In presenting the dissertation as a partial fulfillment of the requirements for an advanced degree from the Georgia Institute of Technology, I agree that the Library of the Institute shall make it available for inspection and circulation in accordance with its regulations governing materials of this type. I agree that permission to copy from, or to publish from, this dissertation may be granted by the professor under whose direction it was written, or, in his absence, by the Dean of the Graduate Division when such copying or publication is solely for scholarly purposes and does not involve potential financial gain. It is understood that any copying from, or publication of, this dissertation which involves potential financial gain will not be allowed without written permission.

7/25/68
AN ELECTROMYOGRAPHIC STUDY OF THE ELBOW JOINT

A THESIS
Presented to
The Faculty of the Division of Graduate Studies and Research
By
Mary Jaquelyn Kirkpatrick

In Partial Fulfillment
of the Requirements for the Degree
Master of Science
in Mechanical Engineering

Georgia Institute of Technology
June, 1972
AN ELECTROMYOGRAPHIC STUDY OF THE ELBOW JOINT

Approved:

Date approved by Chairman: 6/21/72
ACKNOWLEDGMENTS

This interdisciplinary thesis study involved the efforts of many persons in addition to the author. Individuals at the Emory Regional Rehabilitation Research and Training Center and the Georgia Institute of Technology worked with me during equipment assembly and experimental operation. Acknowledgment is given to Mr. R. C. Grim and Mr. H. J. Carr who built the Load Device; to Mr. Jim Perry who made the electrodes and photographed the equipment; to Mrs. Gail Super and Mrs. Mary Hill who assisted during the experiments; to Mrs. Eleanor Regenos, the Coordinator of Laboratories, whose advice and assistance helped speed completion of the experimental phase; to Mr. Charles Clayton for help with the electronics; to Mr. Glenn Shine who not only developed all the electronic equipment but also acted as a technical sounding board throughout the study from equipment design to thesis writing; to Dr. M. S. Scharf who did electrode insertions for the first eight experiments and assisted in defining the anatomical model; to Mr. Steve Wolf who did electrode insertions for the last four experiments and acted as an anatomical consultant; and to Mr. J. R. Reeves at the Rich Electronic Computer Center.

The always willing experimental subjects were Mr. Richard Hutchinson, Mr. William Pugh, Mr. Samy Marcos, Dr. Alex Alkidas, Mr. Richard Hess, Mr. David Smith, Mr. C. Runkle Jameson, Mr. Richard Sheppard, Dr. M. S. Scharf, Mrs. Angie Sheppard, and Mr. C. S. Kirkpatrick. Each subject was personally recruited and their
participation was motivated by friendship and interest in the project.

I thank Dr. W. D. McLeod for acting as thesis and academic advisor and for providing initial contact with the Emory Rehabilitation Research and Training Center. Dr. Jay H. Schlag, Dr. W. M. Williams, and Dr. J. V. Basmajian completed the Thesis Reading Committee. All three men were most helpful in their comments and suggestions. Special appreciation is extended to Dr. Basmajian who arranged for my use of the experimental equipment, coordinated technical assistance, arranged funds to pay the subjects and personally contributed invaluable anatomical input.

Dr. S. P. Kezios and Dr. P. Durbetski arranged for me the NDEA Fellowship Funds. Without this support, full-time attendance at Georgia Tech would have been impossible.

My husband, Stony, helped design the Load Device, served as a subject, proof read the thesis as it was being written and was a ready source of advice and consultation. I thank him for those inputs. However, I appreciate most his encouragement and complete willingness for me to pursue my interests and goals.
# TABLE OF CONTENTS

<table>
<thead>
<tr>
<th>Chapter</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ACKNOWLEDGMENTS</td>
<td>ii</td>
</tr>
<tr>
<td></td>
<td>LIST OF TABLES</td>
<td>vi</td>
</tr>
<tr>
<td></td>
<td>LIST OF ILLUSTRATIONS</td>
<td>vii</td>
</tr>
<tr>
<td></td>
<td>SUMMARY</td>
<td>xii</td>
</tr>
<tr>
<td>I</td>
<td>INTRODUCTION</td>
<td>1</td>
</tr>
<tr>
<td>II</td>
<td>ANATOMICAL INFORMATION</td>
<td>6</td>
</tr>
<tr>
<td></td>
<td>2.1 Anatomy and Physiology of a Motor Unit</td>
<td></td>
</tr>
<tr>
<td></td>
<td>2.2 Musculature and Bone Structure of the Elbow Joint</td>
<td></td>
</tr>
<tr>
<td>III</td>
<td>LITERATURE SURVEY OF PARAMETERS</td>
<td>45</td>
</tr>
<tr>
<td></td>
<td>3.1 Electromyographic Studies of the Elbow Joint</td>
<td></td>
</tr>
<tr>
<td></td>
<td>3.2 Studies of EMG-Load Relationship</td>
<td></td>
</tr>
<tr>
<td></td>
<td>3.3 Experimental Reproducibility</td>
<td></td>
</tr>
<tr>
<td></td>
<td>3.4 Muscle Fatigue</td>
<td></td>
</tr>
<tr>
<td></td>
<td>3.5 Miscellaneous Parameters</td>
<td></td>
</tr>
<tr>
<td>IV</td>
<td>EXPERIMENTAL EQUIPMENT</td>
<td>77</td>
</tr>
<tr>
<td></td>
<td>4.1 Load Device</td>
<td></td>
</tr>
<tr>
<td></td>
<td>4.2 EMG Equipment</td>
<td></td>
</tr>
<tr>
<td></td>
<td>4.3 Electrodes</td>
<td></td>
</tr>
<tr>
<td>V</td>
<td>EXPERIMENTAL METHODS</td>
<td>94</td>
</tr>
<tr>
<td></td>
<td>5.1 Subjects</td>
<td></td>
</tr>
<tr>
<td></td>
<td>5.2 Electrode Placement</td>
<td></td>
</tr>
<tr>
<td></td>
<td>5.3 Experimental Procedure</td>
<td></td>
</tr>
<tr>
<td></td>
<td>5.4 Reproducibility Experiments</td>
<td></td>
</tr>
<tr>
<td>VI</td>
<td>DATA REDUCTION</td>
<td>102</td>
</tr>
</tbody>
</table>
TABLE OF CONTENTS (Concluded)

Chapter

VII. DATA ANALYSIS AND INTERPRETATION .............. 106
   7.1 Graphs: Integrated EMG versus
       Integrated Torque
   7.2 Numerical Methods of Data Analysis

VIII. CONCLUSIONS ...................................... 143

Appendices

   A. GRAPHS: AVERAGE INTEGRATED EMG VERSUS AVERAGE
      INTEGRATED TORQUE ............................. 149
   B. SAMPLING RATE DETERMINATION .................. 234

BIBLIOGRAPHY ........................................... 238
<table>
<thead>
<tr>
<th>Table</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Number of Muscle Fibers per Motor Unit in Human Muscle</td>
<td>18</td>
</tr>
<tr>
<td>2. Prominent Elbow Flexor Muscles (PEF)</td>
<td>109</td>
</tr>
<tr>
<td>3. IEMG at Maximum Torque (Prone 90°)</td>
<td>111</td>
</tr>
<tr>
<td>4. IEMG at Maximum Torque (Supine 90°)</td>
<td>112</td>
</tr>
<tr>
<td>5. Effect of Forearm Position and Elbow Angle on IEMG at Maximum Torque for Each Muscle</td>
<td>114</td>
</tr>
<tr>
<td>6. Effect of Forearm Position and Elbow Angle on Ordering of Muscles According to IEMG at Maximum Torque</td>
<td>116</td>
</tr>
<tr>
<td>7. Experiments No. 6, 9, 11 for Subject SST, Values of IEMG at Maximum Torque of Experiment No. 11</td>
<td>128</td>
</tr>
<tr>
<td>8. Experiments No. 6, 9, 11 for Subject SST, Muscle Ordering According to IEMG at Maximum Torque</td>
<td>129</td>
</tr>
<tr>
<td>9. Least Squares Curve Fitting Technique Results</td>
<td>140</td>
</tr>
</tbody>
</table>
# LIST OF ILLUSTRATIONS

<table>
<thead>
<tr>
<th>Figure</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. A Motor Unit and Spinal Cord Cross-section</td>
<td>8</td>
</tr>
<tr>
<td>2. A Muscle Fiber</td>
<td>8</td>
</tr>
<tr>
<td>3. Distribution of Muscle Fiber Diameters</td>
<td>10</td>
</tr>
<tr>
<td>4. Firing Frequency of a Motor Unit as a Function of Tension</td>
<td>10</td>
</tr>
<tr>
<td>5. A Synovial Joint</td>
<td>24</td>
</tr>
<tr>
<td>6. Right Humerus</td>
<td>26</td>
</tr>
<tr>
<td>7. Right Radius and Ulna</td>
<td>28</td>
</tr>
<tr>
<td>8. Lateral View of Right Radius and Ulna</td>
<td>30</td>
</tr>
<tr>
<td>9. Forearm in Pronation</td>
<td>30</td>
</tr>
<tr>
<td>10. Bones of the Shoulder Joint</td>
<td>31</td>
</tr>
<tr>
<td>11. Bones of the Right Hand from in Front</td>
<td>33</td>
</tr>
<tr>
<td>12. Right Biceps Brachii</td>
<td>35</td>
</tr>
<tr>
<td>13. Cross-section of the Upper Arm</td>
<td>36</td>
</tr>
<tr>
<td>14. Superficial Muscles of the Right Arm (Front)</td>
<td>38</td>
</tr>
<tr>
<td>15. Muscles of the Upper Arm</td>
<td>38</td>
</tr>
<tr>
<td>16. Anterior View of Origins and Insertions of Muscles Studied in This Thesis</td>
<td>41</td>
</tr>
<tr>
<td>17. Posterior View of Origins and Insertions of Muscles Studied in This Thesis</td>
<td>41</td>
</tr>
<tr>
<td>18. Geometric Anatomical Model (90° Elbow Angle)</td>
<td>43</td>
</tr>
<tr>
<td>19. Geometric Anatomical Model (150° Elbow Angle)</td>
<td>44</td>
</tr>
<tr>
<td>20. Top View of Load Device</td>
<td>81</td>
</tr>
<tr>
<td>Figure</td>
<td>Page</td>
</tr>
<tr>
<td>--------</td>
<td>------</td>
</tr>
<tr>
<td>21. Strain Gage Mounting on Beam</td>
<td>83</td>
</tr>
<tr>
<td>22. Strain Gage Bridge Circuit</td>
<td>84</td>
</tr>
<tr>
<td>23. Load Device Calibration Curve</td>
<td>87</td>
</tr>
<tr>
<td>24. Steps in Making Bipolar Fine-wire Electrodes</td>
<td>91</td>
</tr>
<tr>
<td>25. EMG Equipment</td>
<td>92</td>
</tr>
<tr>
<td>26. Insertion of Intramuscular Electrodes</td>
<td>92</td>
</tr>
<tr>
<td>27. Dental Chair and Equipment</td>
<td>95</td>
</tr>
<tr>
<td>28. Subject and Loading Device</td>
<td>95</td>
</tr>
<tr>
<td>29. Diagram of Data Acquisition and Reduction Process</td>
<td>103</td>
</tr>
<tr>
<td>30. Biceps Brachii and Orientation of Humerus Bone</td>
<td>119</td>
</tr>
<tr>
<td>31. Total IEMG Magnitudes Versus Maximum Torque (Prone 90°)</td>
<td>123</td>
</tr>
<tr>
<td>32. Total IEMG Magnitudes Versus Maximum Torque (Supine 90°)</td>
<td>124</td>
</tr>
<tr>
<td>33. Experiment 1, Forearm Prone, Elbow Angle 90°</td>
<td>154</td>
</tr>
<tr>
<td>34. Experiment 1, Forearm Supine, Elbow Angle 90°</td>
<td>155</td>
</tr>
<tr>
<td>35. Experiment 1, Forearm Prone, Elbow Angle 150°</td>
<td>156</td>
</tr>
<tr>
<td>36. Experiment 1, Forearm Supine, Elbow Angle 150°</td>
<td>157</td>
</tr>
<tr>
<td>37. Experiment 2, Forearm Prone, Elbow Angle 90°</td>
<td>160</td>
</tr>
<tr>
<td>38. Experiment 2, Forearm Supine, Elbow Angle 90°</td>
<td>161</td>
</tr>
<tr>
<td>39. Experiment 2, Forearm Prone, Elbow Angle 150°</td>
<td>162</td>
</tr>
<tr>
<td>40. Experiment 2, Forearm Supine, Elbow Angle 150°</td>
<td>163</td>
</tr>
<tr>
<td>41. Experiment 3, Forearm Prone, Elbow Angle 90°</td>
<td>166</td>
</tr>
<tr>
<td>42. Experiment 3, Forearm Supine, Elbow Angle 90°</td>
<td>167</td>
</tr>
<tr>
<td>Figure</td>
<td>Description</td>
</tr>
<tr>
<td>-------</td>
<td>-------------------------------------------------</td>
</tr>
<tr>
<td>43</td>
<td>Experiment 3, Forearm Prone, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>44</td>
<td>Experiment 3, Forearm Supine, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>45</td>
<td>Experiment 4, Forearm Prone, Elbow Angle $90^\circ$</td>
</tr>
<tr>
<td>46</td>
<td>Experiment 4, Forearm Supine, Elbow Angle $90^\circ$</td>
</tr>
<tr>
<td>47</td>
<td>Experiment 4, Forearm Prone, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>48</td>
<td>Experiment 4, Forearm Supine, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>49</td>
<td>Experiment 5, Forearm Prone, Elbow Angle $90^\circ$</td>
</tr>
<tr>
<td>50</td>
<td>Experiment 5, Forearm Supine, Elbow Angle $90^\circ$</td>
</tr>
<tr>
<td>51</td>
<td>Experiment 5, Forearm Prone, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>52</td>
<td>Experiment 5, Forearm Supine, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>53</td>
<td>Experiment 6, Forearm Prone, Elbow Angle $90^\circ$</td>
</tr>
<tr>
<td>54</td>
<td>Experiment 6, Forearm Supine, Elbow Angle $90^\circ$</td>
</tr>
<tr>
<td>55</td>
<td>Experiment 6, Forearm Prone, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>56</td>
<td>Experiment 6, Forearm Supine, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>57</td>
<td>Experiment 7, Forearm Prone, Elbow Angle $90^\circ$</td>
</tr>
<tr>
<td>58</td>
<td>Experiment 7, Forearm Supine, Elbow Angle $90^\circ$</td>
</tr>
<tr>
<td>59</td>
<td>Experiment 7, Forearm Prone, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>60</td>
<td>Experiment 7, Forearm Supine, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>61</td>
<td>Experiment 8, Forearm Prone, Elbow Angle $90^\circ$</td>
</tr>
<tr>
<td>62</td>
<td>Experiment 8, Forearm Supine, Elbow Angle $90^\circ$</td>
</tr>
<tr>
<td>63</td>
<td>Experiment 8, Forearm Prone, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>64</td>
<td>Experiment 8, Forearm Supine, Elbow Angle $150^\circ$</td>
</tr>
<tr>
<td>65</td>
<td>Experiment 9, Forearm Prone, Elbow Angle $90^\circ$</td>
</tr>
<tr>
<td>Figure</td>
<td>Description</td>
</tr>
<tr>
<td>--------</td>
<td>-----------------------------------------------------------------------------</td>
</tr>
<tr>
<td>66</td>
<td>Experiment 9, Forearm Supine, Elbow Angle 90°</td>
</tr>
<tr>
<td>67</td>
<td>Experiment 10, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>68</td>
<td>Experiment 10, Forearm Supine, Elbow Angle 90°</td>
</tr>
<tr>
<td>69</td>
<td>Experiment 11, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>70</td>
<td>Experiment 11, Forearm Supine, Elbow Angle 90°</td>
</tr>
<tr>
<td>71</td>
<td>Experiment 12, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>72</td>
<td>Experiment 12, Forearm Supine, Elbow Angle 90°</td>
</tr>
<tr>
<td>73</td>
<td>Experiment 9, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>74</td>
<td>Experiment 9, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>75</td>
<td>Experiment 9, Forearm Supine, Elbow Angle 90°</td>
</tr>
<tr>
<td>76</td>
<td>Experiment 9, Forearm Supine, Elbow Angle 90°</td>
</tr>
<tr>
<td>77</td>
<td>Experiment 10, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>78</td>
<td>Experiment 10, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>79</td>
<td>Experiment 10 and 12, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>80</td>
<td>Experiment 10 and 12, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>81</td>
<td>Experiment 10 and 12, Forearm Supine, Elbow Angle 90°</td>
</tr>
<tr>
<td>82</td>
<td>Experiment 10 and 12, Forearm Supine, Elbow Angle 90°</td>
</tr>
<tr>
<td>83</td>
<td>Experiment 6, 9, and 11, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>84</td>
<td>Experiment 6, 9, and 11, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>85</td>
<td>Experiment 6, 9, and 11, Forearm Supine, Elbow Angle 90°</td>
</tr>
<tr>
<td>86</td>
<td>Experiment 6, 9, and 11, Forearm Supine, Elbow Angle 90°</td>
</tr>
<tr>
<td>Figure</td>
<td>Description</td>
</tr>
<tr>
<td>--------</td>
<td>--------------------------------------</td>
</tr>
<tr>
<td>87</td>
<td>Experiment 3, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>88</td>
<td>Experiment 3, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>89</td>
<td>Experiment 8, Forearm Prone, Elbow Angle 90°</td>
</tr>
<tr>
<td>90</td>
<td>Experiment 8, Forearm Prone, Elbow Angle 90°</td>
</tr>
</tbody>
</table>
SUMMARY

