms recorded during exp. 28 at a location 7 cm from the thyroid cartilage.</td>
</tr>
<tr>
<td>5.70</td>
<td>Time-dependent spectral display of sound waveforms recorded during exp. 28 at a location 11 cm from the thyroid cartilage.</td>
</tr>
<tr>
<td>5.71</td>
<td>Time synchronized sound and flow signals illustrating the correlation between the onset of sound and turbulence in the expiratory flow pattern.</td>
</tr>
<tr>
<td>6.1</td>
<td>Decay of turbulence within the lung at subcritical Reynolds numbers during inspiration.</td>
</tr>
</tbody>
</table>
NOMENCLATURE

A - constant in King equation (9)
A' - constant in turbulence decay equation (26)
A" - constant in turbulence decay equation (26)
a - radius of a circular tube (10,16,17,21)
B - constant in King equation (9)
b - constant in heat transfer equation (10)
C - lung compliance (1,2,8)
D - diameter of an airway lumen (11)
F(n) - frequency distribution of kinetic energy associated with random fluctuation (19)
f - respiratory frequency (5,7,8,11)
L - characteristic length scale (12)
M - thermal time constant of hot-wire probe
N - exponent in King equation (9)
P - pressure in an airway (9,13,14,16)
Pe1 - elastic recoil pressure (1,2)
Pre - pressure related to resistance (5)
Q - heat removed per unit time from a hot-wire sensor (9)
q - root mean square of turbulence fluctuation (21,24)
R - resistance to breathing (5,6,8)
r - radial distance (13,21)
Rn - Reynolds number (12,25,26,27)
T - period of turbulence measurement (20)
t - time (13,14,23)
U - flow velocity (9,10,12,18,21)
U - mean flow velocity (18)
Vj - assessed solution to flow equation (15,16)
u' - fluctuating component of velocity (18,20)
Vn - normalized flow velocity
Vpn% - peak velocity measured during one time segment
NOMENCLATURE (con't)

V - voltage drop across hot-wire sensor (10)
Vt - tidal volume of breathing (1,2,5,6,8)
Wel - work of elastic recoil (4,7)
Wre - work of flow resistance (6,7)
Wt - total work of breathing (7)
y - non-dimensionalized radius (16)
z - axial distance (13)
a - Womersley parameter (11,17)
∆T - overheat temperature of anemometer (10)
ε - rate of energy dissipation per unit mass (21,23)
µ - coefficient of viscosity (12,16)
ν - kinematic viscosity (11,13,16,17,23)
ρ - density of fluid (12)
ω - angular frequency (14,16,17,25,26,27)
ζ - normalized axial distance (25,26,27)
CHAPTER I

INTRODUCTION

1.1 Background

Diseases of the pulmonary system contribute significantly to morbidity and mortality rates throughout the world (1). Chronic lung diseases such as emphysema, asthma, and bronchitis are debilitating to their victims, and acute attacks cause patient distress (2, 3). Silicosis and asbestosis can also lead to impaired lung function (1), primarily by over distention of the surrounding lung in response to local penetration of fibrous tissue. Certainly the ability diagnose and treat these diseases at an early stage of their development would be welcome, especially in cases where the disease is significantly more amenable to early treatment. It is in these areas of diagnosis and treatment that the author feels the understanding of fluid mechanics within the lung can play an extremely important role.
For example, the elucidation of airflow velocity patterns within the lung would provide the knowledge to describe particle deposition patterns and enhance our understanding of gas mixing within the pulmonary trunk. The understanding of resistance to flow would be within reach if velocity profiles were available, as well as an understanding of abnormal ambient effects such as those experienced by deep sea divers. Velocity profiles would also provide the foundation for the development of new non-invasive pulmonary function tests which would be able to provide differential diagnosis at an earlier time. Velocity profiles may also open the door to comprehension of sound patterns within the lung system, and subsequent diagnosis by auscultation.

The chronic lung diseases mentioned earlier are all characterized by impaired air movement. In asthma, an abnormal responsiveness of the air passages to allergens manifests a widespread narrowing of the smaller airways leading to a significant increase in work requirements for breathing. It is estimated that approximately one percent of the entire population has asthma.

Pulmonary emphysema is an anatomical change in the lungs which includes some destruction of air-sac walls and enlarged air spaces beyond the terminal
bronchioles. These anatomical changes lead to impaired lung function characterized by difficulty in expelling air from the lung, and poor distribution of inhaled gas. This difficulty in expelling air is caused by bronchiolar obstruction resulting from distended air sacs, inflammatory swelling of bronchial walls, and thick secretions. The problem is further aggravated by air continually being trapped in the lobules causing more alveolar enlargement and destruction.

Chronic bronchitis is the inflammation of the membrane that lines the bronchial tubes, causing blocking or narrowing. This disease is often thought to be the precursor to emphysema and, although not often fatal, has accounted for over 22 million work days lost in one year in England (1). This becomes extremely important in terms of manpower and money.

Although the above diseases are clearly different, no test now available can distinguish the functional differences in a simple manner. However, with an increased understanding of the fluid mechanics in conjunction with the same in other areas of basic lung research, it may be possible to isolate these diseases at an early stage of their development.
Olson et al. (5) and Pedley et al. (6) both investigated the distribution of airways resistance below the larynx using model studies and theoretical analysis. These studies yield significant information, but need to be compared with actual measurement before a full understanding is reached. When this distribution of resistance and compliance is understood, the differential diagnosis of these diseases may be possible through a determination of the anatomical impairment of lung function. Such techniques are beginning to appear, as in the case of the Helium isoflow procedures described by Hutcheon (7).

Besides pressure drop diagnostics, the relationship of fluid mechanics to gas mixing can also be employed as a diagnostic tool. Carbon monoxide, often used as a test for gas exchange in the lung, can also be employed to evaluate mixing. When CO is mixed with SF₆ or Helium, the rates of uptake of CO are greatly modified (8), leading to the conclusion that there is significant stratified inhomogeneity, or regional difference in gas concentrations. Velocity profiles would make such uptake information much more comprehensible. Such an understanding would be useful in demonstrating impairment of function. However, the complicated problems of gas mixing during
inspiration and expiration can be better understood when the fluid flow itself is understood.

Phonopneumography, the quantitative analysis of lung sounds, is likely to be enhanced by knowledge of velocity patterns in the lung. At present, confusion regarding terminology and the significance of adventitious and normal respiratory sounds continues to hamper physicians (9). Knowledge of the velocity waveforms may lead to a better understanding of the aerodynamic sound production mechanisms, thus providing the fundamental basis for the interpretation of bronchopulmonary sounds. With these steps taken, auscultation hopefully will be placed on a more objective basis and would become more precise and useful in the diagnosis of bronchopulmonary disease.

Many new avenues of treatment might be opened by the understanding of velocity profiles within the lung. Olson et al. (10) indicate that the deposition and transport of aerosol particles into the lung are greatly influenced by the convective patterns of inspired air in the upper and central airways. Thus, velocity patterns will be helpful in understanding deposition, depth of particle transport, and the distribution of particles to each parallel functioning unit of the lung.
Fluid mechanics plays an extremely important role in the delivery of medications by aerosol to patients with asthma, bronchitis, or emphysema. Knowledge of velocity profiles could very well influence the design of those aerosol agents to be most effective in reaching the target locations. Anesthesia gases are also in this category. Airborne pollution agents such as silicone dioxide play an important role in the development of lung impairment in many individuals. Non-particulate matter such as gases (sulfur dioxide) and liquid droplets may also damage the lung. Such irritants may destroy or retard the normal removal mechanisms thus creating a positive feedback loop leading to more deposition (11) and destruction.

Alterations in fluid mechanics for a patient in distress may be used to provide relief. These alterations would have to be based upon information gained from velocity pattern studies of a particular disease. An example of this type of treatment is positive expiratory pressure applied to patients with pulmonary emphysema. The positive pressure helps prevent the collapse of bronchioles. Intermittent positive pressure breathing is another example. However, many new methods which might involve controlling airways velocities or using different
transport gases may also be employed once an understanding of local velocity patterns is gained.

1.2 Review of Previous Work

Significant information on lung flow has been provided by studies of velocity patterns in models, even though almost all of these were steady state inspiratory flows. Perhaps the most significant of all of these efforts is Olson's work (12). The significance of this work stems primarily from the techniques employed in obtaining three dimensional velocity vectors at every point measured, thus fully describing both the magnitude and direction of the flow at every measurement point. Olson accomplished this by using a dual-wire pulsed heat probe that allowed rotation until a maximum change in resistance occurred in the sensor wire. This indicated that the sensor wire was directly downstream of the heated wire. The time between the transmitted pulse and received heat pulse characterized the local velocity.

Schroter and Sudlow (13) also made hot wire anemometer measurements in rigid tube bifurcation models. Working in planes both perpendicular to the bifurcation and in the same plane as the bifurcation, they mapped velocity profiles. Other investigators (14, 15) also
have made similar measurements including several genera-
tions of tubes. Simple analytical models of inspiratory
gas flow also are available (16). These present informa-
tion very similar to the model studies.

1.3 Objectives of Present Work

The author has selected to research the aspects
of an in vivo airflow velocity pattern encountered in the
upper airways of normal canines. In vivo velocity measure-
ments have been suggested as being too difficult to ac-
complish (17). Admittedly, virtually no work has been
reported where detailed velocity profiles were measured
in vivo. However, based upon the wealth of information
that could be gained and the significant advances possible
with this type of information, it seems a task well worth
attempting. Certainly the areas of diagnosis and treat-
ment of lung disorders would be enhanced through improved
pulmonary function tests, quantitative phonopneumography,
and the deposition and transport of aerosol particles,
all based upon actual velocity profile measurements.

The decision to investigate detailed velocity pro-
files in vivo significantly complicates the technical as-
psects of velocity measurement. However, only in vivo are
all of the influential parameters which act on the flow
conditions encountered. Flow profiles may prove to be quite different from those produced in models, since actual entry conditions and dynamic airways changes will be encountered. The inspiratory/expiratory flow cycle with its concomitant accelerations and decelerations may provide different stability characteristics than the usually encountered steady state inspiratory flows discussed in the literature (13, 14, 18). Elastic wall properties, although simulated in some model studies, may be complicated by neuromuscular interactions (13) in the in vivo case.

Using a single animal model will provide an initial baseline from which comparisons of changes in the diseased state could be made. This could provide the basis for understanding the interrelations of physical changes in disease and the fluid mechanics of lung flow. It is also important to note that, in vivo, the expiratory flow characteristics also will be investigated. A major portion of pulmonary function testing now in use relies heavily upon knowledge of the expiratory flow conditions alone (19, 20).

The primary goal of this investigation, then is to define, by experimental means, the velocity profile characteristics of the normal upper airways in canines.
This task is to be accomplished by the use of hot film/wire anemometry.

A secondary goal is to relate these velocity profiles to sounds recorded within the respiratory tree. Comparisons will be made of frequency spectra and magnitudes of both flow and sounds recorded simultaneously.
2.1 Physiology of Respiration

In order to be able to interpret the value of quantitative velocity measurements in diagnosis and treatment, one must fully understand what role fluid mechanical parameters play in the overall function of respiration. In order to accomplish this, one needs a strong understanding of the stages of respiration and how in each stage the fluid mechanics interact with the lung.

Respiration can be thought of as the transport of oxygen from the atmosphere to cells and the subsequent flux of carbon dioxide from the cells back to the atmosphere. This process is easily broken down into three steps: (1) pulmonary ventilation, that is, the movement of air between the atmosphere and the alveoli; (2) diffusion of respiratory gases between the alveoli and the blood, and (3) transport of oxygen and carbon dioxide in
the blood and body fluids to and from the cells. It is the first of these three steps that involves the internal aerodynamics under discussion here and which is the one to be discussed in detail.

In order for flow to occur within the airways, it is necessary that a pressure potential exist between the atmosphere and the alveoli. These pressure changes are brought about by the expansion and contraction of the lung in response to the respiratory musculature. During normal resting inspiration the lung is expanded, producing about a -3 mmHg intra-alveolar pressure (relative to atmosphere) and thus inducing an influx of air to the lung. During resting expiration, intra-alveolar pressure exceeds atmosphere by about +3 mmHg and outward flow occurs. In periods of maximal respiratory effort, pressure gradients in excess of 80 mmHg have been observed.

In order to create the pressure variations necessary for respiration, the volume of the thoracic cage is varied by skeletal muscles in two ways. The elevation or depression of the ribs to increase or decrease the anteroposterior diameter of the chest cavity is one way to vary the lung volume. This mechanism accounts for over 70 per cent of the maximal volume change of the
thoracic cavity. To understand this process, one must realize that during expiration the ribs extend downward from the spinal column; but during inspiration the ribs move upward to a more directly forward position, thus increasing the anteroposterior diameter. Movement of the diaphragm accounts for the remaining 30 per cent of maximal volume change. The bell-shaped diaphragm extends into the thoracic cavity under resting conditions. During normal inspiration the diaphragm is contracted, causing it to flatten and thus increasing the size of the thoracic chamber. Upon relaxation, the elastic structures of the lung and chest cage return the lung volume to its smaller original volume. For forced expiration, the abdominal muscles undergo contraction causing the abdominal contents to be pushed forcefully into the chest cavity and thus reduce the cavity volume. As mentioned above, the normal expiration is an entirely passive process dependent on the mechanical elastic equilibrium of the lung and chest wall.

The elastic nature of the lung tissue plus the even stronger forces of surface tension in the fluid lining of the alveoli results in a tendency for the alveoli to collapse. The net effect of the intermolecular forces of attraction between surface molecules is to reduce the size
of each alveolus, and thus the lung as a whole. These elastic forces result in a pressure of about -4 mmHg in the intrapleural cavity when the alveolar pressure is at atmospheric. This intrapleural pressure is a measure of the total tendency for lung collapse, and can, at large lung volumes, reach values as great as -12 mmHg. Figure 2.1 demonstrates a flow (volume) versus transpulmonary pressure curve from a supine canine preparation. Upon inspection one can discern the relationship between pressures and lung volume.

The ability to maintain a negative intrapleural pressure is the result of the membranes of the intrapleural space. These membranes absorb any gas or fluid entering the space between the visceral pleura of the lung and parietal pleura of the chest wall. This pressure (about 10-15 mmHg) maintains the pleura in contact with and lubricated by a protein-containing fluid. The lungs slide freely within the chest cavity since they are not physically attached except at the hila.

A graphical representation of the various changes in lung volume during different respiratory conditions can be seen in Figure 2.2. Each of the characteristic maneuvers of breathing has been represented by different
Fig. 2.1 Volume flow and transpulmonary pressure as a function of time.
Figure 2.2 Subdivision of the lung volume *

volumes and capacities. Four volumes are described which then summed result in the maximal volume of the lung.

(1) The tidal volume is the volume of air inspired or expired with each normal breath.

(2) The inspiratory reserve volume is the extra volume of air that can be inspired in excess of the normal tidal volume.

(3) The expiratory reserve volume is the amount of air that can still be expired by forceful expiration at the end of normal expiration.

(4) The residual volume is the volume of air remaining in the lungs even after the most forceful expiration.

In many pulmonary events, a combination of pulmonary volumes can be employed usefully. These volumes or capacities are:

(1) Inspiratory capacity is equal to the tidal volume plus inspiratory reserve volume.
(2) Functional residual capacity is equal to the expiratory reserve volume plus the residual volume.

(3) The vital capacity is equal to the inspiratory reserve volume plus total volume plus expiratory reserve volume.

(4) Total lung capacity is the maximum volume to which the lungs can be expanded with the greatest possible inspiratory effort.

In order to understand more fully the relationship between pressure, volume, and flow in the lung one must understand the concepts of compliance, airway resistance, and viscoelastic behavior of tissue. During breathing the respiratory muscles produce not only the forces required to perturb the elastic components from their mechanical equilibrium, but also the forces required to produce flow of gas through the airways and the sliding of the lung parenchyma and the tissues of the chest wall. These non-elastic forces are of a dissipative nature in that they require work in both directions of movement. Thus, pressure changes at the surface of the lung are not
those caused only by elastic recoil, but also include
pressure fluctuations related to the resistance to the
moving of air and tissue. Hence, if one is to determine
the elastic and flow-resistance properties of the lung,
it becomes necessary to separate their relative contri-
butions to the intrapleural pressure changes. This can
simply be accomplished if one considers a model of the
lung-chest wall system. Figure 2.3 is a representation
of this system in which all flow-resistive effort is
taking place in the tube and all elastic recoil in the
bellows. Figure 2.4 is a graph of volume, flow and
pressure for this model during one respiratory cycle.

In order to extract the elastic properties of
the system one must recognize that at two points in the
cycle, end inspiration and end expiration, all flow has
stopped. Since alveolar (bellows) pressure is now equal
to atmosphere at both instances, the difference in intra-
pleural pressure is balanced only by elastic recoil of
the lung. Expressed as the volume change per unit of
pressure change, the result is termed compliance. Hence,
compliance is the expansibility or distensibility of the
lung.

The flow-resistive properties of the model can
now be determined from the available information. This
Figure 2.3  A simple mechanical model of the lung*

Figure 2.4*

Volume, flow and pressure waveforms from a simple lung model.

is done by assuming that the compliance is linear over the volume range measured. Then at any volume between the minimum and maximum, the elastic recoil pressure component can be predicted. Subtracting this component from the total pressure change yields that per cent of the pressure required to provide flow at that given instant. By dividing the flow pressure component by the volume flow, the flow resistance is determined. This can be done at all points during the cycle.

With the knowledge of compliance and flow resistance of the lungs and chest wall, one can estimate the work done by the muscles of respiration. It is normal to express the work of breathing as the product of pressure and volume. From the geometry of the elastic pressure-volume graph, one can conclude that the elastic work is equal to one-half the product of the tidal volume and the total change in elastic pressure. From the definition of compliance, then

\[
C = \frac{Vt}{Pel} \quad (1)
\]

\[
Pel = \frac{1}{C} (Vt) \quad (2)
\]

\[
Pel = \frac{1}{2} Pel (Vt) \quad (3)
\]

\[
W_{el} = \frac{1}{2}C (Vt)^2 \quad (4)
\]
Thus, the elastic work per breath is inversely proportional to compliance and directly proportional to tidal volume squared. Now for the resistive work, the average resistance may be written

\[ R = \frac{P_{res}}{2(V_t)(f)} \]  

(5)

where \( P_{res} \) is equal to the average flow resistive pressure and \( 2xV_txf \) is the average rate of volume change. When \( F \) is the respiratory rate, flow resistive work is then equal to

\[ W_{re} = 2R(V_t)^2(f) \]  

(6)

and total work per minute is

\[ (W_{el} + W_{re})f = W_t \]  

(7)

or

\[ W_t = \frac{1}{2}C(V_t)^2(f) + 2R(V_t)^2(f)^2 \]  

(8)

Table 2.1 demonstrates the importance of the flow-resistive component at different ventilatory efforts.

2.2 Anatomy of the Canine Lung

The anatomy of the canine respiratory tract will be detailed here in order to provide an understanding of the geometry that plays such a strong role in governing the fluid mechanics of respiration. Naturally, the lung anatomy also defines restrictions as to the type and size of measurements possible within this system. Although it is felt that all structures preceding the trachea will
<table>
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<tr>
<th>Condition</th>
<th>External Work (Kg.m/min)</th>
<th>Tidal Volume (ml)</th>
<th>Resp. Rate (Breaths/min)</th>
<th>Min. Volume (L/min)</th>
<th>O₂ Consumption (ml/min)</th>
<th>RESPIRATORY WORK (Kg.m/min)</th>
<th>(O₂ cost 1 ml/min)</th>
</tr>
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<td>Quiet breathing</td>
<td>0</td>
<td>500</td>
<td>15</td>
<td>7.5</td>
<td>300</td>
<td>0.3</td>
<td>66</td>
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<td>Moderate exercise</td>
<td>620</td>
<td>1600</td>
<td>23</td>
<td>37.0</td>
<td>1500</td>
<td>5.2</td>
<td>57</td>
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<tr>
<td>Heavy exercise</td>
<td>1660</td>
<td>2400</td>
<td>48</td>
<td>115</td>
<td>3500</td>
<td>35.2</td>
<td>30</td>
</tr>
<tr>
<td>Maximal voluntary ventilation</td>
<td>0</td>
<td>1600</td>
<td>120</td>
<td>180</td>
<td>-</td>
<td>65.0</td>
<td>20</td>
</tr>
</tbody>
</table>

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influence the resultant inspiratory flow, no attempt was made to measure any parameter in these areas during experimentation. Instead one will begin here with the trachea as it represents the most proximal location of measurement.

The trachea is a flexible cartilaginous and membranous tube that forms the proximal part of the tracheobronchial tree. Just dorsal to the base of the heart, the trachea bifurcates into the left and right prenupal bronchi after running medially along the ventral aspect of the neck. Four major layers compose the wall of the trachea. From external to internal these are: the adventitia, the musculocartilagenous layer, the submucosa, and the mucosa. The musculocartilaginous layer may play an important role in governing fluid mechanics in this area although the other layers probably do not. The importance of the above mentioned layer rests with its ability to alter significantly the diameter of the trachea by means of contracting the tracheal muscle. The tracheal muscle extends across the gap of the incomplete cartilaginous rings (42-46), each defining the circumference at a given axial position. The
dorsal wall of the trachea, containing no cartilage, is free to draw the gaps in the ring together thus regulating the diameter of the tube.

The lungs, covered by pulmonary pleura and invaginated in the ipsilateral pleural sac in which it is free to move, are the paired respiratory organs utilizing much of the thoracic cavity. Figure 3.1 shows the anatomical relationships of the lobes of the lungs. The left lung consists of three lobes, the apical, the diaphragmatic, and the accessory. Although the relative position of the lobes is of interest, the internal network of the bronchial tree is of much greater concern.

The trachea bifurcates into right and left principal bronchi at the level of the fifth intercostal space. The tracheal bifurcation is situated slightly to the right of the midline, just dorsal to the base of the heart. The right principal bronchus passes caudolaterally from the tracheal bifurcation and enters the lung in the dorsal part of the hilus. Just after the principal bronchus enters the lung, it gives off the right apical lobar bronchus from its lateral aspect.

The right apical lobar bronchus bends cranially and gives off a large bronchus from its dorsal aspect. This dorsal bronchus ventilates the caudodorsal part or
caudal bronchopulmonary segment of the lobe and is known as the caudal segmental bronchus. The continuation of the right apical lobar bronchus passes craniocaudally toward the apex of the lung. It ventilates the cranial bronchopulmonary segment and is known as the cranial segmental bronchus. It gives off a series of dorsal and ventral subsegmental bronchi, of which the dorsal are by far the larger.

After giving off the right apical lobar bronchus the principal bronchus gives off the right middle lobar bronchus from its ventrolateral aspect. The right middle lobar bronchus passes ventrolaterally and slightly caudally and ventilates the right middle lobe. A short distance from its origin it gives off a relatively large bronchus from its lateral aspect which ventilates the dorsal bronchopulmonary segment of the middle lobe and is called the dorsal segmental bronchus. The right middle lobar bronchus continues toward the distal part of the middle lobe as the ventral segmental bronchus, ventilating the ventral bronchopulmonary segment.

Shortly after giving off the right middle lobar bronchus, the principal bronchus gives off the accessory lobar bronchus from its ventromedial aspect. The accessory lobar bronchus passes caudal into the accessory lobe
and divides into dorsal and ventral bronchi. The dorsal bronchus ventilates the dorsal bronchopulmonary segment of the accessory lobe and is called the dorsal segmental bronchus. The ventral bronchus ventilates the ventral bronchopulmonary segment of the lobe and is called the ventral segmental bronchus. After the principal bronchus has given off the accessory lobar bronchus, it continues caudal into the diaphragmatic lobe as the diaphragmatic lobar bronchus. It gives off, from its ventrolateral aspect, two bronchi which run in a ventral, lateral and caudal direction, and from its dorsal aspect two bronchi which run in a dorsal and caudal direction. The first ventrolateral bronchus ventilates the cranio-ventral part or ventral basal bronchopulmonary segment of the diaphragmatic lobe and is called the ventral basal segmental bronchus. The second ventrolateral bronchus ventilates the lateral basal bronchopulmonary segment of the diaphragmatic lobe and is called the lateral basal segmental bronchus. The first dorsal bronchus ventilates the cranial dorsal bronchopulmonary segment of the diaphragmatic lobe and is called the cranial dorsal segmental bronchus, while the second dorsal bronchus ventilates the caudal dorsal bronchopulmonary segment of the diaphragmatic lobe and is called the caudal dorsal segmental
bronchus. After giving off the ventral basal, the lateral basal, the cranial dorsal and the caudal dorsal segmental bronchi, the diaphragmatic lobar bronchus ventilates the dorsal basal bronchopulmonary segment of the diaphragmatic lobe and is called the dorsal basal segmental bronchus.

The left principal bronchus passes laterally and slightly caudally to enter the left lung at the hilus. After entering the lung, the principal bronchus gives off the apical lobar bronchus from its lateral aspect. The apical lobar bronchus is short and terminates by dividing into two bronchi. One bronchus bends cranially and passes toward the apex of the lung; it ventilates the cranial part, or the cranial bronchopulmonary segment, of the apical lobe and is called the cranial segmental bronchus. It gives off a series of dorsal and ventral subsegmental bronchi. The other bronchus passes ventrolaterally and slightly caudally to ventilate the caudal part, or bronchopulmonary segment of the apical lobe and is called the caudal segmental bronchus. The principal bronchus continues as the diaphragmatic lobar bronchus, ventilating the diaphragmatic lobe. The arrangement and distribution of the segmental bronchi are
similar to those of the right lung. In the distal part of the bronchial tree the terminal bronchioles lead into respiratory bronchioles.

This description completes the picture necessary for understanding measurements taken within the lung during these experiments.
Fig. 2.5 Dorsal view of the teased lungs of the dog
3.1 Fundamental Parameters

In this project there are a number of parameters that must be measured or monitored throughout each experiment. Airway velocity, volume flow, sound production, and lung geometry are the variables of prime concern. Velocity measurement will be accomplished by hot-wire/film anemometry, volume flow by a pneumotachograph, sound production by air-coupled microphones and lung geometry by a pediatric gastrointestinal fiberscope. Because of the extreme complexity of velocity measurements in vivo, that topic will be discussed initially.

3.2 Hot-wire Anemometry

Although the possible methods of anemometry are many, including laser doppler techniques, corona-discharge methods, as well as particle visualization approaches, none lends themselves to the working environment of this
research as well as does hot-wire/film anemometry. This device embodies the necessary frequency response, at the flow rates encountered, to accurately describe, not only the mean flow, but also the fluctuating components found in highly disturbed flow of the nature experienced in respiration.

The basic theory of hot-wire/film anemometry is reasonably simple. In essence, power is supplied to a wire or film element submerged in the flow. Energy is removed from the sensing element primarily by convection to the flow. The amount of power necessary to compensate for the lost heat is a measure of the gas velocity. Operating in the constant current mode, the anemometer system maintains a fixed current flowing through the probe, and changes in resistance of the sensor reflects the velocity changes in the medium. Compensation for the thermal inertia of the wire, which attenuates its amplitude response to high frequency fluctuation by a factor \((1+\omega^2M^2)^{-1/2}\), where \(M\) is the time constant, is effected by feeding the hot-wire input through an amplifier whose gain rises with frequency as \((1+\omega^2M^2)\). Thus, the effective time constant = 1/bandwidth represents the wire time constant divided by the maximum gain of the
amplifier. Since the time constant depends on the operating conditions, the compensation gains must be controlled manually.

The hot-film/wire anemometer can also be used in a constant temperature configuration. In this mode of operation the sensing element's temperature (resistance) is maintained constant by an amplifier in a feedback loop, usually a wheatstone bridge, which has the effect of compensating for the thermal inertia of the wire. In this configuration the inertia compensation is accomplished automatically. The stream velocity is then reflected in the feedback loop voltage necessary to maintain the temperature of the wire probe.

The fundamental relationship between flow velocity \( U \) and the amount of heat \( Q \) removed per unit was formulated by L. V. King in 1914. This is

\[
Q = P = A + B \cdot U^n
\]

(9)

where \( A \) and \( B \) are constant, assuming a constant fluid temperature and density. In practice \( A \) and \( B \) must be determined experimentally using a calibration method where velocity is known by some other means. Although time-constant fluctuations are automatically compensated for by the constant temperature feedback loop, the output of voltage vs
velocity is still non-linear. The distortion produced by this non-linearity is avoided by using a non-linear amplifier fed directly from the anemometer. This provides an output voltage directly proportional to mean velocity.

Figure 3.1 is a simple schematic of the constant temperature anemometer system. $R_1$ and $R_2$ represent fixed resistances in the Wheatstone bridge configuration. $R_3$ is a variable resistor used in providing the imbalance necessary to control the relative heating of the fourth resistor, which represents the probe. Initially the bridge is balanced to provide zero output for zero flow. $R_3$ is then adjusted to provide the necessary overheat ratio to the probe. Electric current from the servo-amplifier flows through the bridge, thus heating the sensor. The output current of the amplifier is controlled by the error signal between A and B. The amplifier is polarized so that a change in probe resistance due to cooling will result in an increase in amplifier output current. If the probe is now immersed in a flowing medium, the current that is fed-back by the servo-amplifier is proportional to the velocity of the stream. The voltage drop, $V$, across the hot film as a function of $\Delta T$ (overheat temperature difference) is described by

$$V^2 = a + b \left( U \right)^{1/2} \Delta T$$

(10)
Figure 3.1  Basic schematic of a constant temperature anemometer system
The instruments employed in this investigation were a D.I.S.A. anemometer 55D01 and a D.I.S.A. linearizer 55D10. D.I.S.A. probes were employed as well as some hot-film probes manufactured in our own laboratory.