The elbow joint is studied by correlating the electromyographic (EMG) activity generated in the muscles of the elbow joint with the torque sustained by the joint.

A literature survey was made of the parameters involved in electromyographic research in general, and specifically, in EMG studies of the elbow joint.

A study was made of pertinent anatomical information including a brief review of the anatomy and physiology of the motor unit and the electromyographic signal, as well as a study of the elbow joint musculature and bone structure. From this information, the muscles involved in the elbow joint function were determined to be: Brachialis, Biceps Short Head, Biceps Long Head, Brachioradialis, Pronator Teres, Triceps, Extensor Carpi Radialis and Flexor Carpi Radialis. Also, a geometric anatomical model of the elbow joint including these eight muscles was developed.

A device was developed to enable a subject to isometrically apply a measured torque to his elbow joint. An appropriate set of experiments was designed, and ten subjects were tested. Using intramuscular electrodes, the EMG activity in the eight muscles specified above was recorded while a simultaneously recorded torque was applied by the subject to his elbow joint.

Quantitative data reduction was accomplished by integrating
both the EMG and Torque analog information with respect to time. The data is presented in graphical form as Average Integrated EMG plotted versus Average Integrated Torque for each muscle.

The conclusions are based on both the experimental data and the geometric anatomical model. The integrity of the electromyographic model of the elbow joint is evaluated.

The model is concluded to be quite adequate in describing the major elbow joint flexor muscles, Brachialis, Biceps Long Head, and Biceps Short Head. Three explanations are presented for model inadequacies:

(1) The two-joint muscles thought to be primarily associated with the wrist joint should be included in the elbow joint model,

(2) The assumption that integrated EMG is proportional to the load sustained by a muscle is not completely valid, and

(3) The assumption that a single sample of the EMG activity in the muscle is representative of the total EMG activity in the muscle is invalid. To compensate for such invalidity, a muscle size factor indicating number of motor units per muscle is suggested.

The functional relationship between integrated EMG and integrated Torque is concluded to be dependent on the muscle, subject, etc. This relationship varies from a proportional to a cubic function. Elbow angle and forearm position are determined to be important parameters in evaluating elbow joint performance. Hysteresis was demonstrated in muscle and it was shown to vary for each muscle, suggesting that the functional aspects of the muscles are involved. Electrode migration was found to be negligible. Day-to-day reproducibility of data for one subject
tested on different days was poor and concluded to be most difficult to achieve.
CHAPTER I

INTRODUCTION

In this research the elbow joint is studied by correlating the electromyographic activity (EMG) generated in the muscles of the elbow joint with the torque sustained by the joint.

Electromyography is the study of myoelectric signals, that is, voltage potentials which produce muscle contraction. Originating in the spinal cord, a neural signal is carried by a single nerve fiber which activates several muscle fibers causing these fibers to contract. The tetanus of an entire muscle is the result of many individual muscle fiber contractions. The electromyographic signal (EMG) (See p. 12) can be picked up and monitored by surface electrodes on the skin and by intramuscular electrodes which are inserted into the muscle.

Currently used as a clinical tool to study neuromuscular diseased, electromyography has also proved useful in determining muscle function. For instance, in 1957, Basmajian and Latif [1] did an electromyographic study of Brachialis, Brachioradialis, and both heads of Biceps. Movements studied included elbow flexion, extension, maintenance of elbow flexion at elbow angles of 135° and 90°, pronation and supination of the forearm, and flexion and abduction of the shoulder. These movements were investigated with no resistance and with a two-pound weight held in the hand. A photographic record
of EMG activity was made and then interpreted on a basis of five degrees of activity plus no activity. The raw EMG data in many such muscle function studies were similarly judged in a qualitative manner.

As the availability of data processing equipment has increased, a variety of quantitative methods of data reduction have been utilized in electromyographic studies. As early as 1952, Lippold [2] used a planimeter to integrate the raw EMG signal. Using an electronic counter, Close, et al. [3] determined the number of spikes or action potential counts in 0.1 second intervals for ten second muscle contractions. Some investigators that calculated the root mean square values of the electromyogram are DeVries [4] in 1965, Nightingale [5] in 1960, and Zuniga and Simons [6] in 1968. Various frequency analysis techniques have been used. The first report of clinical application of direct frequency analysis was made by Walton [7] in 1952, using an audio-frequency spectrometer. Cenkovich and Gersten [8] in 1963, did a study to determine whether Fourier analysis could be used to detect changes in electromyographic characteristics (see Appendix B). This analysis was performed on a synthetic diphasic wave which simulated the muscle action potential. A similar frequency analysis of the myoelectric signal was presented by Scott [9] in 1967. Using a digital computer, the auto-correlation function was determined, and from this, the power density spectrum was obtained.

The most common means of quantitative data reduction of EMG is time integration, mainly because of the availability of integrating
equipment. It should be recognized that in some instances, a simulated time integration method, actually a low-pass filter, is used. Irman, et al. [10], in 1952, devised a rectifying and filtering circuit which they "called an integrator for lack of a better name". Two studies in which pure time integration was used are Liberson, et al. [11] (1961) and Lovejoy and Basmajian [12] (1970). In both these studies, the raw analog EMG signal was rectified and digitally sampled over a time interval to obtain the area under the EMG versus time analog curve.

In order to study muscle function on an even more quantitative basis, researchers have attempted to investigate the EMG activity in a muscle as a function of the tension developed in the muscle (see Chapter III). Unfortunately, in most of these studies, the applied load was sustained by several muscles rather than by a single muscle. For instance, it would be most difficult to apply a load to Biceps Brachii without loading other muscles as well. Likewise, it would be equally difficult to measure, in vivo, the tension developed in human Biceps Brachii.

In this analysis, an entire muscle system is studied by correlating the EMG activity generated in the muscles as a result of the load applied to the muscle system. The elbow joint musculature was chosen as the muscle system for several reasons:

(1) The elbow joint is relatively complex, involving several muscles.

(2) This joint has been neglected in EMG muscle function studies.
(3) The elbow joint would be fairly simple to load. Time integration was chosen as the method of EMG data reduction as well as load data reduction, because this method is the most commonly used quantitative technique of EMG data reduction.

The purpose of this thesis study is, then, two-fold:

(1) to make a quantitative electromyographic study of the elbow joint, and

(2) to investigate the functional relationship between time integrated EMG and time integrated load.

To accomplish this purpose, the investigation proceeded in the following manner.

(1) A literature survey was made of the parameters involved in electromyographic research in general, and specifically in EMG studies of the elbow joint.

(2) A study was made of pertinent anatomical information including a brief review of the anatomy and physiology of the motor unit and the electromyographic signal, as well as a study of the elbow joint musculature and bone structure. From this information, the muscles involved in the elbow joint function were determined to be: Brachialis (B), Biceps Short Head (BS), Biceps Long Head (BL), Brachioradialis (BR), Pronator Teres (PT), Triceps (T), Extensor Carpi Radialis (WE) and Flexor Carpi Radialis (WF). The best method of loading the elbow joint was determined. The interaction of the elbow joint with the wrist and shoulder joints was studied revealing that the wrist and shoulder joints would have to be unloaded to properly study the elbow joint musculature. Also, a geometric
anatomical model of the elbow joint was developed.

(3) A device was developed to enable a subject to isometrically apply a measured torque to his elbow joint.

(4) An appropriate set of experiments was designed and, ten subjects were tested. The EMG activity in the muscles specified by the anatomical model was recorded while a simultaneously recorded torque was applied by the subject to his elbow joint.

(5) Quantitative data reduction was accomplished by integrating both the EMG and torque analog information with respect to time. The data is presented in graphical form as Average Integrated EMG plotted versus Average Integrated Torque for each muscle monitored.

(6) The experimental data was compared to the geometric anatomical model and conclusions drawn regarding the elbow joint musculature and the EMG-Torque relationships for time integrated data reduction.
CHAPTER II

ANATOMICAL INFORMATION

2.1 Anatomy and Physiology of a Motor Unit

2.1.1 General Introduction

In striated muscle (voluntarily controlled muscle)*, the structural unit of contraction is the muscle fiber. With maximum length of 30 cm and diameter less than 0.1 mm, a muscle fiber will shorten to approximately two-thirds of its resting length during contraction. A group of muscle fibers is enervated by one nerve fiber, so that the muscle fibers of a group contract simultaneously. The cell body of the nerve fiber (axon) is in the anterior horn of the spinal grey matter (see Section 2.1.2). The nerve cell body, its single long axon with the terminal branches, and all the muscle fibers innervated by these branches is defined as a motor unit [13]. A schematic representation of a motor unit is shown in Figure 1.

The motor unit is the physiological unit of reflex and willed contraction in mammalian skeletal muscle. It behaves in an all or nothing manner in that an impulse in the motor nerve fiber produces an effectively synchronous contraction of all muscle fibers it

* There are three types of muscle: (1) voluntary, or striated, or skeletal, (2) involuntary, or unstriated, or smooth, and (3) cardiac, having many of the characteristics of the other two kinds of muscle. Voluntary or striated muscle is muscle under the control of the individual [16].
supplies. Each striated muscle fiber responds once, and only once to each impulse passing down its motor nerve [14]. Mammalian skeletal muscle may be regarded as an electromechanical transducer in which the physiological input is the neural signal originating in the spinal cord, and its output is muscle contraction [15].

2.1.2 Anterior Horn Cells and Axons

Thirty-one pairs of spinal nerves are attached to the spinal cord. Each nerve is attached to the cord by two roots - an anterior (towards the front) and a posterior (towards the back). The anterior roots are efferent (motor) - they issue from the cord; the posterior roots are afferent (sensory) - they enter the cord. A typical cross-section of the spinal cord, as shown in Figure 1, reveals an outer covering of white matter with a central mass of grey matter in the shape of the letter H. The nerve cell bodies whose axons make up the anterior roots, are in the anterior horn of the grey matter of the spinal cord; they are called anterior horn cells [16]. The two types of anterior horn cells are:

(1) Motor cells from which large (6-20μ diameter) motor nerve fibers originate. These motor cells are arranged in columns which extend through many spinal segments and which appear as rounded clusters in transverse sections of the spinal cord. These cells innervate the extrafusal muscle fibers.