Although the theory of hot-wire anemometry seems to imply that their use would be quite simple, the implementation of that theory is fraught with difficulties. By its design, hot-wire anemometers do not possess directional capabilities, making the interpretation of some flow conditions difficult. The temperature of the flowing medium is also extremely important, since overheat is based upon a single ambient temperature. This difficulty must be overcome before application in vivo is possible. Also, the extremely delicate nature of the wire probes used for high frequency measurements makes each measurement costly and frustrating. These and other problems had to be overcome before actual animal measurements could be initiated.

The selection of the type of probe to be employed in these investigations was in itself a major task, governed by a number of parameters. Included among the many constraints were the anatomy of the canine lung, the nature of the flow, the ambient conditions, aspects of the flow requiring measurement (turbulence), and financial
considerations. The final selection of probes used in these experiments is a compromise of all the above. The anatomy of the canine lung demanded the selection of a small probe to reduce the interference in locating a sensor within the flow. Since flows in airways as small as 2 mm in diameter were hoped to be studied, this provided an upper limit on the size of the probes to be used in these experiments. The turbulent nature of the oscillatory respiratory flow was also to be measured. If this were the only criterion, certainly hot-wire sensors would be chosen, since at the flow rates under consideration their frequency response is significantly better than hot-film probes. However, the geometry of the lung during respiration is always changing and the possibility of the probe contacting the lung during a measurement is great. With a hot-wire probe such contact means the destruction of the wire itself. Also, because the flow temperature was seen to vary with inspiration and expiration, temperature compensated probes were considered, only to be rejected because of size limitations. Finally, the decision was to use hot-film probes where only velocity information was to be measured, and to utilize the less robust hot-wire probes only in situations where frequency data was to be obtained. The probes selected were of two designs. The
wire probes were D.I.S.A. subminiature single-sensor probes with cylindrical sensors, type 55A55. The film probes were D.I.S.A. conical film probes 55R41. Before such probes could be used in measuring velocities, their characteristics first had to be ascertained.

The calibration of hot-wire/film probes for the velocity measurements here involved three separate procedures. These included a steady state calibration, a dynamic velocity calibration, and a frequency calibration. Universal correlations for heat transfer from hot-wires do exist, although none are available for films, but the uncertainties about the properties of the probe and fluid are such that individual calibration of probes is desirable if good accuracy is to be obtained.

The steady state flow calibration was accomplished in a fixed geometry flow stand using a rotometer (a calibrated volume flow meter). A fixed volume flow rate is set on the rotometer by controlling the orifice area from a constant pressure source of air. The output of the anemometer is then measured. The flow rate is systematically changed and the output noted each time. Repeating these steps provides an empirical relationship describing anemometer voltage output vs flow velocity. In this procedure the probes were located in midstream of a circular
cross-section tube whose length was adequate to insure fully developed flow profiles. Care was taken in the design to insure mechanical repeatability of the flow by maintaining $R_n$ within the laminar flow range.

A second, dynamic flow calibration was also utilized. In this instance, a sinusoidally oscillatory flow was generated by a Harvard Animal Respirator (Model #805) and calibrated against a pneumotachograph with previously determined characteristics. The output of the pneumotachograph and the hot-wire/film probe was compared. The anemometer system was then adjusted to provide a linear output at different flow rates.

Another property of hot-wire/film probes that must be determined is their frequency response. To quantify the frequency response of our hot-film and hot-wire probes, an adaptation of our steady state flow system was employed. With the probe in position and a steady flow passing over it, a second oscillatory flow was superimposed. This particular velocity field was developed by driving a large Acoustics Research (AR-3A) audio speaker with a signal generator, producing volume changes within the speaker's cabinet. The cabinet communicated with the steady flow tubes by means of a y-tube, thus inducing oscillatory flow past the probe. Using this technique,
adequate frequency response (250 Hz) has been demonstrated for both probe types, although the hot-wire probes perform well at frequencies above 1000 Hz. The response character is determined by observing the anemometer output on an oscilloscope or by ubiquitous frequency analysis.

As mentioned earlier, the lack of temperature fluctuations is extremely important to the adequate performance of the constant temperature anemometer. The approach selected to eliminate this problem in respiratory flow was to control the ambient temperature of inhaled gas to the same level as expired gas. A glass water-air heat exchanger was developed to provide uniform temperature gas to the experimental preparation. The temperature-flow characteristics of the heat exchanger are presented in Figure 3.2.

Effects of humidity on the performance of anemometer/probe systems also were of concern to the investigator. To determine this relationship, a test rig was designed to humidify gas. The humidified gas was passed over the probe at body temperature while flow was controlled by the rotometer and valve system. Little or no effect was seen as a result of the humidity in the air.

One final obstacle had to be overcome before velocity measurements could be accurately made. The position
Breaths per minute at 500 cc. per breath

Water temperature = 62°C
Water temperature = 49°C
Water temperature = 20°C

Fig. 3.2 Temperature flow characteristics of the respiratory gas heat exchanger
of the probe in the airstream had to be controlled and variable. Tracheal measurements were to be made by puncture of the tracheal wall. Based on the anatomical constraints, a simple probe support mechanism that would traverse the tracheal lumen was developed. Each complete turn of the advancing mechanism screw provided a 1 mm change in location. Using this device, the investigator could locate the far wall of the trachea and then move, 1 mm at a time, across the lumen, allowing 3 breaths at each location. Because of its construction, the traversing mechanism maintained its location relative to the trachea throughout a scan.

3.3 Pneumotachometry

The volume flow measurement utilized in these experiments was instituted primarily as a reference point, both for sounds recorded during the respiratory maneuvers and for velocities recorded during the breathing cycle. Because of the large variability in sizes and weights of test subjects, the absolute velocities as measured by anemometry were likely to vary considerably from animal to animal. It was felt that the availability of a continuous real time total volume flow signal would serve as a normalizing parameter, placing the velocities in their proper perspective. A second, and I feel more important
Fig. 3.3 Hot-wire anemometer probe advancing mechanism
aspect of having a volume flow signal, was its use as a
time reference for the other parameters measured in this
study.

The nature of oscillatory flows in tubes is such
that some radial locations within the lumen may begin
their deceleration phase and subsequent reversal prior to
other radial locations within the stream. This effect is
governed by the frequency of the oscillation of the
stream (f), the diameter of the lumen (D), and the kine-
matic viscosity (ν) of the flowing medium. This relation-
ship was described first by Womersley (21) in 1947, and
is characterized by

\[ \alpha = \frac{D}{2} \left( \frac{2\pi f}{\nu} \right)^{1/2} \]  

(11)

when \( \alpha \) is known as the Womersley parameter. In order to
maintain the phase relationships (that is, the relative
timing of flow reversal) between velocity measurements
taken sequentially at different locations, a baseline
signal was used. This baseline was the volume flow rate
signal. Hence, comparing each signal to the baseline
allowed comparisons between all velocity signals.

Sounds recorded during the breathing cycle also
must be oriented with respect to the cycle itself in
order to describe when the sounds occurred as well as
the intensity variations with time. Use of the volume
flow signal allowed a fixed reference system to which sounds can be compared. Also, since intensity of radiated sound is closely related to velocity, the volume flow signal can be used to normalize sound from different experiments and at different conditions.

The volume flow signal is measured by means of a pneumotachograph. This device is of a simple design, consisting primarily of a tube with a resistance area located across its lumen. As air flows through the tube, the pressure is reduced as it passes through the screen. Pressure ports located on either side of the screen measure a pressure differential related directly to the gas flow occurring within the tube. This pressure differential is sensed by an electronic differential pressure transducer. The electronic output of this device is amplified and is available for recording, either on FM tape or on the Brush direct writing chart recorder. For these investigations a Fleisch pneumotachograph, a Statham differential pressure transducer and a Fleisch heater supply were utilized. The heater supply allows approximately 2°F to 5°F overheat to be used in preventing water vapor condensation in the system.

The calibration procedure is also relatively straightforward. A rotometer is connected in series
with a vane pump driven by a variable speed electric motor. The flow rate is then set by observation of the rotometer. The electronic amplifier gain is then set to provide a fixed, convenient, voltage output. The voltage output will then reflect, in a known way, the pressure differential and hence the flow passing through the pneumotachograph.

3.4 Acoustic Recording

The characteristics of respiratory sounds may provide valuable insight about pulmonary dysfunction at many levels within the lung. For this investigation, however, it was only feasible to measure tracheal sounds in the dog. Due to the intensity of tracheal sounds, an analysis of breath sounds at locations further into the lung was not possible. As an integral portion of this investigation, breath sounds were recorded by two different means during many experiments.

In the first technique, a catheter-tip contact microphone (22) consisting of a miniature lead-titanate-zirconate bimorph ceramic piezo-electric element and six feet of cable was positioned in the esophagus of the dog. The position of this device was easily palpable in the exposed trachea preparation. An audio amplifier and speaker system provided the sound signal to the
investigator during the experiment. This technique allowed the investigator to maneuver the catheter-tip to provide an optimal breath sound signal. Such audio feedback was necessary because of the need for good contact of the microphone to the esophageal wall for proper performance. In order to obtain sound recordings at different locations along the trachea, it was a simple matter to reposition the microphone as desired. The usefulness of this sensor was limited, however, because of its extreme sensitivity to movement and contact artifact.

An alternative method for recording tracheal breath sounds was also used. This second method utilized an electret condenser microphone air-coupled to the lumen of the trachea by a 2 mm diameter tube. The tube penetrated the tracheal wall at whatever location was desired for sound recordings. The electret condenser microphone had a flat frequency response to 14,000 Hz with a dynamic range of 98 dB and a weight of 4.5 grams. The microphone and tube systems exhibited identical frequency response curves up to 2,000 Hz. The air-coupled microphone was sensitive to pressure fluctuations making it susceptible to the recording of pseudo-sound as well as sound. No attempt was made during this investigation to separate the relative contributions of these two components to the air-coupled sound recording.
3.5 Bronchoscopy

Bronchoscopy, the observation of the internal condition of the airways, has been utilized for some years in clinical work. The ease of observation and maneuverability of devices used for bronchoscopy has been enhanced greatly with the use of fiberoptics. Fiberoptics relies upon the use of light-carrying fibers that can be arranged to provide a coherent image along a flexible system. The bronchoscope, in the shape of a long flexible tube, can be placed in the lung by means of the mouth and trachea. A controlling mechanism allows manipulation of the distal end of the bronchoscope, providing the operator a means of selecting the desired path of inspection as well as what objects along the path to observe. Such a device was used in this study to provide information concerning probe placement and lung geometry.

The Olympus G.I.F. Type P fiberscope was used during this investigation. Its selection was based upon its applicability to both horses and dogs, as it will be utilized for both although not in this investigation. The physical dimensions of this scope included a working length of 1100mm, a maximum diameter of 7.2mm, and a bending section 60mm in length. The distal portion, operated by the cycle control, is 9.0mm in length and can be bent a full 300° (150° up and 150° down) in one plane.
The optical system provides an angle of view of 65° in the forward position over a range of 7-100mm. The light for the optical system is provided by an external light source which utilizes a second fiberoptic bundle to provide illumination at the observation end of the bronchoscope. An Olympus OM-1, 35mm single lens reflex camera with a focal plane shutter, linked with the external light supply for automatic exposure control, was used for all photography within the lung. Figure 3.4 is an example of a photograph taken to document the location of a hot-wire anemometer probe.

The bronchoscope also proved to be an excellent aid in the retrograde catheterization process described in Chapter IV, allowing the selection of the path to be followed by the retrograde catheter.

3.6 Data Storage

The quantity and nature of data gathered in these experiments placed a certain number of constraints on the data storage technique employed. The frequency content of both velocity and sound waveforms demanded high frequency (2000 Hz) storage capability. Velocity waveforms also possess portions of extremely low frequency oscillation. Both aspects had to be maintained, and phase relationships between sound, velocity, and volume flow had to be preserved as well.
Fig. 3.4 Photograph by bronchoscopy documenting the position of the retrograde probe and airways
Very few data storage techniques embody all of the above capabilities. Frequency modulated magnetic tape recording does include these aspects as well as providing a complete, compact record of all signals. FM tape also provides good interfacing capabilities with such peripheral devices as the ubiquitous frequency analyzer, analog to digital data conversion systems (for later use on digital computers), memory oscilloscopes, and lower frequency direct-writing chart recorders. Tape records can be processed a number of times employing differing techniques if need be. The ability to record information at rapid tape speeds and then process them at slower speeds provides an excellent means of studying short-term interval information, and to more accurately pinpoint specific events in the respiratory cycle. Data storage for these experiments was provided by means of a Tandberg T1R 115 Instrumentation tape recorder. This device provided four channels of available data storage.

Operating at a tape speed of 15 inches per second with a center carrier frequency of 27,000 Hz, the passband limits were at the lower frequency, D.C. and at the upper frequency and 5,000 Hz. A time multiplexed C.R.T. monitor connected to the outputs displays the instantaneous deviation of each channel.
CHAPTER IV

THE EXPERIMENT

4.1 Introduction

In the previous chapter the discussion centered around the selection and application of techniques for in vivo velocity measurement and analysis. In some instances the equipment selection was based upon the use of the canine animal model of ventilation. In this Chapter, the justification of this animal model will be presented as well as the experimental procedure utilized to achieve these measurements.

4.2 Animal Model

The selection of the dog as the research animal of choice in this investigation was by no means a simple one. The characteristic and fundamental differences between the human and canine tracheobronchial architecture are significant. However, the possibility of human studies is far in the future, and the applicability of
the canine to many other studies that could not be carried out in man overbalanced this initially negative aspect. Also, because of the suspected strong dependence of velocity profiles on local geometry changes, little hope was held that extrapolation from one species to another might be possible. Rather, it was hoped that an understanding of fluid mechanical events might be developed as well as the expertise for later studies in man.

One major positive aspect of the canine model selection was its size. The animal is large enough to provide the possibility of measuring velocity profiles in the upper airways, yet small enough to be handled and positioned by a single investigator. This relative ease of handling is also based upon the animal's normally docile nature, making chronic investigation significantly easier. The dog is also a readily available species, thus reducing any waiting period for delivery and making them extremely practical financially (since they neither have to be boarded for long periods of time nor are they initially expensive). A significant quantity of pulmonary function data exists for dogs, and the animal demonstrates heartiness throughout extended investigation.

Research was already underway within the Cardio-Pulmonary Research Group at The Ohio State University
investigating bronchopulmonary sounds in the canine pulmonary edema model. Compatibility with this work presented another reason for selecting the dog over any other animal. Since future chronic disease studies were also anticipated in the dog, it was hoped that a continuation of this work would establish a strong normal baseline as well as take advantage of these studies to show the influence of disease on the internal aerodynamics of the respiratory tract. The advantage would be that all measurements would be in the same species.

Finally, this author has had significant experience with research in canines and thus choosing this model takes advantages of that experience.

4.3 Animal Preparation

The experimental procedure would begin by removing the dog's food supply the day preceding the actual measurements. This was done in response to initial investigations revealing that the animal was prone to vomiting while under the influence of the anesthetics employed, especially in the supine position. This often lead to the aspiration of vomitus and hence to significant changes in sound production and fluid accumulation in the airways.

The following day the animal received by subcutaneous injection, acepromazine at the recommended
pre-anesthesia dose level of 10 mg/kg. A period of a quarter of an hour would then elapse, allowing the pre-anesthesia to take effect. Sodium Pentobarbital (Nembutal) was then administered by intraperitoneal injection at the prescribed 30 mg/kg and another period of 15 minutes would elapse before surgical level anesthesia was reached. Naturally each animal was weighed just prior to anesthesia to determine the appropriate dosage levels. When anesthesia was complete, the animal would be placed in a supporting structure to maintain its selected supine position. Straps were applied to the animal's forelegs to secure its position. At this point in the investigation, the ventral aspect of the animal was shaved in preparation for the surgical exposure of the trachea.

4.4 Surgical Procedure - Trachea

The initial incision was made along the ventral midline extending posteriorly approximately 30 cm from the site of the thyroid cartilage to the anterior process of the sternum. Both of these anatomical points are readily located by palpation. Utilizing blunt dissection techniques, the trachea is fully exposed by separating the sternohyoid muscle along its centerline and by cutting the sternomastoid muscle at its origin, the manubrium of the sternum. Care is taken to expose as much of the
trachea as possible, while not permanently displacing it from its original position within the neck.

The air supply tube was then introduced into the buccal cavity and precautions were taken to assure that all respired air was received from the heat exchanger-pneumotachograph system discussed in Chapter III. The Geddes (22) contact-tip piezo-electric microphone was introduced into the esophagus through this tube in order to record sound near the point of the velocity measurements. The position of this microphone can easily be determined by feel since the esophagus is an extremely flexible structure.

A small crosscut was made in the ventral midline of the trachea between two of the cartilagenous rings approximately 2 cm caudal of the laryns. This small tracheostomy is produced using the Bovie electrosurgical system, thus providing a clean opening with a minimum of bleeding and seeping from the tissue. This is an important point since these fluids foul the probes, causing poor performance, and lead to the accumulation of fluid in the airways.

Provided all has gone well to this point, the probe traversing mechanism then was situated on the trachea and the guide tube pushed through the tracheostomy.
The tube was initially moved into the trachea about 4 mm and then withdrawn to a distance equal to the wall thickness of the trachea. This pulls the traversing mechanism in contact with the tracheal wall and reduces the chance of the wall or guide tube protruding into the lumen. The hot-wire/film probe then was advanced across the lumen of the trachea in 1 mm steps, usually allowing 3 to 4 breaths between changes of position. Upon reaching the dorsal wall of the trachea, the probe is retracted in the same manner as it had been advanced, providing a check on the repeatability of the velocity measurements. Upon completing a scan, the mechanism was removed and the tracheal diameter, both internally and externally, determined. The external diameter was measured by means of direct vernier caliper reading and the internal diameter by using a small glass micropipet. The pipet was used because of its very low mass, which does not displace the dorsal tracheal wall when inserted. Since the dorsal wall does not contain the cartilage rings, but only smooth muscle and fibrous tissue, it is easily moved by most measuring devices. The initial tracheostomy was then sealed and a second introduced in the horizontal plane, that is a 90° rotation in the plane of the rings. The measurement procedure then was repeated
in this plane, as well as new sites along the trachea until it enters the thoracic cavity.

Sound recordings by a second microphone were also made during the trans-tracheal investigation, but not during the profile runs. In these sound studies, a microphone was air-coupled to events in the trachea at the different sites of investigation by means of a flexible plastic tube.

4.5 Surgical Procedure - Upper Airways

A second surgical procedure was employed for a group of experiments designed to measure central airways velocities and to determine the penetration of turbulence into these airways. The initial animal preparation remained the same as outlined in the beginning of this Chapter. However, it was necessary to employ radically different methods to make velocity measurements in the central airways.

The surgical procedure employed in this group of experiments was based upon a retrograde catheterization technique similar to those described by Macklen et al. (23). Our procedure was enhanced by the availability of a pediatric gastrointestinal fiberscope utilized as a bronchoscope in this investigation.
The bronchoscope was advanced with care into the canine lung as far as possible. The path was controlled by the bronchoscope operator who was monitoring the travel of the device tip by the fiberoptics eyepiece. Once a maximum penetration had been obtained, a stainless steel wire was advanced through the biopsy tunnel of the scope until it intersected the lung wall. The wire tip was sharpened to a point and would penetrate the lung wall. With increased pressure the wire would work its way to the thoracic wall and then out between the ribs of the dog. Once trans-thoracic penetration had been completed, the bronchoscope was removed and a catheter fastened to the stainless steel wire at the mouth. The catheter was then drawn into the lung by pulling the wire out through the thoracic puncture. When the catheter is in position, it is sutured to the skin of the thorax. Using the bronchoscope again, the investigator now inserts a 2 meter guide tube into the trans-thoracic catheter from mouth to chest wall. The 2 meter length provides access to the guide tube at both mouth and chest wall simultaneously. The hot-wire probe catheter was then inserted into the guide tube with the probe portion last. The leads from the probe, now extending from the thoracic position of the guide tube, are soldered to a connector and attached
to the hot-wire anemometer. Figure 4.1 demonstrates the basic steps of retrograde catheterization. The probe is then drawn into the guide tube for protection and the guide tube is pulled into the lung airways from the chest position. By following the receding probe with the bronchoscope, the investigator can determine when a measurement position has been reached and if the probe location is in the center of the airway. If everything is in position, then the hot-wire probe is advanced 2 mm past the opening of the guide tube, giving it an unimpeded location to measure inspiratory velocities. Once measurements have been obtained at a given location, the probe is withdrawn into the guide tube and the entire guide tube-probe system is moved to a new location further into the lung where the above procedure is repeated. Finally, the guide tube containing the hot-wire probe is withdrawn through the trans-thoracic catheter and the measurement procedure is completed.

4.6 Fluid Mechanics Parameters

The pulmonary system generates a fluid mechanical situation in which a great number of parameters can and do play an important role. The flow within this system can be described as an oscillatory flow containing secondary motions induced by area changes and curvature, with the
Fig. 4.1 Schematic representation of retrograde catheterization technique. Sequence progresses from left to right.
possibility of being laminar, disturbed and transitional, or even turbulent. Such flow properties as density ($\rho$), viscosity ($\mu$), respiration frequency ($f$) and velocity ($U$) can interact with the geometric parameters encountered in the lung, e.g., the lung configuration, component dimensions, and wall elasticities.

In order to properly describe the nature of the fluid motion encountered in these experiments, several basic fluid mechanical parameters will be used. These are the Reynolds number ($Rn$) and the Womersley parameter ($\alpha$) [26].

The Reynolds number is a non-dimensional parameter calculated from the fundamental flow properties, i.e., the density ($\rho$), the viscosity ($\mu$), the characteristic length scale ($L$) and the flow velocity ($U$). The equation for the Reynolds number is

$$Rn = \frac{\rho UL}{\mu}$$  \hspace{1cm} (12)

This equation represents physically the ratio of inertial forces acting in the flow to the viscous forces acting in the flow. While $Rn$ is small, disturbances are dampened by viscous forces, while when $Rn$ is large transition to turbulent flow may occur. Since inertia and viscosity are the two primary influences on the flow in the respiratory tract, the Reynolds number will provide a basis for
similarity comparison between different flow conditions and different animals. The Reynolds number will be computed for each flow condition measured in these experiments. The Womersley unsteadiness parameter (a) described earlier in this section will be used to characterize the unsteadiness of the velocities encountered in the test subjects.

Womersley (21) provided an excellent analysis of the interaction of the varying pressure gradient and the inertial effects of the flow in both rigid tubes and elastic-walled conduits. In order to get a basic feeling for the effect of a periodically varying pressure gradient, the following simplified approach will first be used.

For a fully developed laminar flow in a straight rigid tube, the velocity along the tube is independent of axial distance (z). The equation of motion in cylindrical coordinates is

\[ \frac{\partial U}{\partial t} = -\frac{1}{\rho} \frac{\partial P}{\partial z} + \nu \frac{\partial^2 U}{\partial r^2} + \frac{1}{r} \frac{\partial}{\partial r} \frac{\partial U}{\partial r} \]  

(13)

Using complex notation, the oscillatory pressure gradient can be written as

\[ -\frac{\partial P}{\partial t} = P e^{i2\pi wt} \]  

(14)

Assuming a solution for \( U \) in the form

\[ U = U_1 e^{i2\pi wt} \]  

(15)
and non-dimensionalizing, \( y = r/a \), where \( a \) is the radius of the tube, then equation 16 becomes

\[
\frac{d^2 U_1}{dy^2} + \frac{1}{y} \frac{dU_1}{dy} - \frac{i(a^2 \pi \omega)}{\nu} U_1 = \frac{Pa}{\nu} \tag{16}
\]

Now it can be seen that the parameter controlling the solution of this equation is

\[
\alpha^2 = a^2 \pi \omega / \nu \quad \text{or} \quad \alpha = (a^2 \pi \omega / \nu)^{1/2} \tag{17}
\]

This is the Womersley parameter, \( \alpha \), and in the limit \( \alpha = 0 \), the result becomes the well known Poiseuille pipe flow problem which is a steady state solution where the flow and the pressure gradient are in phase. When \( \alpha \neq 0 \) but remains small, then the profile will still be parabolic, but the flow will be slowly time varying. If \( \alpha \rightarrow \infty \) then we can neglect viscous effects and the pressure and flow are shifted 90° out of phase.

The Womersley (26) or unsteadiness parameter physically represents the radius of the tube, divided by the Stokes viscous boundary layer thickness. Alternatively, \( \alpha \) represents the time for viscous diffusion to occur across the vessel divided by the time available in one cycle of periodicity. For large \( \alpha \), the cycle time would limit the amount of viscous diffusion. Womersley as well as Atabek and Lew (24) have considered the problem of elastic walls under conditions described above. Olson (12) indicates that such solutions show that wall
flexibility of the nature encountered within the pulmonary airways has little effect on the motion of the fluid. The fact that the tubes in question are not infinitely long (in general only 200 diameters) and also are curving has been considered by previous researchers (12, 24, 10, 13). Their conclusions imply that a quasi-steady approximation to pulmonary lung flow is adequate. It is also noted that the respiratory pattern is not sinusoidal, but that relatively constant velocities are often encountered over as much as 70 per cent of the inspiratory flow pattern.

The trachea, however, presents a significantly different picture than the remaining airways because the flow encountered at almost all conditions there is disturbed (not laminar) and is not of the entry type flows that are experienced in the other conduits of the lung. Pedley et al. (17) suggested that the effects of unsteadiness on turbulent flows with unsteady mean velocities might be assessed in terms of the eddy viscosity. Pedley calculated a new turbulent unsteadiness parameter. Pedley's results suggest that unsteadiness may have some effect on flow in the trachea, but probably not as significant as first suggested by Chow and Lai (25).
The task of presenting, in a meaningful manner, the enormous quantity of data involving velocity profiles gathered in one experiment is indeed difficult. During a typical experiment, seven axial positions in the trachea might be examined in both the sagittal and lateral planes. In each plane, as many as ten different radial positions would be measured. At each radial position, 6 particular velocities related to the timing of the respiratory flow cycle are analyzed. This procedure results in a total of 840 velocity values calculated for a typical experiment. The author's inclination was to not present data of this magnitude in tabular form, since interpretation of such tables would boggle the mind (at least of this simple graduate student). Instead, a graphical method was employed presenting all of the velocity profiles corresponding to a particular portion of the breathing cycle (for example, after one-fourth of the duration of inspiration) on one graph.

Each graph presents two views of the trachea enabling the reader to envision the profiles taken in either the sagittal plane or the lateral plane at different axial positions along the trachea at the same relative time in the cycle. A total of six graphs, corresponding
to the six time segments of each cycle is presented to represent a single experiment fully. The six time points, at which each velocity profile is evaluated, are characterized in Figure 4.2. Inspiration and expiration are each measured at 3 intervals corresponding to 25%, 50%, and 75% of their total duration. Utilizing this technique of presentation allows a visual analysis of the effects, if any, of the unsteadiness of the flow as well as presenting a sequential development and decay of both the inspiratory and expiratory velocity distributions.

Velocity values depicted in each of the graphs described above have been normalized to the peak velocity encountered at any location measured at that particular time segment. Hence, each value on the graph will be between zero and one. Arrows are used to represent the magnitude of each velocity so that a profile can easily be evaluated. Un (the normalized velocity) was calculated by dividing the actual measured velocity (U) by the peak velocity (U_{pn}) at the time segment under evaluation (n%). Figure 4.3 demonstrates this approach at 25% of inspiration. The peak velocity, time segment, and Reynolds number are presented so that comparisons between experiments are readily possible. The normalization
Fig. 4.2 Characteristic velocity profile with the six time measurement points identified.
Fig. 4.3 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF INSPIRATION ILLUSTRATING THE DATA PRESENTATION FORMAT

Z = DISTANCE FROM THYROID CARTILAGE (CM)
technique was selected for two reasons. First, this method allows for a much enhanced graphical representation thus making interpretation of the results simpler. Normalizing in this manner also yields a much more straightforward means of comparison between experiments and between conditions during an experiment.

As mentioned earlier, the velocity waveforms encountered in the trachea of canines during quiet breathing often appeared to be turbulent. Turbulence may be characterized as a high-disturbed flow regime in which chaotic, random fluctuations of the flow are manifest. The perturbations to the mean flow are three-dimensional in nature possessing a broad flat frequency distribution and are usually on the order of 10% of the mean flow (27). Large pressure drops, increased friction and energy dissipation, and sound generations are all associated with the occurrence of turbulence.

Although turbulence is often characterized on the basis of its intensity and scale, the turbulence spectrum will be employed in this effort as the means of identifying and quantifying this phenomenon. It is convenient in the discussion of turbulent flows to define the total flow \( (\mathbf{u}) \) as the sum of the mean flow \( (\overline{U}) \) and a fluctuating component \( u' \). That is

\[
U = \overline{U} + u'
\]  

(18)
\[ \int_0^\infty F(n)dn = 1 \tag{19} \]

thus
\[ \frac{1}{T} \int_0^T u'dt = \overline{u'} = 0 \tag{20} \]
defines a fluctuating component \( u' \) superimposed upon the mean flow (U). \( T \) must be long enough in duration for \( U \) to be independent of time. Hence, the mean kinetic energy associated with the \( u' \) term can be presented as \( u'^2 \).