(2) Motor cells from which small (1-5μ diameter) fibers originate. The smaller cells are more diffusely arranged in the spinal grey matter. They innervate the intrafusal fibers of the
Figure 1. A Motor Unit and Spinal Cord Cross-section [13].

Figure 2. A Muscle Fiber [16].
muscle spindles (see Section 2.1.8) [17,18].

The axons or nerve fibers of the anterior horn cells provide the only motor pathway to the muscles. Conduction rate in the large motor nerve fibers is about 40-70 msec while signals are conducted in the small fibers at about half that rate [18].

2.1.3 Muscle Fibers

Muscle fibers are cylindrical in shape, varying in diameter from 30 to 90μ, with an average diameter of 50μ in large extremity muscles such as human Biceps Brachii. The muscle fiber diameter [18] varies within a particular muscle of one subject in the manner of a normal distribution, as shown by Figure 3. The mean muscle fiber diameter varies between different muscles in the same individual, and between different individuals in the same muscle. Variations in the latter case are explained in part by differences in build and muscular development. The number of muscle fibers in a muscle increases throughout foetal life, with little change in the diameter of the fibers. From birth on, the growth of the muscle results from increase in length and thickness of fibers, without further increase in number.

With a length varying from a few mm to 30 cm muscle fibers may extend from one end of a muscle to the other, or may be attached to tendinous intersections within the muscle. Terminal parts of the muscle fibers may be finely tapered, or rounded, or expanded. Fibers are arranged in fasciculi, each of which is a bundle of more or less parallel fibers bound together and surrounded by connective tissue containing blood vessels and nerves. Muscle fibers often pass from
Figure 3. Distribution of Muscle Fiber Diameters [17].

Figure 4. Firing Frequency of a Motor Unit as a Function of Tension [19].
one fasciculus to another, and distribution of motor units is such
that their pattern does not necessarily conform to the pattern of
muscle fiber bundles.

As mentioned previously, in normal striated muscle, the
structural unit of contraction is the muscle fiber. It contains
within its sarcoplasm* longitudinally oriented myofibrils, all
of which is enclosed by a delicate membrane called the sarcolemma,
as shown in Figure 2. Because of the orderly arrangement and
lateral alignment of the myofibrils an appearance of transverse
banding is given to the muscle fiber which permits its division into
structural subunits, sarcomeres. Within the sarcomere is an orderly
array of two types of protein filaments, actin and myosin. The
myosin filaments are arranged longitudinally within the central
portion of the cylindrical sarcomere, while the actin filaments,
also arranged longitudinally, radiate from the boundaries of the
sarcomere, and interdigitate with the myosin filaments. The shortening
produced during the muscle contraction is brought about by a sliding
of the actin filaments between the myosin filaments, a mechanism
which results in a net shortening of the sarcomere, while the length
of the actin and myosin filaments remains unaltered. It is thought
that the tension developed by muscle fibers is related to the number

* The specialized protoplasm of the muscle fiber. Protoplasm is a
substance composed of hydrogen, oxygen, nitrogen, chlorine,
calcium, carbon, iron, sodium, potassium and phosphorous, and
having the unusual characteristic of possessing life. Every
living thing is essentially an aggregation of protoplasmic units
and their products [16].
of bridges, and that the maximum tension a single muscle fiber can develop is related to the extent of actin/myosin overlap [17].

2.1.4 Myoneural Junction

The points of contact between the terminal part of the motor axons and the muscle fibers innervated by it are known as the myoneural junctions or motor end-plates. Each motor end-plate is a composite structure, belonging partly to the motor nerve or axon and partly to the muscle fiber. The actual area of contact or synapse, between the axon and the muscle fiber is hardly visible in the light microscope, but appears on the electron microscope examination as a double membrane which is folded in a complex [17] manner. The end-plate is confined to discrete zones and is usually situated around the middle portion of the muscle fiber. In the Biceps Brachii, the end-plate zone is about 15 per cent of the fiber length [18].

2.1.5 Motor Unit Potential

"When an impulse reaches the myoneural junction, a wave of contraction spreads over the muscle fiber resulting in a brief twitch followed by rapid and complete relaxation. During this brief twitch, a minute electrical potential is generated which is dissipated into the surrounding tissues. The duration of this fiber potential is about 1 to 4 msec. Since not all muscle fibers of a motor unit contract at exactly the same instant, the electrical potential developed by the single twitch of all fibers in the motor unit is prolonged to about 5 to 12 msec" [13].

These motor unit potentials (MUP) are picked up by electrodes
of two types:

(1) skin or surface electrodes which are applied directly to the skin, and

(2) intramuscular electrodes which are inserted, usually with a needle, into the muscle.

Of course, the motor unit potentials observed on an oscilloscope or other monitoring device are greatly affected by the distance the electrodes are from the units and the type of electrodes, amplifiers, and other equipment used. The observed MUP does not derive necessarily from all the fibers of the motor unit, but rather from those fibers situated close to the electrodes. This factor is a function of the distribution of the muscle fibers of a motor unit (see Section 2.1.6). Multielectrode studies indicate that a maximum number of 10 to 30 muscle fibers contribute to the MUP [18].

Motor unit potentials are identified by four main characteristics:

(1) wave shape (number of phases),

(2) peak-to-peak amplitude (mv),

(3) duration (msec), and

(4) rate of discharge (cps).

All four characteristics vary considerably for motor unit potentials in different areas of the same muscle.

Pinelli and Buchthal [20] reported that the wave form of over 80 per cent of normal muscle potentials is either diphasic or triphasic. The amplitude, peak-to-peak, of motor unit potentials vary considerably with electrode type and with distance between electrode and muscle fiber. The majority of motor unit potentials have an amplitude of
0.5 mv. Kaiser and Petersen [21] found with coaxial needle electrodes, a statistical distribution in Biceps Brachii of amplitudes from 0.05 to 0.3 mv.

The mean duration of motor unit potentials has been shown definitely dependent on the type of electrode used. A mean duration of 9 msec has been reported for clinical needle electrodes, while a figure of 5 msec is stated for fine wire electrodes (see Section 4.3). Potentials obtained through fine wire electrodes are also reported as much more polyphasic than those obtained through needle electrodes [22]. Kaiser and Petersen [21] recorded shorter durations in the long head of Biceps than the short head. The duration of motor unit potentials varies with electrode site within the same muscle (from 3 to 15 msec in Biceps reported by Buchthal [23]). Petersen and Kugelberg [24] using concentric needle electrodes, reported 3 to 18 msec in Biceps.

The maximum rate of discharge of a motor unit is generally accepted as 50 discharges per second (see Figure 4) [13]. Kaiser and Petersen [21] (1965) report repetition rates from 5 to 50 per sec in Biceps Brachii. Bigland and Lippold (1954) [19] also report these figures. Buchthal [23] also states findings of maximum repetition rate of 50 per second in limb muscles. However, frequencies of up to 500 discharges per second have been claimed to occur in vocal muscles during voice production by Husson [25] (1956). This has not been confirmed. Two groups of researchers, Bjork and Kugelberg [26] and Marg, Tamler, and Jampolsky [27] have claimed repetition rates in extraocular muscles as high as 200 - 270 per
second. Repetition rate or rate of discharge of a motor unit is not to be confused with frequency analysis of the electromyographic signal, in which the frequency range of significant energy is determined. This topic is discussed later in Appendix B.

Man can consciously control the firing rate of individual motor units, an ability which is being applied in the control of myoelectric prostheses, psychology, neurological studies, and even for treatment of reading disorders. For further discussion pertaining to motor unit training, the reader is referred to Chapter 5 of *Muscles Alive* by Basmajian [13].

Increased contraction of a muscle is produced by two mechanisms. The more important mechanism of muscle contraction gradation is recruitment of new motor units. The normal recruitment pattern is agreed to be smaller units ("smaller" meaning fewer muscle fibers per motor unit as well as small diameter fibers) recruited first and the larger potentials being recruited as the load on the muscle increases. The other mechanism is increase in rate of discharge of all the recruited motor units. Figure 4 indicates rate of discharge as a function of tension in a muscle. In a recent study [49], in the Biceps Brachii, the properties of a motor unit are reported to be determined by its tension threshold. Units of low threshold have a wide (7-20 discharges per second) firing frequency range and tend to be located deep in the muscle. The higher the tension threshold of the unit, the higher is its lowest firing frequency, the narrower is its firing frequency range, and the more superficially it is located.
2.1.6 Size and Distribution of Motor Units

Size. The size of a motor unit can be described by the innervation ratio, which is the ratio of the number of nerve fibers to the number of extrafusal muscle fibers in a muscle. In 1873, Tergast made counts in animals and observed that muscles which were concerned with fine movements (eye) have smaller innervation ratios (1:3) than those performing cruder movements (limb, from 1:8 to 1:20) [17]. The number of muscle fibers innervated by a single nerve fiber is also probably related to the inertial load requirement of a muscle. For example, to move the mass of a lower limb will require the simultaneous action of many muscle fibers, and consequently in the muscles responsible for such movements, a large number of muscle fibers per nerve fiber is found. The innervation ratio varies from 1:5 to 1:5000 [14].

The innervation ratio is often calculated by determining the total number of muscle fibers in a muscle and the total number of nerve fibers in its motor nerve and dividing the former by the latter [13]. There are a number of complicating factors to be considered when evaluating the innervation ratio in this manner. Correction is required for

(1) the fact that 40 per cent of the fibers among the large nerve fibers are sensory fibers,

(2) the small motor nerve fibers supplying the intrafusal muscle fibers of the muscle spindles, and

(3) multiple innervation [18].
Table 1 [18] shows some values for the number of muscle fibers per motor unit in human muscle. In most limb muscles, the motor units contain 400-1700 muscle fibers [18].

The question of multiple innervation (one muscle fiber being innervated by more than one nerve fiber) is still not completely answered. In the frog and cat, there is evidence of multiple innervation of muscle fibers derived either from one branching nerve fiber or from different motor neurons. In man there was no evidence of multiple innervation in the Superior Rectors of the eye, the Semitendinous (hamstring muscle) and the Femoral Rectus. In Biceps Brachii and the Oppones Pollicis (thumb flexor) there are 1.3 to 1.5 motor end-plates per muscle fiber indicating multiple innervation of some fibers. The extent of the end-plate zone in these muscles does not differ from that of muscles in which there is only one end-plate per muscle fiber. Hence, the two end-plates must be assumed to be situated at a short distance from each other and multiple innervation can hardly play any role in reducing the time required for activation of the fiber. Such controversial evidence, nonetheless, substantiates the uncertainty associated with the possibility of multiple innervation of muscle fiber [18].

Distribution. The anatomical distribution of muscle fibers activated by a single motor neuron varies somewhat. Animal research has indicated that the individual muscle fibers of a motor unit are scattered and intermingled with fibers of other units in the rabbit and monkey [28] and the rat [29,30,31]. However, Knott, et al. [32]
Table 1. Number of Muscle Fibers per Motor Unit in Human Muscle [18].

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Number of Nerve Fibers</th>
<th>Number of Large Nerve Fibers</th>
<th>Number of Motor Units</th>
<th>Total Number of Muscle Fibers</th>
<th>Number of Muscle Fibers Per Motor Unit</th>
<th>Mean Diameter of Muscle Fibers, µ</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Rectus Oculi Ext. (eye muscle)</td>
<td>4150</td>
<td>2900</td>
<td>1740</td>
<td>22000</td>
<td>13</td>
<td>15</td>
</tr>
<tr>
<td>2. Tensor Tympani (internal ear muscle)</td>
<td>146</td>
<td>91</td>
<td>55</td>
<td>1100</td>
<td>20</td>
<td>not listed</td>
</tr>
<tr>
<td>3. Platysma (neck muscle)</td>
<td>not listed</td>
<td>1826</td>
<td>1096</td>
<td>27100</td>
<td>25</td>
<td>20</td>
</tr>
<tr>
<td>4. Opponens Pollicis (thumb flexor)</td>
<td>790</td>
<td>221</td>
<td>133</td>
<td>79000</td>
<td>595</td>
<td>not listed</td>
</tr>
<tr>
<td>5. Biceps Brachii</td>
<td>3790</td>
<td>1290</td>
<td>774</td>
<td>520000</td>
<td>750</td>
<td>50</td>
</tr>
<tr>
<td>6. Tibialis Anterior (foot dorsiflexor)</td>
<td>not listed</td>
<td>742</td>
<td>445</td>
<td>270000</td>
<td>610</td>
<td>57</td>
</tr>
<tr>
<td>7. Gastrocnemius (calf muscle)</td>
<td>not listed</td>
<td>965</td>
<td>579</td>
<td>1000000</td>
<td>1730</td>
<td>54</td>
</tr>
</tbody>
</table>
found in the cat that the muscle fibers of a motor unit were not uniformly scattered over a muscle, but each had a distinct localized territory. Work has been done on human muscle by Buchthal, Guld, and Rosenfalck [33] (1957) and Buchthal, Erminio, Rosenfalck [34](1959), which indicates widespread distribution of the muscle fibers of a motor unit. They demonstrated that in Biceps Brachii, the muscle fibers of a motor unit were localized in a circular region with an average diameter of 5 mm. However, some contributions to the motor unit potential could be found within a 20 mm diameter region. They also showed that the 5 mm region contains the fibers of many overlapping motor units.

2.1.7 Neuromuscular Transmission [35,36]

The steps in the spread of excitation from nerve to muscle are as follows:

1. Arrival of the nerve impulse from the motor cell at the myoneural junction with subsequent liberation of acetylcholine, ACh.

2. Depolarization of the end-plate or the myoneural junction by the released ACh.

3. Initiation of propagated action potential in the neighboring region of the muscle fiber.

4. Spread by conduction of the muscle action potential along the muscle fiber membrane.

5. Inward conduction of excitation from the surface of the muscle fiber to the myofibrils lying within it.
(6) Contraction of the myofibrils.

Transmission at the neuromuscular junction is chemical, the transmitter substance being acetylcholine (ACh). At the junction, there is a time delay (about 0.5 msec in mammals) between the arrival of the impulse at the motor nerve ending and its further propagation in the muscle fiber. In this time delay, steps 2 and 3 listed above are accomplished.

The motor nerve terminal at rest releases ACh in packets or quanta, of uniform size, at random. These liberated quanta depolarize a portion of the end-plate region of the muscle fiber membrane. These depolarizations, termed miniature end-plate potentials, are less than a millivolt in amplitude, and hence give rise to no propagating action potentials. Arrival of the nerve impulse at the motor nerve ending causes simultaneous release of hundreds of quanta of ACh. This release produces the end-plate potential and gives rise to a propagated muscle action potential.

The creation of end-plate potential begins when ACh molecules combine with certain receptors. This combination is followed by an increased ionic permeability of the muscle membrane, which has a muscle resting potential of 90 mv. That resting potential is caused by high concentrations of Na\(^+\) and Cl\(^-\) ions in the extracellular fluid and high concentrations of K\(^+\) ions in the intracellular fluid. This unequal distribution of inorganic ions across the muscle fiber cell membrane gives rise to an electrical potential difference, 90 mv with the inside negative, between the interior of the cell and its environment. Increasing ionic permeability of the muscle membrane
changes the resting potential with the ion exchange. This potential change is the end-plate potential. Such change, or depolarization, generates the muscle action potential.

The muscle action potential is propagated longitudinally along the muscle fiber by a mechanism similar to the mechanism by which the end-plate potential is produced. The muscle action potential is propagated by changes in ionic permeability of the muscle fiber cell membrane resulting in transmembrane potential changes. In Biceps Brachii, the velocity along the muscle fiber of action potential propagation is 4 to 5.5 m/sec. Thus, when a nerve impulse arrives at the end-plate the whole muscle fiber is activated almost simultaneously. Activation speed is further enhanced by the fact that the end-plate is located in the middle of the muscle fiber.

Inward conduction of the excitation from the fiber surface to myofibrils is quite complex and still subject to speculation. The contraction of the myofibrils has been described in Section 2.1.3.

2.1.8 Muscle Spindles

The muscle spindles are sensory structures, responding to tension in the muscle; that is, they provide a feedback mechanism which controls the force of muscular contraction. Muscle spindles are elongated structures, each several millimeters in length, made up of a collection of specialized intrafusal muscle fibers together with nerves and blood vessels and surrounded by a connective tissue capsule [17].
2.2 Musculature and Bone Structure of the Elbow Joint

2.2.1 Anatomy of the Elbow Joint

The following section was derived from Primary Anatomy by Dr. J. V. Basmajian [16]. For convenience, anatomists have defined the anatomical position of the human body as reference position for all descriptions. The anatomical position is that position of a living body standing erect with arms by the sides and palms facing forward. All descriptive terms refer to this position. Some of the more common terms are defined below:

(1) Anterior: toward or at the front.
   Posterior: toward or at the back.

(2) Lateral: away from the body midline.
   Medial: toward the midline.

(3) Proximal: referring to limbs, closer to the trunk.
   Distal: referring to limbs, farther from the trunk.

(4) Superior: higher or above.
   Inferior: lower or below.

Anatomists refer to three planes of the body:

(1) Sagittal plane: that plane which divides the body into right and left parts.

(2) Frontal plane: any vertical plane dividing the body into anterior and posterior sections.

(3) Horizontal plane: the body plane which cuts the long axis of the body at right angles.

Another convention, used by anatomists, is the manner in which
muscle attachments to bones are described. The proximal attachment of a muscle in the limbs is termed the origin; the distal attachment is called the insertion. Generally, the origin of a muscle extends over a larger area than the insertion. Usually the insertion is near the joint moved by the muscle.

The elbow joint is a synovial joint. The four distinguishing features of this type joint shown in Figure 5 are:

1. a joint cavity,
2. lubricated articular cartilage,
3. fibrous capsule, and
4. a synovial membrane.

The joint cavity separating the bones is bounded by a sleeve of fibrous tissue, the fibrous capsule, which is all that unites the bones. Friction between the bones is minimized by the articular cartilage on the bone ends and by cells of the inner synovial membrane of the fibrous capsule which transude a lubricating fluid.

The elbow joint is the hinge joint between the lower end of the humerus bone and the upper ends of ulna and radius bones. The three bones of the elbow joint are known as long bones, consisting of a hollow shaft and expanded ends. The shaft is encased in an outer shell of compact bone almost ivory-like in hardness and density. Within this shell, the bone assumes a trellis-like appearance and is termed spongy bone. At the middle of the shaft is a space known as the medullary cavity filled by yellow marrow. The expanded ends of long bones are particularly spongy, the compact bone being quite thin. The interstices of the spongy bone are filled with red marrow which
Figure 5. A Synovial Joint [16].
manufactures red blood cells. Covering the outside of bone is a tough fibrous membrane, periosteum, the inner surface of which consists of cells of bone growth and repair. At the expanded ends of long bone, periosteum is replaced by smooth and slippery cartilage. This articular cartilage is the persistent portion of the original cartilaginous precursor of limb bones, which in the embryo, are formed from cartilage.

The bone of the upper arm is the humerus, shown in Figure 6. The head, which is at the upper end of the bone, is the ball of the ball-and-socket shoulder joint. In close proximity to the head are two tuberosities or tubercles - a greater and lesser. Both are protrusions that serve as insertion points for numerous muscles surrounding the shoulder joint. These tubercles are separated by the bicipital groove in which lies the tendon from the long head of Biceps Brachii. The lower end of humerus curves forward to form two joint surfaces, one for ulna, medially, and the other laterally for radius. The humerus and ulna together form a secure hinge joint with a surface (the trochlea) on humerus fitting into a deep, rounded notch on ulna. Immediately above the trochlea are two hollows, the coronoid fossa, in front, and the olecranon fossa in back, which accommodate corresponding parts of ulna when the joint is flexed or extended. Adjoining the lateral part of the trochlea is a small rounded ball of bone, the capitulum, which articulate with the head of radius. A prominent process, the medial epicondyle, projects medially from the trochlea and provides origin for flexor muscles of the forearm. Similarly, the lateral epicondyle projects laterally
Figure 6. Right Humerus [16].
from the capitulum and provides origin for extensor muscles. A line through these epicondyles defines the center of rotation for the elbow joint.

The ulna is the medial of the two bones of the forearm and is shown in Figure 7. The upper end resembles an open end wrench; the upper jaw is the olecranon process (recall olecranon fossa on humerus) and the lower jaw is the coronoid process (corresponding to the coronoid fossa on humerus). Between these two projecting ledges of bone is the trochlear notch into which fits the trochlea of humerus. On the lateral side of the coronoid process is a small, shallow surface, the radial notch, into which fits the side of the head of radius. Below the coronoid process is the ulna tuberosity which receives insertion of Brachialis, the chief elbow flexor muscle. The shaft of ulna is triangular in cross-section and tapers to a small disc head from which projects a peg of bone, the styloid process.

Radius, shown in Figure 7, is the lateral of the two bones of the forearm and is united with ulna by a flexible interosseous membrane. Whereas the important functions of ulna lie at the elbow, those of radius lie at the wrist where the bone is charged with carrying the hand. Hence, while the head of radius at the elbow is small, the shaft becomes progressively more massive until at the wrist it is twice as wide as at the elbow. The head of radius articulates with the capitulum of humerus and is surrounded by a circular ring consisting of the radial notch of ulna and an annular ligament attached to ulna. Below the head on the medial side of the
Figure 7: Right Radius and Ulna [16].
shaft is the prominent radial tuberosity, which is the insertion point for Biceps Brachii. There is an obvious lateral convexity or bowing of the shaft. At the lower end is the concave ulnar notch into which the head of ulna fits. On the lateral side, is the styloid process which is larger than that of ulna, and projects farther distally, as shown in Figure 7. With the palm facing forward as in the anatomical position, that movement of turning the hand over such that the palm faces backward is pronation of the forearm. In this movement, at the wrist, the lower end of radius swings around the front of the head of ulna until the shafts are crossed. The reverse movement is supination of the forearm. If the shafts of ulna and radius are viewed from the side, as in Figure 8, it is obvious that radius is anterior to ulna. This relationship makes it possible for radius shaft to cross in front of ulna shaft while the radial head at the elbow revolves to perform forearm pronation. Thus, when the forearm is prone, as shown in Figure 9, the palm faces backwards; when the forearm is supine, as in the anatomical position, the palm faces forward.

Since Biceps Brachii, short and long heads, and Triceps long head originate on the scapula, a brief description of this bone is in order. The scapula is a thin, flat, triangular plate of bone on the upper back side of the rib cage, as shown in Figure 10. The region of scapula (sometimes called shoulder blade) which receives the head of humerus at the shoulder joint is a shallow, concave, oval area called the glenoid cavity (or fossa). Immediately
Figure 8. Lateral View of Right Radius and Ulna [16].

Figure 9. Forearm in Pronation [16].
Figure 10. Bones of the Shoulder Joint [16].
medial to the upper end of the glenoid cavity is a stout, hooked, bony projection, the coracoid process. The short head of Biceps originates from the tip of the coracoid process, while the long head arises from the root of the coracoid process immediately above the upper end of the glenoid fossa. The long head of Triceps Brachii originates from a tubercle immediately below the glenoid fossa in the back.

The insertion of certain wrist flexor and extensor muscles at the metacarpal bones, warrants further description of the wrist and hand bones. The eight small bones that provide wrist flexibility are known as carpal bones. Arranged in two rows, the carpal bones of the proximal row fit into the concavity of radius at the radio-carpal or wrist joint. The five bones in the palm of the hand are the metacarpals, and the bones in the fingers are the phalanges; each finger possesses three phalanges. Flexor Carpi Radialis, a wrist flexor used in this study, inserts at the base of the second and third metacarpal of the anterior surface. Extensor Carpi Radialis, the extensor counterpart of Flexor Carpi Radialis, inserts at the base of the second metacarpal on the posterior surface. The bones of the wrist and hand are shown in Figure 11.

Listed in Primary Anatomy as "muscles acting in the flexion of the elbow joint" are Brachialis, Biceps, Brachioradialis, and Pronator Teres, and as "muscles acting in extension of the elbow joint" are Triceps and Anconeus. As a general rule, however, any muscle which crosses a joint (that is, the joint lies between
Figure 11. Bones of the Right Hand from in Front [16].
muscle origin and insertion) can act on that joint. Several muscles, not listed above, cross the elbow joint. These muscles function primarily as wrist extensors and flexors but could conceivably act on the elbow joint since they cross it. A representative pair from this group of wrist flexors and extensors was considered in this thesis study. These muscles are Flexor Carpi Radialis and Extensor Carpi Radialis Longus. In all, eight muscles were monitored in the experiments: Brachialis, Biceps Short Head, Biceps Long Head, Brachioradialis, Pronator Teres, Triceps, Extensor Carpi Radialis Longus, and Flexor Carpi Radialis. The following information about these muscles is taken from Primary Anatomy.

The great flexor of the elbow joint is Brachialis. (see Figure 12, 13, 15) This muscle lies beneath Biceps Brachii, which usually gets credit for work done by Brachialis. From an extensive origin on the anterior side of the whole lower half of humerus shaft, Brachialis crosses the elbow joint as a wide fleshy muscle. It is inserted by a stout tendon into the tuberosity of the ulna situated immediately below the coronoid process of the ulna (refer to Figure 7).

Biceps Brachii is a two-headed muscle; that is, it originates in two parts from two different places, yet has a common insertion (see Figure 12). Biceps Short Head originates from the tip of the coracoid process on the scapula (above and anterior to the top of humerus head). The long head originates at the root of the coracoid process immediately above the glenoid fossa. This elongated, round tendon traverses the interior of the shoulder joint and runs down
Figure 12. Right Biceps Brachii [16].
Figure 13. Cross-section of the Upper Arm [16].
the bicipital groove of humerus (Figure 6). In the upper half of the arm, the two muscle bellies are separated by a fibrous membrane or septa in the sagittal plane (see p. 26). Half-way down the arm, the bellies fuse and the muscle is inserted by a cord-like tendon into the back of the radial tuberosity (see Figure 7). Biceps Brachii is a forearm supinator, and it aids Brachialis as an elbow flexor against strong resistance.

Brachioradialis is a strong elbow flexor only with heavy loads or during rapid flexion. It also acts to prevent the extension of the semiprone, flexed forearm as demonstrated by a person carrying his overcoat over his arm. Brachioradialis originates over the distal third section* of humerus on the lateral side. It passes down the lateral side of the forearm and is inserted by a long ribbon-like tendon on the radial styloid (see Figures 14 and 15).

Pronator Teres (see Figure 14) also aids in elbow flexion when great force is required, however, it is chiefly a forearm pronator, as indicated by its location. This muscle crosses the anterior side of the forearm upper half by running obliquely down from its origin on the medial epicondyle of the humerus to its insertion on lateral side of radius shaft where the bowing of radius is maximum (midway down the shaft)(refer to Figure 7).

Triceps Brachii, the only important extensor muscle of the elbow, and the only muscle at the back of the arm, has three heads:

(a) Long Head, originating from a tubercle, immediately

* for descriptive purposes, a limb is often sectioned in thirds.
Figure 14. Superficial Muscles of the Right Arm (front) [16].

Figure 15. Muscles of the Upper Arm [16].
below the glenoid fossa of the scapula,

(b) Lateral Head, arising from a 2-3 inches long area posterior to the greater tubercle of the humerus, and

(c) Medial Head, which originates from the back of humerus about two-thirds down on the upper arm.

These three Triceps heads along with long and short heads of Biceps, Brachioradialis, and Brachialis, are shown by cross-section in Figure 13. The three Triceps heads converge in the lower part of the arm and insert by a stout tendon into the posterior edge of the olecranon process on ulna (refer to Figure 7).