\( F(n) \), then, represents the frequency distribution of the mean kinetic energy. Such distributions usually peak about a frequency described by \( U/L \), where \( U \) is the propagation velocity of the turbulence and \( L \) is the scale of the turbulence (usually about the characteristic length of the geometry).

Energy containing eddies will exist on both sides of this frequency but the distribution will go to zero in both directions. Nerem and Seed (41) indicate that for a periodic unsteady flow the frequency distribution will be a combination of the unsteady velocity waveform frequency components and the fluctuating components. This overlapping may well obfuscate the distinction between laminar and turbulent flows.

4.7 Frequency Analysis

The Federal Scientific Model U.A.-14A Ubiquitous Frequency Analyzer performing with 400 line resolution was employed to develop frequency data from the hot-wire
anemometry measurements. Essentially, the analyzer
low pass filters the input to prevent the introduction
of unwanted signal components in the sampler and then
converts the analog input signal to a digital form to
be stored in parallel in a recirculatory digital memory.
A contraction of the signal's time base is accomplished
thus providing an expansion of the frequency spectrum,
i.e., if the time required for a complete cycle of a sine-
wave is halved, the frequency is doubled. The signal is
then fed to a narrow-band crystal filter (24dB/octave)
which observes frequency resolution elements. The output
of the device is stored to await selection of a presenta-
tion mode. The modes are linear, log, or square-law
spectrum amplifiers.

A device designed and built by a member of The
Ohio State University (28) provides a time-dependent fre-
quency analysis when used in conjunction with the ubiqui-
tous analyzer. The output of this instrument includes
both the amplitude spectrum and the real time flow wave-
form. Figure 4.4 is an example of this type of data pre-
sentation. The velocity waveform is displayed in the
left border of the oscillographic display, time increas-
ing from top to bottom and magnitude increasing from
left to right. Since the hot-film anemometer system
does not possess a directional capability, both inspiration and expiration are seen as positive displacements on this curve. The device provides a partially updated spectrum every 100 milliseconds, corresponding to that portion of the input signal directly preceding the spectrum and defined by one bandwidth. This display is seen on the right three-fourths of the oscillographic display. Frequency is from left to right in a linear fashion while the amplitude corresponding to a given frequency is on the vertical axis. The amplitude spectrum is displayed every 100 milliseconds and is presented directly to the right of the portion of the velocity waveform that was analyzed. Such a presentation format allows direct comparison of amplitude spectra at different intervals during the breathing cycle as well as providing a sequential picture of the changing amplitude spectrum during a given period of time. The frequency range is variable and is the gain of the spectral signal. Both the frequency range and the amplitude can be displayed in either linear or logarithmic fashion. The amplitude can also be displayed as a square-law function characterizing the energy distribution.
There is another aspect of this device which is also useful. Since the information is provided from magnetic tape, one signal can be displayed in the time-amplitude mode while another can be frequency analyzed. The result then is a display demonstrating the frequency content of any parameter during a period of the second signal. For example, the flow waveform is presented in the left border display; and if the magnitude spectrum is determined from the sound signal occurring simultaneously to the flow, then the display provides information revealing the frequency content of the sounds occurring at the corresponding velocities. Such a display is useful in itself, but when compared to the same analysis of the velocity spectrum provides direct sequential comparison of the flow and sound spectra at the same moment during the respiratory cycle.
CHAPTER V

RESULTS

5.1 Introduction

This Chapter presents the results of the airways velocity and sound measurements described in the previous chapters. Table 5.1 and all related figures are presented. Table 5.1 summarizes the experiments performed in this research effort. Included in the tabulation is the number of measurement sites, the type of probes used and pertinent comments for each experiment. Velocity profiles, time-dependent velocity spectra, and time-dependent sound spectra are all employed to characterize lung airflow and therefore comprise the major results of this investigation. Time-dependent velocity spectra are also presented in a qualitative manner for airflow below the trachea.
5.2 Velocity Profiles

Normalized velocity profiles are presented for six time segments during each respiratory cycle, corresponding to 25%, 50% and 75% of both the inspiratory and expiratory phase. The particulars of the normalization techniques was discussed in Section 4.6. Figures 5.1 through 5.36 depict complete velocity waveforms for six representative experiments. These characterize the observations of tracheal air flow patterns. In general such velocity patterns appear as blunt or flat profiles characteristic of highly disturbed flow where rapid momentum transfer is possible. The profiles presented are also asymmetric. This is particularly notable near the more cranial aspects of the trachea where variable and non-symmetric geometry is governing the flow. The velocity patterns themselves are time-dependent, but the normalized profiles do not depict strong changes in shape within a given phases. A possible exception is that at the higher flow rates the profiles are somewhat flatter than at lower rates.

Several particularly interesting aspects of the flow are illustrated in selected experiments. In some instances the flow seems to rotate as it moves along the
trachea. In Figures 5.31, 5.32, and 5.33 the peak inspiratory velocities are found near the right lateral wall 2 cm from the thyroid cartilage. As the flow progresses, a more blunt profile appears in the ventral view at 7 cm from the thyroid cartilage. At 9 cm from the thyroid cartilage the peak flows are now near the left lateral wall. The sagittal view appears to reinforce this flow pattern seen in Figure 5.33, suggesting that the flow rotates. Figures 5.19, 5.20, and 5.21 suggest such a pattern of flow rotation. This is possibly introduced by the tracheal curvature in the ventral-dorsal plane of the supine dog. Olson (10) suggests such a mechanism in the human lung.

In three of the six experiments illustrated (Figs. 5.10, 5.14, 5.36), a strong core component of velocity seems to develop during expiration in the ventral-dorsal plane, i.e., the plane normal to the plane of the first bifurcation. In Figure 5.36 the bi-modal pattern of the two bronchi are clearly defined at the location 17 cm from the thyroid cartilage, 1 cm from the carina in this instance (measurements made by bronchoscope manipulation of probe). In the plane of the bifurcation this pattern alternately appears and disappears as the flow moves up the trachea, decaying in about 5
diameters. Schroter and Sudlow's (13) model studies suggest that such a pattern might develop as a result of secondary motions concentrating momentum in the bifurcation plane and dissipating momentum at the wall in the normal plane. At first this result seems in contradiction of our observations; however, when considered with the earlier suggested flow rotation pattern, secondary motions could account for the observed flow patterns.

The strongly skewed profiles of Figures 5.31 through 5.36 and 5.19 through 5.24 clearly indicate an asymmetric aperture for flow entrance and exit to and from the trachea. Such an orifice allows a preferential flow pattern to develop which seems to persist about 5 diameters into the trachea. This aspect of the velocity patterns measured in vivo is extremely important as it indicates that it is the anatomy itself that strongly dominates much of the flow characteristics measured in this study.

Flow patterns at different time segments within a given phase, e.g., inspiration or expiration, are strongly similar indicating that the flow moves with a quasi-steady pattern.
5.3 Velocity Spectra - Trachea

The frequency display techniques discussed in Chapter IV are utilized to present time-dependent velocity spectra for three typical experiments. Centerline velocity traces and their time-dependent spectra are presented in geometric sequence starting at the location nearest the thyroid cartilage and progressing down the trachea to the final, most caudal measurement site. These results appear in Figures 5.37 through 5.50. In all of these figures frequency and amplitude are on linear scales, the frequency range being from 0-2000 Hz. An effort has been made to maintain signal levels approximately equal so that direct comparison of frequency and amplitude can be made from signal-to-signal. Such a procedure serves as a rudimentary means of normalizing for different absolute velocities and thus for different total energy levels. These results indicate that during peak expiratory flows, velocity perturbations are consistently observable at frequencies up to 1500-2000 Hz. This result is consistent with turbulent flow being expected in the trachea. Inspiratory flows also contain high frequency disturbances approaching 2000 Hz. An analysis of sequential locations does not demonstrate any measureable decay of the frequency spectra with locations within the trachea.
5.4 Upper Airways

Two methods were employed in order to assess the nature of flow occurring in the upper airways. Time-dependent velocity spectra as well as turbulence intensities were studied to provide information concerning the flow. The results of typical retrograde velocity experiments are documented in Figures 5.51 to 5.61.

Figures 5.51, 5.52, 5.53, 5.54 and 5.55 represent a serial presentation of waveforms taken in the baseline, trachea, first, second, and fourth generations, respectively. A trend of diminishing frequency content can be identified when the waveforms are studied in succession.

5.5 Lung Sounds

Figures 5.62 through 5.70 are the time-dependent sound spectra obtained along the trachea for consecutively increasing distances from the thyroid cartilage. Three experiments are represented in this series of figures. The spectra are broad uniform distributions extending to 2000 Hz. These spectra resemble the spectra generated from turbulent velocity waveforms also measured in the trachea. This similarity suggests that the velocity fluctuations and their concomitant pressure fluctuations may be related to the pressure fluctuations measured as sound in the vicinity of the trachea. Figure 5.71 presents the synchronized...
velocity and sound signals recorded in the trachea. The beginning of recordable sound is seen to be closely related to the rapid onset of disturbed flow within the trachea.
## TABLE 5.1

### SUMMARY OF EXPERIMENTS

<table>
<thead>
<tr>
<th>Experiment #</th>
<th>Type of Probe</th>
<th># of Measurement Sites</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-11</td>
<td>Hot Film and Hot Wire</td>
<td>variable</td>
<td>experiments performed to develop technique - no useful data obtained</td>
</tr>
<tr>
<td>12</td>
<td>Hot Wire</td>
<td>4 lateral planes</td>
<td>first complete measurement set including sound</td>
</tr>
<tr>
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<td>Hot Wire</td>
<td>4 lateral planes</td>
<td>strong bimodal distribution during expiration</td>
</tr>
<tr>
<td>14</td>
<td>Hot Film</td>
<td>4 lateral planes</td>
<td>poor high frequency resolution</td>
</tr>
<tr>
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<td>Hot Film</td>
<td>3 lateral plane</td>
<td>poor high frequency resolution</td>
</tr>
<tr>
<td>16</td>
<td>Hot Film</td>
<td>3 lateral plane</td>
<td>poor high frequency resolution</td>
</tr>
<tr>
<td>17</td>
<td>Hot Wire</td>
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<td>some jet-like flow patterns</td>
</tr>
<tr>
<td>18</td>
<td>Hot Wire</td>
<td>5 lateral plane</td>
<td>strong jet patterns during respiration good example of rapid flow development</td>
</tr>
<tr>
<td>19</td>
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<td>5 lateral plane</td>
<td>dog expired before completed run</td>
</tr>
<tr>
<td>Experiment #</td>
<td>Type of Probe</td>
<td># of Measurement Sites</td>
<td>Comments</td>
</tr>
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<td>---------------------</td>
<td>------------------------</td>
<td>------------------------------------------------------------</td>
</tr>
<tr>
<td>20</td>
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<td>1 dorsoventral plane</td>
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<tr>
<td>21</td>
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<td>1 lateral plane</td>
<td>incomplete</td>
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<tr>
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<td></td>
<td>3 dorsoventral plane</td>
<td></td>
</tr>
<tr>
<td>22</td>
<td>Hot Wire Epoxy</td>
<td>2 lateral plane</td>
<td>moderate frequency</td>
</tr>
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<td></td>
<td>2 dorsoventral plane</td>
<td></td>
</tr>
<tr>
<td>23</td>
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<td>3 lateral plane</td>
<td>response but inadequate for frequency analysis</td>
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<td></td>
<td>2 dorsoventral plane</td>
<td></td>
</tr>
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<td></td>
</tr>
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</tr>
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<td>3 dorsoventral plane</td>
<td></td>
</tr>
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<td></td>
<td></td>
<td>4 dorsoventral plane</td>
<td></td>
</tr>
<tr>
<td>27</td>
<td>Hot Film</td>
<td>3 lateral plane</td>
<td></td>
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<tr>
<td></td>
<td></td>
<td>3 dorsoventral plane</td>
<td></td>
</tr>
<tr>
<td>28</td>
<td>Hot Wire</td>
<td>5 lateral plane</td>
<td>good example of bimodal expiration patterns good, high frequency information</td>
</tr>
<tr>
<td></td>
<td></td>
<td>3 dorsoventral plane</td>
<td></td>
</tr>
<tr>
<td>Experiment #</td>
<td>Type of Probe</td>
<td># of Measurement Sites</td>
<td>Comments</td>
</tr>
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<td>---------------</td>
<td>------------------------</td>
<td>------------</td>
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<td>Hot Film</td>
<td>Generations 0,2,3,4</td>
<td></td>
</tr>
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<td>Hot Film</td>
<td>Generations 0,1,2,3</td>
<td></td>
</tr>
<tr>
<td>R-4</td>
<td>Hot Film</td>
<td>Generations 0,1,2,4</td>
<td></td>
</tr>
<tr>
<td>R-5</td>
<td>Hot Film</td>
<td>Generations 0,1,2,3</td>
<td></td>
</tr>
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Fig. 5.1 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF INSPIRATION
EXP. 12 \( v=100 \text{ cm/SEC} \) RN=1111
NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 50% OF INSPIRATION

Exp. 12  V=200 cm/sec  RN=2222
Fig. 5.3 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF INSPIRATION
EXP. 12 V=100 CM/SEC RN=1111
Fig. 5.4  NORMAlIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF EXPIRATION
EXP. 12  V=512 CM/SEC  RN=5688
Fig. 5.5  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 50% OF EXPIRATION
EXP. 12  V=200 CM/SEC  RN=2222
Fig. 5.6 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF EXPIRATION
EXP. 12 V=70 CM/SEC RN=777
5.7 Z DISTANCE FROM THYROID CARTILAGE (CM)
NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF INSPIRATION
EXP. 13 V=280 CM/SEC RN=2379

Fig. 5.7

Z= DISTANCE FROM THYROID CARTILAGE (CM)
NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF INSPIRATION
EXP. 13 V=280 CM/SEC RN=2379
Fig. 5.8 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 50% OF INSPIRATION

EXP. 13 V=240 CM/SEC RN=2039
Fig. 5.9 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF INSPIRATION
EXP. 13 V=172 CH/SEC RN=1461

Z = DISTANCE FROM THROAT CARTILAGE (CM)
Fig. 5.10 NORMALIZED VELOCITY PROFILES IN CANINE TRACHEA AT 25% OF EXPIRATION
EXP. 13 V=400 CM/SEC RN=3398
Fig. 5.11 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 50% OF EXPIRATION
EXP. 13 \( v=150 \text{ CM/SEC} \) RN=1274
Fig. 5.12  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF EXPIRATION
EXP. 13  V=50 CM/SEC  RN=424
Fig. 5.13  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF INSPIRATION
EXP. 17  V=190 CM/SEC  RN=2359
Fig. 5.14 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 50% OF INSPIRATION