Extensor Carpi Radialis Longus arises along with three other wrist extensor muscles from the lateral epicondyle of humerus. This muscle is distinguished from the other three muscles in that a part of the muscle arises from above the lateral epicondyle. It is next to Brachioradialis which runs parallel down the lateral side of the forearm (see Figure 14). Extensor Carpi Radialis Longus inserts on the back of the base of the second metacarpal. This muscle helps extend the wrist and move the hand from side to side at the wrist.

Flexor Carpi Radialis is the wrist flexor counterpart of Extensor Carpi Radialis. It arises, along with two other wrist flexor muscles and Pronator Teres, from the medial epicondyle of humerus, and runs in close proximity with Pronator Teres obliquely across the forearm to insert as a stout tendon on the front of the second and third metacarpals (refer to Figure 15). The tendon of this muscle can be felt slightly lateral to the wrist midline on the
anterior side of the forearm.

All eight muscles discussed are shown in Figures 14 and 15. Also, all origins and insertions are shown on the upper limb in Figures 16 and 17.

2.2.2 Geometric Anatomical Model

In order to determine the muscles involved in the elbow joint, a geometric anatomical model of the joint was developed with the assistance of Dr. M. S. Scharf. Any model is by nature an approximation, and in view of that fact, the following assumptions and implied inaccuracies are recognized:

(a) This model is a two-dimensional model in the sagittal plane (refer to p. 22) containing humerus, radius, and ulna. This statement is an approximation for radius and ulna, which obviously are not in the same sagittal plane. It is also an approximation for several muscles, most notably Pronator Teres, Brachioradialis, Flexor Carpi Radialis, and Extensor Carpi Radialis, as all four muscles run obliquely across the forearm.

(b) Muscles were assumed to originate and insert at points. Since muscles usually insert as tendons over small areas on the bone, this assumption is quite valid for insertion points. Origins, however, typically extend over an area up to several square inches, in which case, the origin "point" was estimated to be the geometric center of the origin area.

(c) Relative distance estimations were made from a "typical" elbow joint personified in form of an anatomy instruction skeleton.
Figure 16. Anterior View of Origins and Insertions of Muscles Studied in this Thesis [16].

Figure 17. Posterior View of Origins and Insertions of Muscles Studied in this Thesis [16].
(d) The two heads of Biceps insert at a common point and three heads of Triceps are shown to insert at the same point, but separately, rather than as fused tendons.

(e) The system is static, i.e., in equilibrium. That is, the sum of the moments about the elbow joint is zero.

(f) Zero torque is sustained by the shoulder joint and/or wrist joint.

(g) Finally, this model yields relative geometric information only. As an initial approximation, it provides insight into the amount of leverage of each muscle as a result of origin and insertion points.

Figure 18 shows the model for an elbow angle of \(90^\circ\) and Figure 19 is for elbow angle of \(150^\circ\).
Figure 18. Geometric Anatomical Model, Elbow
Angle = 90°.

Relative Moment Arm Lengths (arbitrary units)

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Length</th>
</tr>
</thead>
<tbody>
<tr>
<td>Biceps Short (BS)</td>
<td>2.3</td>
</tr>
<tr>
<td>Biceps Long (BL)</td>
<td>2.2</td>
</tr>
<tr>
<td>Brachioradialis (BR)</td>
<td>1.6</td>
</tr>
<tr>
<td>Brachialis (B)</td>
<td>1.3</td>
</tr>
<tr>
<td>Pronator Teres (PT)</td>
<td>0.7</td>
</tr>
<tr>
<td>Flexor Carpi Radialis (WF)</td>
<td>0.7</td>
</tr>
<tr>
<td>Extensor Carpi Radialis (WE)</td>
<td>0.7</td>
</tr>
<tr>
<td>Triceps Long (TL)</td>
<td>0.2</td>
</tr>
<tr>
<td>Lateral (TLA)</td>
<td></td>
</tr>
<tr>
<td>Medial (TM)</td>
<td></td>
</tr>
</tbody>
</table>

Note: subscript "o" designates muscle origin
subscript "i" designates muscle insertion
Relative Moment Arm Lengths (arbitrary units)

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Length</th>
</tr>
</thead>
<tbody>
<tr>
<td>Biceps Short (BS)</td>
<td>1.40</td>
</tr>
<tr>
<td>Biceps Long (BL)</td>
<td>1.25</td>
</tr>
<tr>
<td>Brachioradialis (BR)</td>
<td>0.80</td>
</tr>
<tr>
<td>Brachialis (B)</td>
<td>0.55</td>
</tr>
<tr>
<td>Flexor Carpi Radialis (FR)</td>
<td>0.37</td>
</tr>
<tr>
<td>Extensor Carpi Radialis (ER)</td>
<td>0.35</td>
</tr>
<tr>
<td>Pronator Tertes (PT)</td>
<td>0.32</td>
</tr>
<tr>
<td>Triceps (IT)</td>
<td>0.20</td>
</tr>
<tr>
<td>Long (TL)</td>
<td></td>
</tr>
<tr>
<td>Lateral (TLA)</td>
<td></td>
</tr>
<tr>
<td>Medial (TM)</td>
<td></td>
</tr>
</tbody>
</table>

Notes: Subscript "o" designates muscle origin; subscript "i" designates muscle insertion.

Figure 19. Geometric Anatomical Model (150° Elbow Angle).
CHAPTER III

LITERATURE SURVEY OF PARAMETERS

3.1 Electromyographic Studies of the Elbow Joint

Four electromyographic studies of elbow joint muscles were found in the literature [1,37,38,39]; all but one are qualitative.

In 1957, Basmajian and Latif [1], performed a detailed electromyographic study of both heads of Biceps, Brachialis, and Brachioradialis. Twenty subjects participated in experiments using bipolar concentric needle electrodes. A permanent photographic record of EMG activity was made and then interpreted on a basis of five degrees of activity plus no activity. Movements studied included flexion, extension, maintenance of elbow flexion at $135^\circ$ and $90^\circ$, pronation and supination of the forearm, and flexion and abduction of the shoulder. All movements were studied with no resistance and with a two-pound weight held in the hand. Results pertinent to this thesis are:

1. In all movements, no set pattern of the order in which the muscles' activity began or ended was observed.

2. There was evidence of fine interplay between muscles, with a wide range of response from each muscle from subject to subject.

3. Long head of Biceps showed more activity than short head in the majority of subjects during slow flexion of the forearm and
supination of the forearm against resistance, but little difference was observed in the activity of the two heads during isometric contraction.

(4) Biceps Brachii was generally active during flexion of supine forearm; however, with forearm prone, in the majority of subjects, Biceps played little if any role in flexion.

(5) Brachialis was found to be a flexor of the supine, semi-prone, and prone forearm in slow or quick flexion, with or without resistance.

(6) Brachioradialis did not play any appreciable role during maintenance of elbow flexion and during slow flexion. However, this muscle was quite active in all three forearm positions during quick flexion. Brachioradialis is reserved for speedy movements.

(7) All three muscles differed in their flexor activity in the three positions of the forearm. All act maximally when a weight is lifted during flexion of semiprone forearm.

Using the same apparatus and data interpretation methods, Basmajian and Travill [37] in 1961 made a study of Pronator Teres during elbow flexion. They concluded that Pronator Teres contributes to elbow flexion only when resistance is offered to the movement. No activity in this muscle was observed during unresisted flexion in all three forearm positions.

Fine copper wire bipolar electrodes were used by Pauly, Rushing, and Scheving [38] (1967) to study Triceps, Biceps, Anconeus, Brachialis, and Brachioradialis muscles. Eighteen subjects participated. During isotonic flexion (see p. 78) of the elbow with a five-pound
weight in the hand, activity in the Brachialis preceded that from the Biceps Brachii which in turn preceded that from Brachioradialis. The authors state complete agreement of their results with Results no. 4, 5, and 6 from Basmajian and Latif, as stated on previous page. The EMG was rectified and integrated, but results were presented qualitatively.

In 1969, Bankov and Jorgensen [39] used surface electrodes to record EMG activity in Biceps Brachii and Brachioradialis of three subjects during maximum isotonic and isometric contractions (see p. 78). Torque was applied and measured by a motor-driven strain gage dynamometer. Maximum isometric contractions were made for forearm either supine or prone and elbow angle $90^\circ$, $45^\circ$, or $135^\circ$. Three subjects were used. The EMG activity was integrated and both torques and IEMGs were expressed as percentages of the maximum isometric values with an elbow angle of $90^\circ$ and the forearm supine.

In general the authors concluded that isometric maximum torque of the elbow flexors depends on the elbow angle and on the forearm position. Results of the isometric study are as follows:

1. IEMG from Biceps for prone forearm was 50 per cent of that obtained with supine forearm.

2. No significant differences in IEMG for Biceps for different elbow angles.

3. No significant differences between IEMG of Brachioradialis with different elbow angles or forearm positions.

4. Torque produced with forearm prone was 85 per cent of that obtained with forearm supine.
(5) Maximum torque at elbow angle of 45° and 135° was only about 75 per cent of the value obtained with 90°.

As revealed by this literature review, the elbow joint has been neglected in EMG muscle function studies. Most of the information that is available pertaining to this complex joint is qualitative in nature. Also, in none of these investigations is the complete elbow joint musculature studied.

3.2 Studies of EMG-Load Relationship

A literature search for previous studies of relationships between the electromyogram and muscle tension has revealed sixteen such investigations, using a variety of muscles, loading methods, EMG retrieval and treatment methods, etc. Except for one paper [40] which proposed a logarithmic relationship in poliomyelitis (a nerve disease affecting muscle) muscle, two schools of thought seem to predominate. One suggests a linear relationship and the other a nonlinear, or more specifically, quadratic function. Because of lack of sufficiently sophisticated measuring and data reduction equipment coupled with tendencies toward oversimplification in original studies, one might logically postulate that, in general, the earlier studies probably report linear relationships, while later studies reveal a quadratic relationship. In fact, Inman, et al. in 1944 [40] reported that "experiments have determined a direct relation between tension developed in a muscle and action current potential, . . . this relation, however, is not a linear one, the precise relationship being the function of the square, and is expressed as a quadratic

The differences in results proposed by these two schools of researchers may be explained by grouping studies according to muscles examined and by comparing experimental methods used to study each muscle. All studies were made with voluntary contraction, and all involved isometric contraction, though three studies [10,3,41] also included information regarding isotonic contraction (see p. 78).

Triceps muscle was investigated by three groups: Inman, Saunders, and Abbott, 1944 [40], Inman, et al., 1952 [10], and Lenman, 1959 [42]. Inman, Saunders, and Abbott included very little detailed description of the experimental method. In a study of the shoulder joint, "electrodes were implanted in the muscle" of 15 normal subjects and eight poliomyelitic subjects. A plot of action current potential, in arbitrary units, versus tension, ranging 0 - 10 kg, is presented for all subjects studied. The authors state that the relationship for normal muscle is quadratic, and that for poliomyelitic muscle is logarithmic. Later, in 1952, Inman, et al. [10] states that "no one, as far as we are aware, has demonstrated existence of a simple relation between EMG and tension."

In a study of eleven amputees, using three types of electrodes (skin, coaxial needle, and wire), action potentials and muscle tensions from a strain gage dynamometer were recorded simultaneously. The
action potentials were rectified, integrated and plotted as a
function of time. The tension was also plotted versus time.
Contractions varied from slight effort, to maximal effort, and then
decreased. Inman, et al. concluded that the "integrated electro-
myogram closely parallels tension in human muscle contracting iso-
metrically," [10]. As a result of isotonic contraction
studies, they further state that "no quantitative relation between
EMG and tension exists when a muscle is allowed to change in length"
[10]. Lenman [42] used surface electrodes, one subject on
five different occasions, and a dynamometer loading method with elbow
angle of 90°, to obtain five graphs, 10 data points each, showing
a linear relationship between rectified, mean voltage EMG (0.1-2mv)
and tension (2-20 lb). Correlation coefficient with a straight line
was 0.985, and regression coefficients* varied from 0.1 - 0.15 mv/lb.

Biceps Brachii was included in the papers by Inman, et al. [10]
and Lenman [42] with the same results as discussed for Triceps.
Eight other papers deal with Biceps: Travis and Lindsley, 1931 [43],
Snyder, 1953 [44], Knowlton, 1956 [45], Liberson, et al., 1962 [11],
Devries, 1965 [4], Devries, 1968 [46], Zuniga and Simons, 1969 [6],
[43] and [44] by stating linear equations which he considered
representative of their reported relationship between EMG and tension:

\[ A = K_1 + K_2 N \] [43], forearm flexor muscles, wrist
flexors

* In the equation for a straight line, \( y = mx + b \), the regression coefficient
is \( m \).
\[ A = K_1F \]

where \( A \) = amplitude of electromyogram

\( F \) = force or load on muscle

\( K \) = constants

Knowlton and associates [45] measured average maximum peak-to-peak EMG voltage (0.1 - 2 mv) with surface electrodes over 56 Biceps muscles (short head), as weights up to 64 oz in 14 steps were lifted by the forearm. The resulting graph shows EMG to be a linear function of load, with regression coefficient of 0.107 mv/lb. Note the close agreement of this figure with Lenman’s [42] coefficient (0.1 - 0.15 mv/lb), for Triceps. In a study of isometric exercises, Liberson, Dondey, and Asa [11] obtained a plot (6 data points) of integrated EMG in arbitrary units (surface electrodes) versus tensiometer readings up to 30 pounds. This graph, assumed to be from one subject, is described as linear, though it appears that a higher degree equation might be a better approximation. Linearity was also the conclusion of DeVries and associates both in 1965 and 1968. The first study [4] examined low levels of muscle contraction (EMG range of 0 - .01 mv). The forearm was midway between prone and supine, and elbow angle was 75°. The EMG was recorded from surface electrodes over the belly of right Biceps Brachii. The composite plot of 19 observations on one subject demonstrates linearity. Also observed, was lack of hysteresis effect of loading and unloading the muscle. DeVries' second study [46] reported linearity for EMG
range 0 - 0.7 mv and load range 0 - 25 lb in three subjects for the elbow flexor muscle group. This study was experimentally similar to the study in 1965.