EXP. 17 \( \dot{V}=250 \text{ cm/sec} \) RN=3104

\( Z = \text{DISTANCE FROM THYROID CARTILAGE (cm)} \)
NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF INSPIRATION

Fig. 5.15  EXP. 17  V=173 CM/SEC  RN=2148
NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF EXPIRATION

**Fig. 5.16**

Z = DISTANCE FROM THYROID CARTILAGE (CM)

EXP. 17 V = 490 CM/SEC RN = 6084
Fig. 5.17 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 50% OF EXPIRATION
EXP. 17 $V=222$ CM/SEC RN=2756

SAGITTAL VIEW TRACHEA

VENTRAL VIEW TRACHEA

Z = DISTANCE FROM THYROID CARTILAGE (CM)

Fig. 5.18  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF EXPIRATION

EXP. 17  V=142 CM/SEC  RN=1763
Fig. 5.19  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF INSPIRATION
EXP. 18  V=217 CM/SEC  RN=2478
Fig. 5.20  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 50% OF INSPIRATION
EXP. 18  V=307 CH/SEC  RN=4213
NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF INSPIRATION
EXP. 18  v=153 cm/sec  RN=7099
Fig. 5.22 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF EXPIRATION

EXP. 18 V=561 CM/SEC RN=7699
Fig. 5.23  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 50% OF EXPIRATION
EXP. 18  V=250 CM/SEC  RN=3431
Fig. 5.24  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF EXPIRATION
EXP. 18  V=70 CF/SEC  RN=966
Fig. 5.25  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF INSPIRATION

EXP. 19  V=200 CM/SEC  RN=2091
Fig. 5.26  
NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 50% OF INSPIRATION  
EXP. 19  V=270 CM/SEC  RN=2823
SAGGITAL VIEW TRACHEA

VENTRAL VIEW TRACHEA

Fig. 5.27 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF INSPIRATION
EXP. 19 V=171 CM/SEC RN=1780
Fig. 5.26 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF EXPIRATION
EXP.19 V=400 CM/SEC RN=4183
Fig. 5.29 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 50% OF EXPIRATION
EXP. 19 V=304 CM/SEC RN=3179

Z = DISTANCE FROM THYROID CARTILAGE (CM)

SAGITTAL VIEW TRACHEA

VENTRAL VIEW TRACHEA
SAGITTAL VIEW TRACHEA

VENTRAL VIEW TRACHEA

Z = DISTANCE FROM THYROID CARTILAGE (CM)

Fig. 5.30 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF EXPIRATION
EXP. 19 V=120 C.M/SEC RN=1254
Fig. 5.31
NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF INSPIRATION
EXP. 28 V=68 CM/SEC RN=535

Z = DISTANCE FROM THYROID CARTILAGE (CM)

RADIAL POSITION R (CM)

SAGITTAL VIEW TRACHEA

VENTRAL VIEW TRACHEA
Fig. 5.32  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 50% OF INSPIRATION

EXP. 28  \( V = 181 \text{ cm/sec} \)  RN=1735
Fig. 5.33 NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF INSPIRATION

EXP. 28 V=185 CM/SEC RN=1445
Fig. 5.34  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 25% OF EXPIRATION

EXP. 28  V=429 CM/SEC RN=4424

Z = DISTANCE FROM THYROID CARTILAGE (CM)
Fig. 5.35 Normalized velocity profiles measured in canine trachea at 50% of expiration.

Exp. 28 V=238 cm/sec RN=2454
Fig. 5.36  NORMALIZED VELOCITY PROFILES MEASURED IN CANINE TRACHEA AT 75% OF EXPIRATION
EXP. 28  V=95 CM/SEC  RE=975

Z = DISTANCE FROM THYROID CARTILAGE (CM)
Fig. 5.37 Time-dependent spectral display of velocity waveform measured at trachea centerline location 3 cm from the thyroid cartilage, experiment #12
Fig. 5.38  Time-dependent spectral display of velocity waveform measured at trachea centerline location 7 cm from the thyroid cartilage, experiment #12
Fig. 5.39  Time-dependent spectral display of velocity waveform measured at trachea centerline location 13 cm from the thyroid cartilage, experiment #12
Fig. 5.40 Time-dependent spectral display of velocity waveform measured at trachea centerline location 2 cm from the thyroid cartilage, experiment #13
Fig. 5.41  Time-dependent spectral display of velocity waveform measured at trachea centerline location 7 cm from the thyroid cartilage, experiment #13
Fig. 5.42 Time-dependent spectral display of velocity waveform measured at trachea centerline location 6 cm from the thyroid cartilage, experiment #13
Fig. 5.43 Time-dependent spectral display of velocity waveform measured at trachea centerline location 10 cm from the thyroid cartilage, experiment #13
Fig. 5.44 Time-dependent spectral display of velocity waveform measured at trachea centerline location 2 cm from the thyroid cartilage, experiment #17
Fig. 5.45 Time-dependent spectral display of velocity waveform measured at trachea centerline location 5 cm from the thyroid cartilage, experiment #17
Fig. 5.46  Time-dependent spectral display of velocity waveform measured at trachea centerline location 8 cm from the thyroid cartilage, experiment #17
Fig. 5.47 Time-dependent spectral display of velocity waveform measured at trachea centerline location 3 cm from the thyroid cartilage, experiment #18
Fig. 5.48  Time-dependent spectral display of velocity waveform measured at trachea centerline location 5 cm from the thyroid cartilage, experiment #18
Fig. 5.49 Time-dependent spectral display of velocity waveform measured at trachea centerline location 8 cm from the thyroid cartilage, experiment #18
Fig. 5.50  Time-dependent spectral display of velocity waveform measured at trachea centerline location 12 cm from the thyroid cartilage, experiment #18
Fig. 5.51

Time-dependent spectrum of a zero velocity waveform demonstrating noise level.
Fig. 5.52 Time-dependent spectral display of velocity waveform measured in the 0th generation by retrograde catheter in experiment #29
Fig. 5.53  Time-dependent spectral display of velocity waveform measured in the 1st generation by retrograde catheter in experiment #29
Fig. 5.54  Time-dependent spectral display of velocity waveform measured in the 2nd generation by retrograde catheter in experiment #29.
Fig. 5.55  Time-dependent spectral display of velocity waveform measured in the 4th generation by retrograde catheter in experiment #29.
Fig. 5.56 A composite of velocity waveforms measured during experiment #29. From top to bottom, the 0th, 2nd, 3rd and 4th generations, respectively.
Fig. 5.57 Time-dependent spectral display of velocity waveform measured in the 0th generation by retrograde catheter in experiment #30.
Fig. 5.58  Time-dependent spectral display of velocity waveform measured in the 1st generation by retrograde catheter in experiment #30.
Fig. 5.59 Time-dependent spectral display of velocity waveform measured in the 2nd generation by retrograde catheter in experiment #30.
Fig. 5.60 Time-dependent spectral display of velocity waveform measured in the 3rd generation by retrograde catheter in experiment #30.
Fig. 5.61 A composite of velocity waveforms measured during experiment #30. From top to bottom the 0th, 1st, 2nd and 3rd generations, respectively.
Fig. 5.6.2 Time-dependent spectral display of sound waveforms recorded during experiment #12 at a location 2 cm from the thyroid cartilage.
Fig. 5.63 Time-dependent spectral display of sound waveforms recorded during experiment #12 at a location 7 cm from the thyroid cartilage.
Fig. 5.64 Time-dependent spectral display of sound waveforms recorded during experiment #12 at a location 3 cm from the thyroid cartilage.
Fig. 5.65 Time-dependent spectral display of sound waveforms recorded during experiment #13 at a location 2 cm from the thyroid cartilage.
Fig. 5.66  Time-dependent spectral display of sound waveforms recorded during experiment #13 at a location 6 cm from the thyroid cartilage.
Fig. 5.67 Time-dependent spectral display of sound waveforms recorded during experiment #13 at a location 10 cm from the thyroid cartilage.
Time-dependent spectral display of sound waveforms recorded during experiment #28 at a location 2 cm from the thyroid cartilage.
Fig. 5.6.9 Time-dependent spectral display of sound waveforms recorded during experiment #28 at a location 7 cm from the thyroid cartilage.
Fig. 5.70 Time-dependent spectral display of sound waveforms recorded during experiment #28 at a location 11 cm from the thyroid cartilage.
Fig. 5.71 Time synchronized sound and flow signals illustrating the correlation between the onset of sound and turbulence in the expiratory flow pattern.
CHAPTER VI

DISCUSSION OF RESULTS

6.1 Introduction

It is the author's intent in this discussion to interpret the many aspects of the results and to compare the results obtained with the findings of other researchers who have studied airways aerodynamics. From this comparison, it is hoped to provide, as completely as possible, a description of the airflow pattern in canine upper airways. Tracheal flow patterns, upper airways flow patterns, and lung sounds all will be reviewed. However, the majority of data comprising this effort consists of tracheal airflow patterns, and hence, this subject suggests itself as the first topic of discussion.

The results of this investigation have demonstrated tracheal flow patterns in the canine to be highly asymmetric, time-dependent and characteristically turbulent.
Secondary flows and possible regions of flow separation were also suggested by the time resolved velocity profiles measured in this study. Velocity profiles did not change significantly between the three measurement times during inspiration or expiration. Although in general similar, the inspiratory and expiratory phases of flow did exhibit some differences and will therefore be treated separately. Since the inspiratory phase has been investigated in the literature in much greater detail than the expiratory phase, inspiratory flow patterns will be discussed first.

6.2 Tracheal Flow - Inspiration

Perhaps one of the most striking characteristics of the inspiratory flow patterns measured in this experiment is the asymmetry of the flow. Figures 5.31 through 5.36 and 5.19 through 5.24 clearly demonstrate this aspect of the inspiratory flow measurements. Such flow asymmetry is considered to be the result of local geometric influences. In these samples, the degree of skewing is not constant from experiment-to-experiment suggesting that entrance conditions were highly variable and of strong influence. This is quite likely as the degree of anesthesia, animal size, and upper airways anatomy all
varied from subject-to-subject. The predominant picture of these skewed profiles, however, is one in which the highest velocities were measured in the right dorsal aspect of the trachea in the plane nearest the tracheal entrance. These profiles contained peak flows that exceeded the minimum velocities in a given plane by as much as 500 per cent. This is strongly indicative of a jet formed at the entrance of the trachea.

Such flow patterns are not unique to this investigation, but are described in many investigations related to lung aerodynamics. Iliff, Olson, and Sudlow (29) used hot-wire anemometry to investigate velocity patterns between the larynx and lobar bronchi in a cast replica of the human upper and central airways. They found that the larynx strongly influences flow patterns in airways immediately downstream (the trachea and main bronchi). Their investigation also revealed that the asymmetry of the flow (i.e., high velocities close to the posterior wall) strongly reflected the local geometry, in this instance the V-shaped aperture of the larynx. Dekker (18) also used cast models of the human airways, including the complete larynx and main bronchi, in order to investigate the transition between laminar and turbulent flow in the trachea. Although the turbulent nature
of the flow will be dealt with in another section, Dekker's results indicate that the local geometry plays an important part in the transition to turbulence. He concluded that during inspiration the influence of the position of the glottis and aditus laryngis on tracheal turbulence is very great.

Olson, Sudlow, Horsfield and Filley (10) studied in great detail the convective patterns of flow during inspiration. These measurements were performed in full scale replicas of the human pulmonary system, consisting of the airways from the mouth to the subsegmental bronchi. Velocities were obtained by hot-wire anemometers and impact pitot tubes. These investigators found that in the vicinity of the larynx reverse flow regions were encountered as well as maximum velocities toward the posterior aspect of the cricoid cartilage. Olson et al. (10) also found a higher velocity on the posterior or membranous portion of the trachea. They attributed this result to the aperture of the larynx being wider posteriorly than anteriorly and thus allowing more flow to extrude through the larynx in the posterior area. These anatomical characteristics are indicated to have a strong influence on flow by Olson et al. (10).
The results of the present investigation certainly support the findings of previous studies in pulmonary aerodynamic studies. Although the studies mentioned were carried out in casts with non-variable dimensions, good agreement of a qualitative nature exists between this in vivo work and these earlier studies. However, because such a strong influence of geometry is seen to exist for tracheal flow patterns, it is reasonable to assume that changes in inlet conditions during the breathing cycle in vivo would provide different quantitative results. Such time variations associated with respiration in local anatomy should therefore be considered in future studies.

The strong asymmetries discussed above may very well induce some of the disturbances associated with the turbulence measured during the inspiratory phase of flow in our investigations. The results of this investigation indicate that flow in the trachea during inspiration is turbulent in nature \((600 < Rn < 4200)\), containing velocity fluctuations as high as 2000 Hz. The relative intensity and frequency distribution of these fluctuations do not change with location along the trachea. A brief analysis of a single spectrum representing an entire waveform defined an amplitude peak at about 300-400 Hz. The peak flow during this inspiratory cycle was 284 cm/sec and
the diameter of the trachea was 1.3 cm (Rn=2366). The corresponding estimated center frequency U/L is predicted to be on the order of 200 Hz, which is a result of the correct magnitude considering possible flow transitions during the complete cycle.

The fluid mechanical regime occurring in the airways has long been the center of interest in many investigations. An early yet highly illustrative work was that of Dekker (18). In this investigation casts of the human trachea, larynx, and main bronchi were studied with dye-tracer flow visualization techniques as well as some measurements with a hot-wire anemometer. The critical volume flow rate on inspiration was 47 ml/sec. Based upon the assumption of a 2 cm diameter trachea, this value is comparable to a flow velocity of approximately 17 cm/sec (Rn=220). Such velocities are characteristically low compared with the velocities obtained during the present investigation. Dekker noted that the flow disturbing influence of the narrowing and subsequent enlargement which occurred in the glottal part of the airways was clearly visible. During rapid inspiratory flow a reversal of the turbulent flow could be observed along the walls at the abrupt tracheal enlargement beyond the glottis. These findings led Dekker to indicate that airflow in the trachea
of most individuals is turbulent during the greater part of normal inspiration. This finding is in clear agreement with the study under discussion.

Fluctuations superimposed on the mean velocity were detected at all the inspiratory flow rates studied (575 Rn 8214) by Iliff, Olson, and Sudlow (29). This work carried out in cast models of the human bronchial tree. In a later and extremely detailed study, Olson et al. (10) also discussed instabilities in tracheal airflow patterns. In these investigations a flow rate of 200 ml/sec was sufficient to induce a fluctuating velocity component (Rn=800) in the central airways. Olson suggests that turbulence is produced by the flow undergoing separation just below the larynx.