The previous eight papers discussed have reported a linear relationship between EMG and load in Biceps. Zuniga and Simons [6] in 1969, suggested that the relationship is in fact nonlinear, probably quadratic. They report two plots: one composite of 42 plots from one subject, and the other a composite of 54 plots from nine subjects. The EMG (0 - 1.0 mv) was recorded from surface electrodes, electronically averaged, and input to an X-Y plotter. Load, ranging from 0-18 lb, was measured by a strain gage tensiometer and also input to the X-Y plotter. The authors conclude that the parabola provides a much better fit than the straight line. The support their conclusion by stating the mean square deviation about regression of only 0.0004 for the quadratic as compared to 0.0011 for the straight line.

Zuniga and Simons suggest several possible explanations for previous reports of linearity. Two studies [4,45] limited investigation to low levels of muscle contraction in which the curvature of the graph is rather obscured. In fact, at levels less than 20 per cent maximum contraction, the relationship is indeed quite linear. All of the previous investigators, except Snyder [44], examined the data for linearity but did not mention a check for nonlinearity. Data reported by both Liberson, et al. [11] and Knowlton, et al. [45] appears to be nonlinear and quite possibly quadratic. Zuniga and Simons state that half of Knowlton's pub-
lished plots appear to fit a non-linear regression at least as well as a linear regression. This statement suggests that possibly investigators were too demanding in their criteria for improved fit, however, only three of the papers [45, 42, 44] reported a regression coefficient and only one of those [42] reported a correlation coefficient.

Komi and Buskirk (1970) [53], in a study of reproducibility of EMG measurements with intramuscular and surface electrodes, report a significant non-linearity in the relationship between integrated EMG recorded from surface electrodes and muscle tension. Details of the experimental methods are discussed in Section 3.3 dealing with electrode position. Their results are presented as a graph of IEMG (0.2 - 1.4 mv) versus isometric tension in per cent of maximum tension. The graph is a composite plot of 29 subjects each tested on two different occasions. The following quadratic equations are given:

\[
\text{Day 1: } \text{EMG} = 7.75 + 5.9(\text{Tension}) + 0.10(\text{Tension})^2
\]

\[
\text{Day 2: } \text{EMG} = 22.32 + 2.32(\text{Tension}) + 0.12(\text{Tension})^2
\]

Regression coefficients are given as 0.73 for Day 1 and 0.69 for Day 2. There was a three day interval between test days.

Two investigations were made of the wrist flexors. Previously mentioned in the discussion of literature pertaining to Biceps, the study by Travis and Lindsley (1931) [43] was referenced by Nightingale [5] who deduced a linear relationship from data reported by Travis and Lindsley for the forearm flexor muscles and wrist flexors.
Travis and Lindsley [43] measured EMG activity in wrist flexors with needle and surface electrodes, while the subject exerted various intensities of wrist flexion indicated on a hand dynamometer. The EMG activity was recorded on film and amplitude of action potentials was measured in millimeters. Twenty-five subjects participated. The authors presented results as a plot of amplitude in millimeters (0-10) intensity in kilograms (0-70), which appears possibly non-linear. The results are somewhat implied to be linear in the following description: "There is an increase in the amplitude of action currents with an increase in the intensity of contraction," (p. 379). Nightingale also referenced Dempster and Finerty [48]. Dempster and Finerty used surface electrodes placed over muscle motor points for 10 wrist flexor muscles. With seven subjects, the elbow angle at 90°, the forearm horizontal and midway between pronation and supination, weights, which were multiples (2, 3, 8, 16, 32, 64 times) of the hand torque (3.2 kg-cm), were applied as torque to the wrist joint. The results, presented as plots of muscle potential times lever arm length versus gravity torque, appear nonlinear, possibly quadratic or logarithmic.

Calf muscle (gastrocnemius and soleus muscles) in the leg was used in five studies [2, 41, 3, 5, 15]. Only one [5] suggested nonlinearity, though another [3] reported nonlinearity for isotonic contraction. Also worth noting is a reference by personal communication to Edwards and Hardy, by Bigland and Lippold [41] stating "certain muscles including Tibialis Anterior follow a quadratic
relation rather than a linear one," ([4]p.214). Bigland and Lippold, using eighteen subjects studied isotonic contraction with a load dynamometer varying 0 - 30 pounds. EMG was recorded from both needle and surface electrodes, and then electronically rectified and integrated. Results were presented as five-data-point plots, each data point representing the mean of ten repetitious observations from one subject. These plots for both electrode types, appear to be linear (correlation coefficient 0.88 - 0.93) for constant speed of flexion ($\theta/dt = $ constant, for ankle joint angle $\theta$) for both muscle lengthening and muscle shortening. The slope of the plot increases for increasing flexion speed (0.1 - 1.0 rad/sec). A previous study by Lippold [2] in 1952, indicates a linear relation for isometric plantar flexion with linear regression correlation coefficient of 0.93-0.99. The EMG recorded from surface electrodes was mechanically integrated with a planimeter and reported in arbitrary units. The load, applied with a dynamometer, ranged from 2-20 pounds. Regression plots were presented for ten subjects, each plot consisting of 12 data points.

As preface to their discussion, Close, et al. [3] state, "Although the density of the interference pattern of the EMG appears to increase with strength of contraction, a quantitative relationship has not yet been demonstrated," (p. 1208). As an alternative to integration of the raw EMG signal, they proposed to count the number of action potential counts per unit time. They support this alternative by postulating, "If the motor unit conforms to the all or none law, the total number of action potentials may
be of greater significance than total electrical activity, which is a function of amplitude and duration, as well as action potential count." (p. 1208). Recorded from internal copper wire electrodes (0.006 in diameter), 1 inch apart, the action potentials were counted by an electronic counter. An integration was performed on the EMG as a comparison with the action potential count. In an effort to directly measure the tension developed in Soleus muscle, a force gage was attached to a pin inserted in the tibia very close to the muscle's insertion point. EMG activity in Peroneal and Posterior Tibial muscles was suppressed with practice by the subjects, and the knee was flexed to 90° to eliminate any possible participation of Gastrocnemius (calf muscle). Total counts (up to 1800) of 10-second isometric contractions at six different muscle lengths were plotted against tension up to 80 pounds, with four or five data points comprising each straight line. The results are described as linear, although the data is somewhat scattered, and no attempt was made to fit the data to nonlinear functions. Action potential is stated to decrease with increasing muscle length. Comparing action potential counts and integrated EMG, the authors claim that "the integrated curve of total electrical activity may be replaced by the action potential count of the EMG," (p. 1220). Results for isotonic contraction indicate nonlinearity.

Nightingale [5], in 1960, using surface electrodes in a posture study, reported results as a plot of the rms mean amplitude value of EMG (0 - 0.07 mv) recorded over a 30-second period, versus tension (0-95 lb). This plot was constructed from data recorded from
one subject on two separate occasions. To load the Soleus muscle, the subject was placed with a foot on each of two scales, and asked to lean slightly farther forward or backward from a comfortable stand-at-ease position. Using a geometric model of the ankle joint, calf muscles, and line of weight, the force in the calf muscles was calculated given the displacement of the line of weight. The two experiments compared favorably, both being linear up to the upper (greater than 60 lb) values of tension, at which values the plot curves upward. Nightingale, referencing Edwards and Lippold [50] (1956) and Snyder [44] (1953), postulates that these increasing values of EMG at the higher loads may be a result of fatigue.

In the most recent (1971) study found, Gottlieb and Agarwal [15] recorded EMG activity in Soleus muscle with surface electrodes. The subject pursued a computer-controlled target with a marker which he moved by varying the isometric torque exerted by his foot strapped to a rigid plate. The computer recorded the foot torque and filtered EMG and plotted this information as a function of time. These experimental results were compared with results generated by linear model simulation on an analog computer. Input to this model was actual filtered EMG, and output or model response was simulated foot torque.

Model response compared quite favorably with experimental results, and the authors concluded that there is a "definite functional relationship between Soleus EMG and foot torque during voluntary isometric plantar flexion . . . that relation being a simple linear second-order differential equation with constant
coefficients," ([15], p. 343).

This literature search demonstrates that the EMG-Tension relationship has been the subject of a considerable amount of research. Investigators seem convinced that the electromyographic signal contains information regarding the load sustained by a muscle. However, it is ironic that in none of these studies was the tension in a muscle or the load sustained by a muscle measured. In fact, such a measurement would be most difficult to make. Instead, the applied and measured load was sustained by several muscles, from only one of which was the EMG activity monitored. In this thesis study, an attempt was made to measure the EMG activity in all muscles sustaining the load.

3.3 Experimental Reproducibility

In any quantitative electromyographic study, one of the most difficult-to-control variables is electrode position. Since surface electrodes pick up action potentials from a large number of motor units, a reasonable assumption is that electrodes placed over the belly of a muscle pick up EMG activity representative of the entire muscle. With intramuscular electrodes, which pick up EMG activity in a small localized muscle area, such assumption is questionable due to the nonhomogeneous nature of muscle structure and muscle control (see Chapter II). In this thesis study, attempts to control intramuscular electrode position included using needles of the same length for all subjects and having the same anatomist insert all electrodes with care to maintain consistency of insertion.
Even with these precautions, electrode position in the muscle no
doubt varies to some extent between subjects. Electrode position
also varies during an experiment, for electrodes tend to migrate
into the tissue with repeated muscle contraction.

Recently three investigations have been made to determine the
effects of electrode position of experimental reproducibility in a
quantitative study. One of these studies [51] was concerned with
intramuscular electrodes, another [52] used surface electrodes,
and the other study [53] involved both types of electrodes. Also
a recent study [54] was made to investigate electrode displacement
during repeated muscle contractions.

Jonsson and Reichman (1967) [51], in a series of experiments
using Brachioradialis in isometric and isotonic contraction, studied
reproducibility with unipolar needle electrodes, concentric needle
electrodes, and wire electrodes. The wire electrodes used were
very nearly the same as used in this thesis study (see Section 4.3).
Detailed results for wire electrodes only will be discussed.
Suffice it to say, however, that with all three electrode types, EMG
activity was definitely affected by electrode position. For six
subjects EMG activity was recorded about five minutes after the electrodes
had been positioned, and then again about 15-20 minutes later.
Between recordings, the muscle was at rest. For all subjects, the
experiment was repeated on another occasion. The interval between
the two experiments varied from three to seven days. The electrodes were
inserted in approximately the same place each time. Muscular
activity was recorded with the subjects standing with their back and
left upper arm resting against a wall and their elbow joint flexed at 90°. The subject held a 250 ml beaker filled with water in his left hand to maintain the semiprone forearm position. No force or torque other than the beaker of water was applied to the joint. The degree of muscular activity was assessed according to the frequency of the action potentials. No exact calculation of the number of action potentials per unit time was made. Four subjective degrees of activity were used:

0 = no activity
1 = slight activity
2 = moderate activity
3 = marked activity.

The results of this study are as follows:

(1) The difference in the degree of activity in the two recordings made on the same occasion was slight. In one subject activity in one experiment was moderate in the first recording and slight in the second. In two other subjects, there was "slight"\* difference between recordings.

(2) The differences in degree of activity in the various experiments were somewhat greater than differences in recordings made on the same occasion.

a. Recordings made five minutes after electrode insertion: In three subjects there were slight differences between experiments.

\* In those cases in which differences in intensity of activity were on the border between two degrees of activity, the difference has been described as "slight".
b. Second recordings: In two subjects, there were slight differences, and in one subject there was a distinct difference (slight activity on one occasion, and moderate activity on another).

(3) No appreciable differences in the degree of EMG activity were observed when comparing the different types of electrodes. The authors attribute differences in EMG activity to three factors pertaining to electrode position:

(1) Alterations of electrode position during muscle contraction. Substantial changes in activity can take place even if the location of the electrode is only altered by a few mm.

(2) Distance between the two wires of bipolar wire electrodes. Changes of electrode position during muscle contraction will probably alter this distance.

(3) Intramuscular bleeding. Minor bleeding around the electrode probably always takes place when intramuscular electrodes are used. A hematoma at the recording surface of the electrode increases the distance from the active motor units, resulting in few motor units being recorded.

Regarding reproducibility in quantitative EMG studies, the authors concluded that "the 4-point scale described here would appear to represent the highest degree of accuracy that is practical," ([51]p.89).

Komi and Buskirk (1970) [53] investigated reproducibility of EMG measurements with both intramuscular wire electrodes and surface
electrodes. They mention that repeatability of measurements is "perhaps an area, which has been most neglected in electromyography. Duplicate observations have seldom been made, or if made not reported," ([53]p.357). Experimental Method (pertinent information): Biceps Brachii in submaximal and maximal isometric contractions; wire and surface electrodes; elbow angle 90°; forearm supine; muscle tension sensed with strain gages; eight subjects; three separate testing days with one day of rest between them. Contractions, each a percentage of maximum tension (20 per cent, 40 per cent, 60 per cent, 80 per cent, 100 per cent) were performed for four seconds each, with thirty second rest intervals. The entire test was administered twice on each day with 10 minute rest between. Reproducibility of EMG measurements with surface electrodes was much better than with inserted wire electrodes. Reproducibility was better between recordings made during the same test than between those recordings made on different occasions.

The authors offer several explanations for poorer reliability using wire electrodes:

1. Difficulties in re-inserting the wires exactly into the same place.

2. Limited pick-up area of the wire electrode. "Basmajian (1967) suggested that the wire electrode technique provides a pick-up area which covers the whole muscle. We feel that this is not the case for the whole or even for the different heads of the Biceps muscle," ([53]p.366).
(3) "It seems illogical to expect excellent repeatability of measurement in such a complex biological phenomenon as whole muscle activity. It is impossible to control incoming nervous activity for voluntary contractions. Thus the input signal varies," ([53] p. 366).

Zuniga, Truong, and Simons (1970) [52] investigated the effects of skin electrode position on the amplitude of averaged electromyographic potentials during isometric flexion of the elbow. EMG activity in Biceps Brachii was recorded with Beckman monopolar surface electrodes, identical to those used in this thesis (see Section 4.3). Isometric tension was measured by a strain gage dynamometer. Three subjects participated. The forearm was midway between prone and supine, and elbow angle was 90°. The experiment was performed in two parts:

(1) longitudinal position test, in which nine skin electrodes were placed equidistant along the longitudinal axis of Biceps Long Head, and

(2) transverse position test, in which five electrodes were placed equidistant across the belly of the long and short heads of Biceps.

The subject exerted a five-second, steadily increasing isometric flexion to maximum effort. Ten runs were made in one session with each subject. Results were as follows:

(1) The plot of averaged EMG potentials at maximum isometric tension versus longitudinal electrode position follows a bell-shaped curve with a rather prominent peak corresponding to electrode
position approximately at mid upper arm.

(2) A similar plot for transverse position shows peak at the position of division between short and long heads of Biceps, a minimum at position over long head, and a slight rise at position dividing Biceps and Brachialis.

(3) Electrode position showed a definite effect on the electromyogram-tension relationship, though this relationship remained parabolic for all electrode positions. The authors conclude that "skin electrodes can pick up EMG signals from quite distant muscles and therefore have significant limitation in differentiating the respective activities of underlying muscles. The use of fine bipolar intramuscular electrodes appears to be the safest way to ascertain the presence or absence of activities in a certain muscle," ([52]p.271).

Jonsson and Bagge (1968) [54] investigated the displacement and deformation of wire electrodes identical to those used in this thesis. Little is known about the effects of muscle contraction on wire electrodes. The authors state that muscle contraction tends to displace the wires in directions parallel to muscle fibers. This displacement, of course, changes the electrode pick-up area. Deformation of the electrodes was observed during experiments for this thesis in that curled and twisted wires were withdrawn from muscles after an experiment. Effects of such deformation upon EMG activity recorded is unknown. In Jonsson and Bagge's study, bipolar wire electrodes were placed in the Gastrocnemius (calf) muscle in 23 subjects. The subjects then walked around a lecture
room for 20 minutes. The length of that part of wire outside the skin was measured before and after the walking exercise. The wires were then carefully removed and deformation observed. Care was taken to avoid deforming the wires when withdrawing them from the muscle. In order to study whether wire deformation is due to muscle contraction, eighteen pairs of wire electrodes (six in each of three subjects) were inserted into a relaxed muscle and immediately withdrawn again. The wires were then examined under a microscope. Results of this study were as follows:

1. In all cases, the distance from the skin to the external end of the wire was less after 20 minutes walking. The wire had been drawn in a mean of $14.8 \pm 0.61$ mm. The wires were pulled in the largest distance under the skin during the first contractions of the muscle.

2. Deformation of the wires was noted in all cases after 20 minutes walking. They were found to have one or several kinks. The wires that were inserted into a relaxed muscle and immediately withdrawn displayed insignificant deformation. That result indicates that muscle contraction definitely causes electrode deformation. The authors conclude that "electrode displacement during a kinesiologic study may thus produce effects that can be misinterpreted as a change in the degree of contraction of the muscle," ([54]p.344).

All four of these investigations indicate that experimental reproducibility in quantitative EMG studies is extremely difficult to achieve. In this thesis study, a check of experimental reproducibility was made for both day-to-day reproducibility as well as
reproducibility within an experiment. The literature definitely demonstrates the need for particular attention to electrode placement, especially with intramuscular electrodes. The other major cause for poor experimental reproducibility, simply variations within a biological system, is recognized and accepted as virtually impossible to define and/or control.

2.4 Muscle Fatigue

Electromyographic activity in a muscle is definitely altered by fatigue. Since the EMG-tension relation is also affected by fatigue, muscle fatigue is considered an important parameter in experimental designs for this thesis. A literature search was conducted to attempt to answer the following questions:

a. What specific changes are produced by fatigue in the electromyogram and in the EMG-tension relation?

b. What are the limits within which the EMG-tension relation holds? What limits on load and work rate determine the onset of muscle fatigue?

c. What rest period is required to restore fatigued muscle to rested state?

3.4.1 Question a.

Several studies were found which investigated changes in the electromyogram produced by fatigue. Cobb and Forbes [55] in 1923 established that in rapid fatigue there is an increase in the amplitude of the summated action potentials. After referencing this study, Knowlton, et al. [56] in 1950, investigated the relation of load and
work rate to the fatigue electromyogram. EMG activity was recorded from Biceps Brachii with surface electrodes, one placed over the motor point of the Biceps Short Head, the other at the tendon end of that muscle. With forearm supine, six subjects lifted a hand-held weight through an elbow angle range of 95-165° at a given rate to the point of subjective fatigue. Action potentials were displayed on an oscilloscope and recorded on film. The EMG record was analyzed for the number of contractions to fatigue, and then the action potential voltage was measured for the first three contractions and three contraction each about the points of 20, 40, 60, 80, and 100 per cent of the lifts to fatigue. The experiment investigated the fatigue electromyogram as a function of two parameters:

a. Relation of the fatigue electromyogram to load: The work rate was constant at 58 lifts per minute. Six subjects each lifted four loads (5, 10, 15, 25 lbs) on four different occasions with two days rest between experiments. Observed for the first three loads was an early increase in action potential amplitude, which became greater as the fatigue point was reached. For the 25 lb load, however, amplitude increased to a maximum at 60 per cent of fatigue level and then decreased slightly.

b. Relation of the fatigue electromyogram to work rate: The load was constant at 15 lbs, while three work rates (29, 58, 116 lifts/min) were used. For 29 lifts/min, the EMG amplitude appeared to oscillate, increasing initially, then decreasing, increasing again, decreasing, and finally increasing to maximum value at maximum number of lifts performed. The amplitude for 58 lifts/min
steadily increased. The results for 116 lifts/min are similar to those for 29 lifts/min with amplitude increasing, decreasing, and finally increasing to maximum value. In all three cases, however, the maximum action potential amplitude was attained at the highest number of lifts to subjective fatigue. The authors conclude that there is an intermediate load and work rate where the EMG voltage increase is maximal.

In agreement with Knowlton, et al., Scherrer and Bourguignon [57] (1959) observed an increase in the integrated EMG in an experiment in which a constant load was supported at a constant height. Their results, presented as a plot of IEMG (integrated) versus time (duration of work) show slight increase in IEMG up to 50 - 60 seconds, at which time the increase is more pronounced.

The findings of Lippold, Redfearn, and Vučo (1960) [58] also support the two studies just discussed. Using both surface electrodes and concentric needle electrodes, these researchers studied fatigue EMG activity in several muscles. They found that with a constant tension, maintained for several minutes, both frequency and amplitude of the action potential increased. A plot of IEMG versus time, for which a constant isometric contraction of 10 kg was maintained in Gastrocnemius (calf) muscle, is curvilinear with a sharp increase in IEMG after three minutes. An interesting phenomenon is demonstrated by the also nonlinear decline in IEMG activity in Extensor Digitorum Communis (finger extensor muscle) supporting 200 grams on the forefinger. After five minutes the activity had almost ceased, yet the weight was still supported. Needle electrodes showed that the
load had been transferred to another muscle, Extensor Indicis Propius.

With surface electrodes, DeVries (1967) [59] recorded activity in knee extensors and elbow flexors while holding continuous isometric contractions which were 30, 40, and 50 per cent of the measured maximal voluntary contraction. Results presented as plots of mean EMG voltage versus time duration of contraction show linear increase in EMG with increasing contraction time for all loads, with the slope increasing with increasing load.

Edwards and Lippold (1965) [50] investigated the relation between the electrical activity and the tension of isometrically contracting muscle under fatigue conditions. EMG activity in Soleus (calf) muscle was recorded with surface electrodes and electronically integrated. Tension (range: 9 - 15 per cent maximum subject capability) was measured with a strain gage dynamometer. The loading sequence was performed twice, before and after muscle fatigue. To fatigue the muscle, continuous contraction to about 25 per cent of maximum was maintained for five minutes. This entire experiment, two loading sequences and fatiguing contraction, was repeated after a 15 minute rest. Eight subjects performed 15 experiments which produced comparable results. Results are summarized below:

(1) During the fatiguing contraction, the electrical activity always decreased slightly during the first minute and then gradually increased, with some oscillation, for four minutes. During the last minute of contraction, activity increased sharply without
oscillations. These results support conclusions of the five papers discussed [56,57,58,59,60].

(2) When the loading sequence was repeated with fatigued muscle, the integrated EMG was higher, as might be expected from the results just previously stated. The IEMG-tension relationship remained linear, with straight line correlation coefficient of 0.985. However, the slope of the line increased significantly for fatigued muscle.

Findings by Lenman in 1959 [60], agree with Edwards and Lippold. Isometric tension was recorded from a strain gage dynamometer, and mean EMG voltage in 12 subjects was monitored from surface electrodes over Triceps muscle. A typical plot of mean EMG voltage versus tension (10-30 lbs) demonstrated linearity for experiments performed before fatigue as well as for experiments performed after a five minute sustained isometric contraction. The relation for fatigued muscle, however, was of steeper slope.

To summarize the answer to question (a), as indicated by the literature,

(1) Muscle fatigue, produced by sustained isometric contraction at constant load, causes increase in amplitude of electromyographic activity.

(2) The amount of increase in EMG activity varies as the sustained load is increased.

(3) The slope of the EMG-tension relationship increases in fatigued muscle.
3.4.2 Question b.

In a sustained isometric contraction, the time at which fatigue appears to affect the electromyogram appears to be a function of load and work rate. The experiments for this thesis were designed to avoid muscle fatigue entirely and thereby stay within those bounds where EMG-tension relations for rested muscle are applicable. The isometric contractions involved 12 loads which were approximately 5.5, 11.0, 16.5, 22.0, 27.5, 33.0, 38.5, 44.0, 55.0, 66.0, 77.0, and 88.0* per cent of maximum isometric contraction. Each load was sustained approximately four-five seconds, for a total contraction time of 48-60 seconds. Between every other contraction was a two minute rest period, and between every four contractions was a five minute rest period. The four-five seconds each load was established as a minimum based on suggestions by Dr. J. V. Basmajian. Dr. Basmajian stated that each load should be sustained for at least two or three seconds before recording the one second of data in order to insure neuromuscular system stability at each load; that is, after two-three seconds, the appropriate number of motor units would be recruited and firing frequency established. The maximum total contraction time of one minute was established by results of four investigations.

Edwards and Lippold [50] presented a plot of integrated EMG versus time duration of isometric contraction which shows a sharp 

* estimated as percentage of maximum isometric contraction load as stated in reference [61] as 71.3 kg or = 32 lbs. "mean maximum force" Typical loads in these thesis experiments were 0.5 - 8 divisions with calibration indicating 3.5 lbs/div.
increase in EMG after four minutes, but only slight oscillatory changes up to that time. Scherrer and Bourguignon [57] also obtained a plot of integrated EMG versus time in sustained isometric contraction. This plot shows no change in EMG activity until one minute at which time a gradual increase in EMG activity begins followed by sharp increase at five minutes and an even greater EMG increase at ten minutes.

In two recent studies by Lloyd [61, 62] isometric contractions were maintained at 25 per cent and 30 per cent of maximum voluntary contraction, while EMG activity in several muscles, including Biceps, was recorded with surface electrodes. The subjects were asked to rate the pain in the muscles on a five-point scale, the minimum value being noticeable pain and the maximum value unbearable pain. Results were presented from both investigations as two graphs:

(1) a plot of time (0-200 sec) versus subjective levels of pain (1-5), and

(2) plot of mean EMG amplitude (1/4 mv) versus subjective levels of pain.

By combining the two graphs into one plot of mean EMG amplitude versus time, the results indicate a slight increase at 60 seconds and a sharp increase at 80-90 seconds. These results, as well as those obtained by Edwards and Lippold [50] and Scherrer and Bourguignon [57], indicate that a one-minute sustained isometric contraction at 33 per cent of maximum contraction* is not long.

* Since all four investigations involved constant load isometric contractions, for comparison purposes, an average value of the 12 loads in this thesis experiments was computed to be 33 per cent of maximum isometric contraction.
enough to fatigue the muscle.

### 3.4.3 Question c.

The rest periods were established from results of a non-electromyographic fatigue study by Mundale [63], in which strength of hand grip as indicated by a dynamometer was monitored during ten minute repetitive isometric exercises. Decrease in strength was most evident in exercises requiring exertion greater than 20 per cent of maximum strength. A four-minute rest period was shown to restore a fatigued muscle to rest strength capability. In view of these results, the two minute rest period after two one-minute contractions and the five minute rest period after four one-minute contractions were considered sufficient to eliminate fatigue problems from this thesis experiment.

### 3.5 Miscellaneous Parameters

Mentioned in the literature are several other variables which warrant consideration. In a study of isometric elbow flexion, Holt, et al. [64] in 1969, investigated the influence of antagonistic contraction (see p. 132) and head position on EMG activity in elbow flexor muscles. The authors referenced Preo [65] (1967) who found that submaximal isometric contraction of the antagonists (elbow flexors) produced greater agonistic (Triceps) force. Also mentioned are Hellebrandt, et al. [66] and Waterland and Hellebrandt [67], who demonstrated the effects of head position on EMG activity in wrist flexors and extensors. Holt further explored the influence of four independent variables upon the total load capability of the elbow
flexors and upon the EMG activity in Biceps Brachii and Brachioradialis muscles. The four independent variables were:

(1) Maximum isometric contraction of the antagonists (to elbow flexion, i.e., Triceps), with head in anatomical position (refer to p. 22), prior to the maximum isometric contraction of the agonists (elbow flexors).

(2) Maximum contraction of agonists only, with head in anatomical position.

(3) Maximum contraction of agonists only, with head turned completely to the right.

(4) Maximum contraction of agonists only, with head turned completely to the left.