Owen (30) analyzed, theoretically, the nature of flow in the trachea and central airways with respect to particle deposition. This analysis presumed the flow to be detached as it moved through the upper airways constrictions. This pattern was then considered to be responsible for jet formation in this region and the resulting jet to be the site of turbulence generation (Rn=3100). Owen suggested that turbulence spread both inwards across the jet and outwards to the wall of the trachea. Because
the boundary layer was influenced by the turbulent, sheared, jet-like flow, the boundary layer grew more rapidly than flows with a uniform central stream. This result suggested that the flow at the lower end of the trachea was close to being fully developed. Many of the profiles measured in this study appeared to be fully developed during inspiration. These results tend to confirm the theoretical ideas of Owen.

The asymmetric jet-like profiles encountered during inspiration in this investigation also served as markers in the flow. These high velocity regions were identified as they moved from the entrance of the trachea to the exit. Naturally, because of the turbulent mixing occurring within the flow, the persistence of the jet was variable but could often be identified for several diameters along the flow path. The movement of these high velocity regions within the flow field were suggestive of secondary motions, in particular a spiraling of the flow along the trachea. Because the hot-wire anemometers used do not possess a directional capability, any description of these secondary flows is limited.

This spiraling flow pattern is indicated in Figures 5.31, 5.32 and 5.33. Location A in these figures identifies the peak inspiratory velocities near the
right lateral wall 2 cm from the thyroid cartilage. At 7 cm from the thyroid cartilage in the ventral view (location B), a more blunt profile is shown. Location C, 9 cm from the thyroid cartilage, illustrates a flow pattern where the peak velocities are near the left lateral wall. The sagittal view of Figure 5.33 also suggested this type of flow pattern. Figures 5.19, 5.20 and 5.21 also indicate this characteristic spiral. The source of such a secondary flow is likely to be the curved nature of the trachea in the supine dog. Several researchers have identified similar patterns in related work.

Olson et al. (10) identified a secondary flow with this character during inspiration in the human trachea. These investigators attributed this flow rotation to the aortic curvature of the trachea. Although the source of curvature is different there than the one in this investigation, the result appears to be similar. Dekker (18) does not mention the spiraling nature of inspiratory flow in his models, although a photograph from his work of a laminar flow situation indicates the flow does rotate. Dekker does indicate that a spiral pattern exists in expiratory flow. Olson (12) also investigated the nature of flow in a curved tube. His results were not amenable to generalization, but did in some instances give rise to flow profiles where
peak velocities were seen to shift from one side of the tube to the other as the flow progressed along the curved section. The interpretation of our qualitative observations of secondary motions could also be interpreted as oscillating in this manner.

It is likely that our measurements of secondary flow rotation do reflect the situation within the trachea, especially in light of confirming evidence from outside sources. The diffuse nature of these observations is not surprising when considered in light of the many interacting variables encountered in vivo.

A review of the time-dependence of inspiratory profiles indicated that the flow patterns behaved in a quasi-steady fashion. No characteristic changes in profiles could be associated with the timing of events within the inspiratory cycle. This is reasonable for a number of reasons. The calculation and interpretation of Womersley's (21) unsteadiness parameter \( \alpha \) indicates that the flow should approach a quasi-steady pattern. This result is based upon a sinusoidal flow waveform. However, the inspiratory flow waveforms encountered in these studies were not sinusoidal, but more flat. This type of waveform would induce profile changes very early and very late in the flow but little acceleration would be seen
during the majority of the flow cycle. Since profiles were measured only at 25%, 50% and 75% of the cycle, profiles might be expected to change very little between these measurement profiles.

Olson et al. (10) observed profiles in the trachea to become flatter at high flow rates and more peaked at low rates. No change of this nature was observed in the present study. This difference is most likely explained by the differing peak flow rates. Flow rates in the present study were those of normal tidal breathing in a resting dog. Olson's peak flow measurement was at values of 150 liters/minute, a value far in excess of that encountered in this study.

Many investigators, including Schroter and Sudlow (13), Pedley, Schroter and Sudlow (6), Iliff, Olson and Sudlow (29), Dekker (18), Flescher and Bridge (16), Sekehara, Olson and Filley (31) have employed the quasi-steady flow assumption in their model studies within the lung. The results of the present study tend to confirm the validity of this assumption at tidal breathing rates. This is a strong validation of this assumption since it appears to be true even under the considerable influences to be encountered in the in vivo state. This is the first work to confirm this suggestion in vivo.
6.3 Tracheal Flow - Expiration

The observed nature of expiratory flow was also both interesting and informative. Turbulence, asymmetries, time variation, and secondary flows were observed during this phase of respiration just as they were during inspiration. The particular aspects of each of these phenomena were often considerably different than those discussed earlier. Such differences were to be expected, however, since upstream conditions and velocity waveforms were totally different during this phase of respiration.

Perhaps the most exciting difference between inspiration and expiration was the nature of turbulence onset. During inspiration, the velocity waveform was almost always disturbed. The velocity waveform during expiration however was characteristically undisturbed during the initial phase of flow acceleration. This undisturbed velocity amounted to about 10% of the total expiratory period, usually about 0.1 sec to 0.2 sec. This quiescent flow was maintained until the flow velocity approached peak expiratory velocity (3000<Rn<7700), at least in most cases, and the initiation of disturbed flow in general corresponded to the beginning of
expiratory velocity deceleration. The onset of turbulence was very abrupt during expiration, whereas the turbulence intensity developed gradually with inspiratory flow. Because of the repeatability of most respiratory cycles, the differentiation of time to turbulence, critical velocity for turbulence, and the nature (acceleration or deceleration) of flow at turbulence initiation is not completely possible. The analysis of velocity waveforms with differing peak expiratory velocities suggests, however, that the onset of turbulence is not governed by reaching a critical velocity. This conclusion is based upon the observation that the flow will not become unsteady at a fixed velocity in a given system, but rather at a point where the expiration began to slow. It is possible that, as the flow begins its deceleration phase, the velocity profile may manifest an inflection point leading to the triggering of instabilities. These inflections, however, were not observed with the time resolutions employed in our analysis.

No research efforts were found that dealt directly with dynamic expiratory flows within the lung. Several authors did review constant-flow models of expiration. West and Hugh-Jones (32) measured critical expiratory flow velocity in a cast of the lower half of the trachea.
and the first branches of the segmental bronchi. Water was used as the flowing medium. The critical flow rate was found to be 220 ml/sec. With a tracheal diameter of 0.8 cm, this value corresponds to a velocity of 86 cm/sec, and a Reynolds number of 1000. During the acceleration phase of expiration, velocities as great as 300 cm/sec were not seen to be turbulent in our study, although velocities of 100 cm/sec were observed to be highly disturbed during the deceleration phase of expiration.

Dekker (18) also studied the critical flow velocity necessary for turbulent flow generation in casts. His steady state expiratory flow models using both air and water yielded values of 122 and 127 ml/sec, respectively. These values correspond to a flat profile value of about 50 cm/sec and a Reynolds number of 600. Normal tidal breathing in the anesthetized dog produced values of Reynolds number generally greater than these values during all but very limited periods during the flow cycle. No decay of turbulence with time was seen in the velocity waveforms measured during expiration. This is perhaps because of the high Reynolds number (1000 < Rn < 7700) prevailing throughout most of the flow period. Dekker's results confirm these in that once the flow becomes turbulent it remains so for the duration of the cycle. The unsteady
nature of the flow was suggested by Dekker as possibly having a strong influence on the character of the flow. He did not indicate, however, in what way.

Another means of assessing turbulence measurements made during expiration was the relative intensity of fluctuating nature of the flow relative to the mean value of the flow. Characteristically, the intensity of centerline fluctuations during inspiration amounted to as much as 70% of the mean flow value. Expiratory flows, however, exhibited significantly reduced fluctuation intensities, corresponding to about 10% of the velocity. It seemed likely that these differences in the intensity of turbulence are attributable to the differences in entrance conditions for the two flow patterns. Owen (30) indicates that turbulent pipe flow and turbulent jet flow may develop in different manners. The smooth pipe requires ten or more diameters to become fully developed at a Reynolds number of 3000. The inspiratory cycle, however, is a different situation since flow most likely enters as a jet. Turbulence then would spread both inward across the jet and outward to the wall. Such a mechanism would lead to rapid full development of the flow and to high levels of flow fluctuation. The frequency spectra corresponding to similar mean velocities
in both phases of the respiratory cycle bear out this apparent difference in fluctuating intensity. At the same mean flow velocity, inspiratory spectra contained frequencies to about 2000 Hz, whereas during expiration only frequencies of about 1000 Hz and less were observed at this velocity. This is not to say that higher frequencies did not exist, but that at fixed gain settings the intensities were greater during inspiration.

Observations made concerning expiratory flow patterns were not limited to turbulence, but included other phenomena such as asymmetries and secondary flows. Because of the strong interaction between flow asymmetries and secondary flows during expiration, these flow phenomena will be discussed together. The anatomy of the canine lung suggests that expiratory flow entrance into the trachea will not be a uniform fully developed flow profile, but rather a pattern reflecting the branching tube geometry. For laminar flows, as an example, the entrance length would be \( L = 0.05D Rn \). Based upon this relationship, the length of the trachea required for fully developed flow would be over 300 cm. Turbulent pipe flows require at least ten diameters or more to become fully developed. In this case, more than 20 cm of trachea would
be needed. Even for jet type flow, entrance lengths of six diameters are called for, and with this type of entry very little of the trachea could possibly contain fully developed flow.

Inspection of flow patterns developed during expiration indicate that this is indeed the case. Figure 5.36 (location A) depicts a bimodal pattern characteristic of the branching at the carina. These observations, within 1 cm of the carina, were made only infrequently since they required the bronchoscope to be located deep within the trachea. These patterns seemed strongly coupled to local geometry effects. This bimodal velocity profile can be identified in several other experiments as well, but the profiles are further removed from the carina in such instances (see Figures 5.11A, B, C; 5.12A, B, C, D; 5.17A, B, C, D, E).

In Figure 5.36 in the plane of the first bifurcation, the bimodal velocity profile appears (location A), disappears (location B) and reappears (location C) as the flow moves up the trachea. A possible explanation for such behavior is that the entire flow is rotating as it moves along the trachea. This possibility is also suggested by the flow behavior in the sagittal view (locations D
and E) of Figure 5.36. Without the ability to resolve the direction of the flow or the existence of supporting evidence, such suggestions of secondary motions are open to question.

However, evidence of such spiraling secondary flows during expiration does exist from several sources. Schroter and Sudlow (13) studied expiratory flow downstream of a single bifurcation. With flat entry profiles, the developing profiles tend to remain flat in the plane normal to the bifurcation, but in the junction plane a bimodal profile first appears and then develops into a profile with a maximum in the center. These resultant profiles are explained in terms of secondary flows. The secondary motions were said to carry momentum from the core to the walls in the normal plane, thus dissipating any potential accumulation. The secondary flows in the junction plane tended to accumulate fluid in the middle of the tube. Although this mechanism does not explain fully the observed spiraling motion, it does suggest one mechanism by which the bimodal velocity pattern shifts its shape. However, when considered in conjunction with curved tube mechanisms discussed earlier, the above flow patterns could be visualized.
In a more empirical treatment, Dekker (18) studied expiratory flow patterns in a cast of the human airways. He used dye-tracer techniques to identify flow patterns in the trachea. Dekker stated that during laminar flow in the expiratory direction a spiral arrangement was often observed. He also indicated that the spiral pattern of expiratory flow persisted during beginning turbulence.

Such mechanisms as the above suggest that the pattern of expiratory flow is very complicated and likely to involve spiraling motions and profile changes. The present results confirm the existence of such patterns within the accuracy of the measurement techniques.

6.4 Upper Airways Flow

Time-dependent velocity spectra also served as the major means of assessing information resulting from the retrograde catheterization experiments discussed in section 4.5. Figures 5.51 through 5.61 document the results of the typical retrograde velocity measurement experiments.

Retrograde waveforms are presented in serial fashion beginning with measurements in the trachea and sequencing one figure for each generation below the
trachea. Figure 5.51 depicts the baseline flow signal and its concomitant frequency spectrum for subsequent comparisons. Figure 5.51 was generated from the real time waveform of the functioning hot-film anemometer exposed to a zero flow situation. This figure characterizes the typical noise level of the total system. Figures 5.52, 5.53, 5.54, and 5.55 represent waveforms taken in the trachea, first, second, and fourth generations respectively for one experiment. This sequence presents an interesting pattern of frequency content in the lower airways. In the trachea the waveform contains a characteristically uniform spectrum comparable to earlier discussed hot-wire measurements, but reduced in amplitude and frequency. In the first generation, taking into consideration the different amplitudes of the waveform, no significant change in frequency content can be identified. Disturbances persist to about 250 Hz with no dominating or identifiable peaks. In the second generation, however, the flow pattern has changed significantly and so does the velocity spectrum. The spectrum contains a resonant spike at about 50 Hz although still containing frequencies up to about 200 Hz. At the fourth generation, and with the dog under positive chest pressure ventilation, a broad frequency band can be seen. This
frequency band, from 136 Hz to 176 Hz, is the only dominant pattern of an otherwise quiescent spectrum. This last generation is suggestive of a non-turbulent flow regime possibly containing shed vortices. Figure 5.60, a fourth generation measurement from a different animal, also contains a characteristic harmonic at 97 Hz. Such information suggests that turbulence, at least in the animals studied, seems to be giving way to less disturbed flow patterns by the fourth generation, although persisting into the fifth generation in all cases. Caution must be employed, however, to not put too much quantitative significance on measurements obtained by hot-film probes under these flow conditions in animal preparations where downstream conditions are less than optimal due to the presence of retrograde catheterization paraphernalia.

Aside from frequency analysis of upper airways velocity measurements, direct velocity waveform analyses also proved rewarding. Table 6.1 indicates the peak inspiratory velocities encountered in each of the generations measured. Also presented are Reynolds number estimates based upon bronchoscope observation of airway diameter, as well as the per cent of mean velocity observed as fluctuation (i.e., turbulence intensity). Peak
<table>
<thead>
<tr>
<th>Generation</th>
<th>Peak Velocity</th>
<th>Rn</th>
<th>Turbulence Intensity</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Exp. R-1</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>160 cm/sec</td>
<td>1464</td>
<td>0.38</td>
</tr>
<tr>
<td>1</td>
<td>220 cm/sec</td>
<td>1297</td>
<td>0.33</td>
</tr>
<tr>
<td>2</td>
<td>200 cm/sec</td>
<td>915</td>
<td>0.30</td>
</tr>
<tr>
<td>4</td>
<td>120 cm/sec</td>
<td>392</td>
<td>0.13</td>
</tr>
<tr>
<td><strong>Exp. R-2</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>250 cm/sec</td>
<td>2694</td>
<td>0.40</td>
</tr>
<tr>
<td>2</td>
<td>203 cm/sec</td>
<td>1142</td>
<td>0.36</td>
</tr>
<tr>
<td>3</td>
<td>93 cm/sec</td>
<td>400</td>
<td>0.25</td>
</tr>
<tr>
<td>4</td>
<td>125 cm/sec</td>
<td>426</td>
<td>0.13</td>
</tr>
<tr>
<td><strong>Exp. R-3</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>220 cm/sec</td>
<td>2371</td>
<td>0.25</td>
</tr>
<tr>
<td>1</td>
<td>190 cm/sec</td>
<td>1482</td>
<td>0.23</td>
</tr>
<tr>
<td>2</td>
<td>140 cm/sec</td>
<td>789</td>
<td>0.22</td>
</tr>
<tr>
<td>3</td>
<td>130 cm/sec</td>
<td>561</td>
<td>0.05</td>
</tr>
<tr>
<td><strong>Exp. R-4</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>250 cm/sec</td>
<td>2700</td>
<td>0.33</td>
</tr>
<tr>
<td>1</td>
<td>125 cm/sec</td>
<td>975</td>
<td>0.21</td>
</tr>
<tr>
<td>2</td>
<td>100 cm/sec</td>
<td>564</td>
<td>0.10</td>
</tr>
<tr>
<td>4</td>
<td>100 cm/sec</td>
<td>432</td>
<td>0.05</td>
</tr>
<tr>
<td><strong>Exp. R-5</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>205 cm/sec</td>
<td>2214</td>
<td>0.23</td>
</tr>
<tr>
<td>1</td>
<td>205 cm/sec</td>
<td>1600</td>
<td>0.18</td>
</tr>
<tr>
<td>2</td>
<td>175 cm/sec</td>
<td>987</td>
<td>0.09</td>
</tr>
<tr>
<td>3</td>
<td>100 cm/sec</td>
<td>432</td>
<td>0.09</td>
</tr>
</tbody>
</table>
velocities remained remarkably constant in each of the first four generations measured. This reflects the characteristic total area curves of the canine lung. Reynolds numbers fell with diameter of the tubes and turbulence intensity appeared reduced by at least a factor of two by the fourth generation.

Other investigators have studied, at least in a limited sense, the nature of flow in the upper airways. Sikehara, Olson and Filley (31) measured velocity profiles in casts of primary through segmental bronchi during steady state inspiratory flows. Using pitot tube probes, these investigators found that disturbed flow consistently disappeared at Reynolds numbers values less than 2000 and seemingly independent of location within the cast. At flow rates great enough to produce local Reynolds numbers greater than 2000, the flow was found to be highly disturbed even in the bronchi. These results are not in agreement with the present study. Our results for both trachea and upper generations (1-4) have demonstrated highly disturbed flow at Reynolds numbers an order of magnitude less than those measured in the above study.

Olson, Sudlow, Horsfield and Filley (10) used hot-wire anemometers to measure velocities in casts of humans. These investigators found fluctuating velocity components
at all flow rates about 200 ml/sec. It was surprising to them that such fluctuations existed to levels of the sublobar bronchi where the Reynolds number was below 100. These authors explained this result by suggesting that the scale of turbulence in the trachea is very large and that time is inadequate for breakdown to occur within the trachea. The larger eddies are rather swept down to branches of inadequate diameter to accommodate them. Subsequently these eddies impact the wall of these tubes and break up into small eddies rapidly dissipated by viscous effects.

Owen (30) presented an analysis of turbulence decay for inspiratory flow within the lung. The form of decay law followed by the turbulence may be inferred from an approximate energy equation in which the rate of change of turbulent energy is equated to the difference between the rate of production of energy and its rate of dissipation. Accordingly, if $q^2$ is a typical turbulent fluctuation in mean square, with components $u_1^{1/2}$, $u_2^{1/2}$ and $u_3^{1/2}$ in the axial ($x$), radial ($r$), and circumferential directions respectively, the energy equation becomes

$$\frac{1}{2} \frac{\partial q^2}{\partial t} = u_1 u_2 \frac{\partial u}{\partial r} - \epsilon$$  \hspace{1cm} (21)
where $e$ is the rate of energy dissipation per unit mass of fluid and $U$ is the axial velocity in the mean flow.

Except very close to the wall, an order of magnitude estimate of $\partial U/\partial r$ gives $\partial U/\partial r = u_T/a$, where $a$ is the radius of the pipe and $u_T$ is the friction velocity defined by $\rho u_T^2 = \tau_0$, in which $\rho$ is the density of the fluid and $\tau_0$ the shearing stress it exerts on the wall.

It may be assumed that

$$u_1 u_2 \sim a_0 \rho q^2$$

(22)

$a_0$ being a constant of order $10^{-1}$.

In order to estimate the magnitude of the dissipation term it may be supposed that the small turbulent eddies responsible for the ultimate degradation of the kinetic energy are locally isotropic, in which case

$$e = 15 \nu (\partial u_1^2/\partial x)^2 - a_2 \nu q^2/a^2$$

(23)

where $\nu$ is the kinematic viscosity of the fluid and $a_2$ is a constant of order 10.

In casting the dissipation term into the above form we have made the crucial assumption that in the decay process the energetic turbulent eddies are bounded in size by the walls of the pipe and therefore exhibit
a typical dimension comparable with the pipe radius. In consequence, the decay may be expected to follow in time, or distance, a different law from that obeyed by an unbounded field of decaying homogeneous turbulence in which the length scale of the energetic eddies increases as the decay proceeds.

If the turbulence is convected with the mean flow,

$$\frac{\partial \omega}{\partial t} = \frac{\partial U \omega}{\partial x}$$

(24)

where $U$ is the mean axial velocity averaged across the pipe. Consequently, the energy equation may be written

$$\frac{d\omega}{d\xi} = 2a_1 \omega - 4a_2 \omega / R_n$$

(25)

where $\omega = q / U$, $\xi = x / d$, $d$ is the pipe diameter and $R_n$ is the Reynolds number, $Ud / v$; $a_1 = a_0 (u_t / U)$ and is of the order $10^{-2}$.

(25) is integrated to give

$$\frac{\omega}{\omega_0} = \exp \left\{ (2a_1 - 4a_2 / R_n)(\xi - \xi_0) \right\}$$

(26)
in which $\omega_0$ is the initial value of $\omega$ at $\xi = \xi_0$. 
Owen applied this equation to normal lung flow and estimated that the turbulence intensity would only be diminished by one-half at the third generation. The equation indicates that as the Reynolds number decreases, however, decay becomes greatly enhanced. Owen concluded that if turbulence is generated in the trachea, its subsequent decay in the bronchi is initially slow. These results fortify our own observations in the upper four generations.

Using Owen’s (30) formulation, present experimental results on the decay of turbulence within the first four generations of the lung can be analyzed. Figure 6.1 shows the results of five experiments in which inspiratory velocity waveforms were measured in the upper airways. These experimental results agree quite well with the experimental results of Sibulkin (33) for pipe flow and Narayanan (34) for channel flow, as well as with the experimental law of decay suggested by Owen. Sibulkin et al. (33) estimated that $A_1=0.001$ and $A_2=17.9$ for a curve fit of their experimental data and using equation as the model. A linear curve fit based upon a least squares method was applied to the data points shown in Figure 6.1. This curve fit yielded the
Fig. 6.1 Decay of turbulence within the lung at subcritical Reynolds numbers during inspiration.
following equation for turbulence decay within the upper airways at subcritical Reynolds number

\[ \log_{10}(\omega/\omega_0) = 0.00057 - 30.79/R_n \]

This result which conforms well with other available data and suggests an initially slow decay of turbulence in the bronchi, but a rapid increase in decay as Reynolds falls in the smaller lumen.

6.5 Tracheal Sound

Time-dependent sound spectra were developed from sound signals recorded within the trachea by microphone techniques as discussed in an earlier Chapter. Figures 5.62 through 5.70 represent three typical experiments presented in aerial fashion in terms of location within the trachea. The sound spectra bear a remarkable similarity to the velocity spectra encountered within the trachea during expiration. The maximum observable frequencies at our gain settings were about 2000 Hz. The spectra do not seem to contain any resonant spikes, but rather a broad uniform distribution of amplitude with frequency. Since the microphones employed demonstrated flat frequency response up to well above the 2000 Hz level, the resultant spectra are likely to be a realistic representation of pressure fluctuations within the trachea. The
sound intensity falls rapidly as does the flow during expiration. The sound intensity during inspiration is many times reduced when compared with the sound generated during expiration. However, the peak velocity amplitude is characteristically only reduced by about one-half during inspiration relative to expiration. Such relationships indicate a strong dependence on velocity relative to sound. This relationship is reflected in the reduced overall amplitude of the sound spectra during inspiration. This relationship is suggestive of the sound radiated from a turbulent jet where the acoustic efficiency is strongly dependent on velocity, resulting in acoustic power governed by velocity to the 8th power.

Careful examination of the velocity and sound signals on an expanded time base (Fig. 5.71) reveals that the onset of sound in the trachea is clearly associated with the development of disturbed flow within the trachea during expiration. Such an association suggests that it is the turbulence induced pressure fluctuations in the flow that give rise to the breath sounds perceived in close proximity to the trachea. McDonald (35), McKusick (36) have discussed the origin of cardiovascular sounds generated by turbulence in the arteries, and indicate that the turbulence gives rise to pressure
fluctuations perceived as sound.

This relationship between turbulent flow and sound production has been proposed before. Forgacs, Nathoo and Richardson (37) stated that the noise of the upper respiratory tract, the trachea, and the central bronchi was generated by turbulent flow of air. An evenly spread frequency distribution between 200 and 2000 Hz was generated by recordings made close to the mouth. These investigators found a linear correlation between sound intensity and inspiratory flow rate.

Banaszak, Kory and Snider (9) also measured sound from the lung field. These researchers believed that breath sounds were generated at the larynx and carina of larger bronchi by turbulent air flow and then selectively filtered by the components of the lung. Since Banaszak's measurements were made on the dorsal aspect of the lower thorax, little correlation of his relative sound intensity measurements can be made in the present study.
CHAPTER VII

CONCLUSION AND REMARKS

7.1 Introduction

The primary objective of this investigation has been to determine, in vivo, the nature of air flow in the upper airways of canines. The present effort has been successful in detailing several aspects of upper airways flow patterns previously unattained in vivo. Velocity profiles describing flow in the trachea in two planes have been presented and discussed. The degree of flow disturbance has been measured for both the trachea and several subsequent generations. A relationship has been determined based on these observations that describes the decay of turbulence within the lung during inspiration.

A secondary goal of this investigation was to relate velocity waveforms to sounds recorded within the lung. To this end little success was achieved. This
aspect will be discussed further in the section describing limitations of the present work.

7.2 Conclusions

Several conclusions were reached as a result of this investigation into airways aerodynamics. The analysis of tracheal velocity profiles during inspiration provided a number of conclusions not previously demonstrated \textit{in vivo}, as follows.

Tracheal velocity patterns during inspiration exhibited characteristic asymmetries as a result of local geometry. The above conclusion suggests that model studies must incorporate local geometries in detail if accurate descriptions of the flow pattern are to be properly evaluated.

A second conclusion based upon inspiratory velocity profiles is that a quasi-steady flow model should be adequate to depict the nature of the flow within the trachea. This \textit{in vivo} work confirms the assumption of quasi-steady flow in models.

As a result of the experimental observations, one also must conclude that the inspiratory flow in the trachea is highly disturbed throughout its phases.
In combination, then, a rather complete picture of inspiration in the trachea emerges as being a flow pattern that is highly disturbed with jet-like asymmetries and secondary motions that are not strongly time-dependent. This same flow approaches a fully-developed turbulent flow as it moves along the trachea.

Conclusions can also be drawn concerning flow within the trachea during expiration. Notable in its difference to inspiration, the onset of turbulence during expiration is not immediate, but is triggered by the beginning of flow deceleration. This delayed onset of turbulence can most likely be attributed to the relatively smooth entrance conditions encountered during expiration.

The expiratory phase of flow in the trachea also supports the conclusion that local geometries play a strong role in the character of the flow within the lung. This conclusion is based upon the strong secondary motions and asymmetries seen during this phase of flow.

Measurements in the first five generations of airways have provided the basis for another conclusion concerning inspiratory lung flow. The decay of turbulence in the lung at subcritical Reynolds numbers can be described as an exponential similar to that employed for pipe and channel flows. The conclusion, based on the data gathered
during this investigation, is that turbulence decay can be described by the equation

\[ \log_{10}(\omega/\omega_0) = \frac{0.0057 - 30.79/Rn}{\xi - \xi_0} \]  

(27)

Results from the sound measurements made during this investigation provided some limited insight into the relationship of flow and sound in the lung. Qualitatively, at least, it can be concluded that sounds recorded near or in the trachea were the result of pressure fluctuations generated by turbulent flow within the trachea.

7.3 Limitations of the Study

This investigation contains many limitations that reflect upon the credibility and accuracy of the data presented. With regard to the tracheal velocity profiles, several important influences must be considered. The development of velocity profiles is based upon individual waveforms taken at the different locations in the trachea at different times, often minutes apart. Complete consistency cannot be expected under such conditions, although volume waveforms were matched to reduce any possible error.

The inability of the constant temperature anemometer to sense direction also imparts significant restrictions on the interpretation of the data presented here.
One must assume, using this device, that flow is occurring mostly in the axial direction, an assumption that may not be true in light of the suggested secondary motion.

Hot-film probes did not demonstrate the same sensitivity as hot-wire probes to air flows in all instances. These film probes add an additional degree of uncertainty to measurements obtained when they were employed.

Another limitation in the evaluation of the data was the number of time segments that could be realistically analyzed. A smaller time resolution may have yielded more information but proved to be too time consuming.

Probe location was also susceptible to a degree of inaccuracy because of the basic nature of in vivo measurements. This was particularly true of upper airways measurements where tube dimensions were small and constant movement interfered with the stability of the sensor.

Sound signals also contained possible sources of error. Air-coupled microphones connected directly into the trachea measured pressure fluctuations. These signals could be the direct result of pseudo-sound (pressure fluctuations traveling at the speed of the flow) as well as
true sounds occurring within the trachea. The piezoelectric microphones placed in the esophagus are particularly sensitive to location and stable contact. These microphones were particularly difficult to use in vivo.

There are obviously many sources of error in this investigation, but in every instance every effort was taken to minimize the effect of these problems. The greatest hindrance to obtaining accurate quantitative data was the environment of the living system itself.

7.4 Future Work

Before work can progress beyond the first several generations within the canine lung studied here, numerous problems must be resolved. The probes must be more sensitive to slower flows, smaller in actual dimension and more easily placed within the lung. They must also be less sensitive to failing. Airways geometry must also be more accurately assessed.

In light of the difficulties encountered in the larger airways and the anticipated increase in hardships caused by the smaller more numerous measurement sites in the lower generations, the author feels that continued in vivo measurements of this type would prove less than fruitful until further steps are taken to improve the measurement procedures and devices.
BIBLIOGRAPHY