Three normal subjects participated. EMG activity in Biceps Brachii and Brachioradialis was recorded with surface electrodes during maximum isometric contraction. The subject was supine (on his back) his left elbow at 90° with forearm midway between prone and supine. Load was indicated by a dynamometer during maximum isometric contraction. Results were as follows:

(1) Reversal of antagonists prior to maximum isometric contraction of agonists produced greater load from the agonists. However, only one subject had increased EMG activity.

(2) Head position definitely affected load output from the agonists (elbow flexors). Anatomical head position produced greatest load scores, and left position load measurements exceeded those of right head position.
In summary, EMG activity as well as load production in the elbow flexors during isometric contraction is significantly influenced by head position and prior antagonistic contraction.

Another parameter investigated by Okamoto [68], is the direction of load application. During isometric flexion of the arm in the sagittal plane (see p. 22) with slight variance in the direction of resistance, EMG activity in 11 muscles, including Triceps Long head and Biceps Long head, was recorded. Results indicated wide variations in the discharge patterns of several muscles, including Biceps Long head, as the direction of resistance is very slightly changed. This information implies need for subject control of load application direction.

A less tangible and virtually uncontrollable variable in the inconsistency between different specimens of all living organisms. Several researchers have stated results to support this inconsistency. Miles, et al. [69] and Sullivan, et al. [70] reported considerable variation in Biceps Brachii in the same subject as well as in different subjects during elbow flexion tests. Okamoto in the study just discussed, stated: "The flexion of the arm among the most fundamental movements is a very simple movement. But, even though the same movements were performed by the well-trained subjects, the myograms obtained demonstrated some differences," ([68]p.113). In 1957, Basmajian and Latif [1], after detailed electromyographic study of Biceps, Brachialis and Brachioradialis, stated:
What is more striking, however, is the wide range of response (among different subjects) from any one muscle. Thus, although a general trend may be described, there is rarely any unanimity of action. For example, the Brachialis is generally markedly active during quick flexion of the supine forearm, but in one of our subjects it was completely inactive. These findings re-emphasize the general biological principle that there is a range of response in any phenomenon. It would seem that anatomists and clinicians have taken too little heed of this wide range of individual pattern of activity in something even so simple as elbow flexion, ([13]p.173).
CHAPTER IV

EXPERIMENTAL EQUIPMENT

Equipment used in the experiments includes the following:

(1) Load Device, by which the subject applied an isometric measured torque to his elbow joint.

(2) EMG monitoring equipment, including the EMG amplifiers, the eight-channel oscilloscope to monitor the EMG activity in each muscle, and a 14-channel Hewlett-Packard recorder to record the EMG activity plus the torque measured by the Load Device.

(3) Electrodes used to pick up the EMG activity in the muscles.

Each piece of equipment is described in detail in the following sections of this chapter. An explanation of how the equipment was used in the experiments is found in Chapter V.

4.1 Load Device

An apparatus was designed to enable a subject to isometrically apply a measured torque to his elbow joint. "Isometric" refers to the state of muscle during contraction. Muscle contraction is accomplished by muscle fibers which shorten to approximately two-thirds of their resting length [13]. During this fiber shortening, however, the overall muscle length, defined by the distance between insertion and origin points, can remain constant, increase or decrease. Biceps Brachii can be contracted while the muscle length remains constant,
in which case the elbow angle is constant; contraction can also occur when muscle length is either increasing or decreasing, i.e., when the elbow angle is varied. A contraction during which muscle length remains constant is defined as being isometric; if muscle length varies during contraction, the contraction is termed isotonic. In order to apply a torque isometrically to the elbow joint, the elbow angle must remain constant during load application, insuring that contracting muscles do not change in length. The requirement that torque be applied to the elbow joint demands precise definition of the point of rotation of the elbow joint. That point was defined to be the line of centers thru the lateral and medial epicondyles of humerus (personal communication with Dr. Scharf). Torque was applied about this point of rotation. Finally, torque must be applied to the elbow alone, eliminating possibilities that part of the load be supported by wrist or shoulder. Thus, only those muscles crossing the elbow joint will sustain the total applied torque.

With anatomical requirements for the device established, the following design specifications were postulated:

a. torque applied to the elbow joint only, about the elbow center of rotation, and with elbow angle constant.

b. torque measured be that torque sustained by the elbow without correction for physical differences between subjects.

c. structural integrity of device assured for subject application of 50 ft-lbs or less torque (Lloyd [62], Singh and Karpovich [71]).

d. output (torque measurement) must be a time continuous
analog signal.

The last specification warrants further explanation. Consider someone holding a mass in his outstretched hand. Assume rigidity at the wrist and no movement of the arm, that is, a perfectly static system analogous to a rigid beam supporting a load, then the load applied to the hand is exactly equal to the product of the mass and the acceleration due to gravity. If, however, the arm wavers slightly or the non-rigid wrist allows the hand to tremor, then the load sustained by the hand is a function of the arm and hand accelerations, and no longer simply equal to the weight. Since no biological system is perfectly rigid, a time continuous or analog measurement of the torque sustained by the elbow is necessary.

These specifications, along with desire that the device be simple, inexpensive, and portable, suggested simply-supported beam to serve as both the measuring and loading mechanisms. This dual function was accomplished by specifying strain gages be mounted on top and bottom of one end of the beam to measure deflection induced by a moment applied by a subject to one end of the beam. Alignment of the moment vector applied to the beam, with the centerline of elbow rotation assures that the moment applied to the beam is identical to that torque sustained by the elbow. Thus, the beam moment measured by the strain gages was made identical with the torque applied to the elbow joint. Output voltage potentials from the strain gages were, of course, easily recorded as an analog signal.
Mechanical components of the Load Device were funded by the Georgia Institute of Technology School of Mechanical Engineering and assembled in the School machine shops. These components are:

(1) torque bar,
(2) four pillow blocks,
(3) two pivot shafts,
(4) end-supported beam, and
(5) arm and hand rests.

Refer to Figure 20. The pillow blocks are bolted to a mahogany base platform. The pivot shafts, mounted in pillow blocks, are free to rotate until the end-supported beam is located in slots milled in the pivot shafts. Set screws in the left pivot shaft clamp the end-supported beam to eliminate horizontal translation of the beam.

Attached to the left pivot shaft and secured by a tapered pin, is the torque bar. A hand rest is mounted to the torque bar by a rectangular protrusion which slides in a horizontal slot milled in the torque bar. This adjustment compensates for differing forearm lengths. The subject's forearm was secured to the hand rest by a leather strap two inches wide which wrapped around the forearm just above the protruding styloid process of ulna (refer to Figure 7). This method of attachment precluded wrist support of any load. The forearm in the vicinity of the elbow lay on the arm rest, with the condyles aligned with the center line of the pivot shaft.

Structural considerations for development of the device included the following factors:
Figure 20. Top View of Load Device.
(1) All torsion applied to pivot shaft supported by calibrated beam at a moment vector line.

(2) The torque imposed by the torque bar be sustained by torsion in the pivot shaft.

(3) Vertical and horizontal load components not be applied to the end-supported beam.

(4) Rigid system to keep loading isometric.

These factors were accomplished by:

(1) Pivot shaft designed such that a sharp edge, which bears on the beam surface, is aligned with the centerline of the shaft.

(2 and 3) Shaft turns in bearings.

(4) Parts sized properly.

The end-supported beam is steel bar stock, 1.5 in. x 10 in. x 5/16 in., mounted in the pivot shaft slots separated 8.0 in. between shaft centerlines thus providing effective beam length of 8.0 in. (see Figure 20). The beam was sized such that Pixie strain gages mounted on top and bottom of the beam 7 in. from the line of moment application, can measure maximum torque of 50 ft-lbs without failure. Strain gage instrumentation was funded by the Emory Regional Rehabilitation Research and Training Center and designed and built in the Neurophysiology electronic shop by Mr. Glenn Shine. Pixie 8101 semiconductor strain gages, manufactured by Endevco Laboratories were bonded to the beam with Kodak 910 adhesive, as shown in Figure 21. The gages form two arms of a Wheatstone bridge circuit (refer to Figure 22). The variable 2.5 KΩ potentiometer serves to balance the bridge, at which time zero output is observed and the conventional
Figure 21. Strain Gage Mounting on Beam.
Figure 22. Strain Gage Bridge Circuit.
ratio holds:

\[
\frac{P_1}{10} = \frac{P_2}{10} \quad \text{(see Figure 22).}
\]

This capability enables the operator to establish a zero baseline on an oscilloscope or other monitoring equipment. The operational amplifier, with 250 Kn potentiometer, provides maximum gain of 25 which was used for all experiments. The 2.2 Kn resistor is the amplifier trim resistor and the 4.7 Kn resistor is the temperature compensator for the amplifier. By locating strain gages on opposite sides of the beam, two things are accomplished:

1. Since the two gages will be subject to strains equal in magnitude but opposite in sign, output from the bridge is doubled, increasing sensitivity.

2. Temperature compensation is secured, since both gages are affected by equal change in resistance due to temperature, the ratio \( \frac{P_1}{P_2} \) will be unaffected.

The Pixie strain gages are proof tested by the manufacturer to 40 grams force applied to the free end of a cantilevered gage in direction to load the semiconductor in tension. Resistance change remains linear through ±10 grams, but deviation from linearity is only slight with loads from 10 - 30 grams. This proof force of 40 grams was the criterion that established beam dimensions to sustain 50 ft-lbs torque. From a compliance value 55 \( \mu \text{in/gm} \), given by the manufacturer, a maximum deflection was calculated as
\[(55 \text{ \text{	ext{	ext{	ext{in/gm}}}}}) \times (40 \text{ \text{gms}}) = 2.2 \times 10^{-3} \text{ in.}\]

Using the deflection equation for a simply supported beam

\[y = \frac{1}{6} \frac{M_o}{EI} (3x^2 - \frac{x^3}{L} - 2lx)\]

required stiffness was calculated such that, for maximum moment of 50 ft-lbs, maximum deflection at the strain gage would be less than or equal to \(2.2 \times 10^{-3}\) in. The Load Device calibration curve (Figure 23) is very nearly linear through 40 ft-lbs. This curve was obtained with a simple rope and pulley arrangement in which weights were hung applying a force exactly 1.0 foot from the pivot shaft centerline.

The output from the strain gages was displayed on an oscilloscope which could be viewed by the subject. In the experiment, the subject was asked to flex his elbow and deflect the scope beam some requested distance. Effectively, then, the subject controlled the measured torque applied to his elbow joint.

4.2 EMG Equipment

As the EMG potentials are on the order of one millivolt or less, high impedance Argonaut LRA O42 AC Differential Preamplifiers were used to amplify the signals before recording. One major difficulty in recording these low voltage potentials is high noise.
Figure 23. Load Device Calibration Curve.
level from lighting and other currents. These interference signals induce alternating voltages between the subject's body and ground. This interference can often be much higher in magnitude than the EMG signal and thus mask or override the EMG signal. To minimize this interference, the subject is grounded. Unfortunately the high resistance of skin prevents perfect grounding. For these reasons, a vacuum tube differential amplifier with ability to discriminate between voltages common to both inputs was used to minimize this noise problem. Amplifier was set with gain of 1000, low frequency response of 1.0 cps, and high frequency response of 1.6 Kcps.

EMG activity in the eight muscles was monitored on a Hewlett-Packard 1309A Monitor Scope with eight channel capacity. This scope is shown in Figure 25 on top of the amplifier bank. The oscilloscope to the right of the amplifier bank was used to check interference levels on the EMG channels.

Just to the left of the amplifiers is the 14-channel Hewlett-Packard 3955A Magnetic Recording System. Ten channels of data were recorded simultaneously at tape transport speed of 15 in./sec, on one-inch wide FM recording tape. EMG and pulse input channels were calibrated to ±1 volt, while the Load Device channel was ±\( \frac{1}{4} \) volts. For data reduction purposes, a one-volt pulse signal was recorded as a trigger signal necessary for computerized data reduction scheme. The data reduction techniques are described in detail in Chapter VI.
4.3 Electrodes

Two types of electrodes were used in these experiments, surface or skin electrodes and fine-wire intramuscular electrodes. A pair of surface electrodes was used to pick up EMG activity in the Triceps muscle. As monitoring of Triceps was to verify its expected inactivity during flexion, the precision afforded by intramuscular electrodes was not required. The other seven muscles were monitored with intramuscular electrodes.

Surface electrodes are convenient for indicating a gross degree of activity or inactivity from superficial muscles. However, the pick-up area is quite wide-spread; activity from several muscles has been recorded from one pair of surface electrodes located over one specific muscle [52].

The surface electrodes used were four mm diameter Beckman silver-silver chloride skin electrodes. They were applied to the skin with a saline paste "electrode jelly" (EKG-SOL) to improve electrical contact by removing body oils having high electrical resistance. Previous to application, light abrasion of the skin removed the dead surface layer of skin. These techniques lowered electrical resistance to about 3-4KΩ. An adhesive collar around the electrode discs secured electrodes to the skin.

The intramuscular fine-wire bipolar electrodes used in the seven other muscles, allow monitoring of deep underlying muscles without interference from activity of other muscles. The electrodes are made from polyurethane-insulated, Karma alloy wire, 25 microns
in diameter, the same order of magnitude as a muscle fiber diameter. These electrodes were developed independently by several research centers in 1960-1962. The method of electrode construction used for these thesis experiments was developed by Basmajian and Stecko, 1962 [13]. The steps for making an electrode are described below and depicted in Figure 24 [13]:

1. A double strand of the polyurethane coated wire is passed thru the shaft of a 27 gage hypodermic needle, leaving a small loop.

2. Insulation is burned from the looped ends and the ends of wire projecting from the base of the needle.

3. The loop is cut, leaving 1 - 1.5 mm of bared wire on each of the two wires. This distance is verified under a light microscope. These bared ends are staggered so that they will not contact and short within the muscle. They are bent back to form a barb which holds the wires in the muscle during needle withdrawal after insertion.

4. At the other end, the wires are bared for a length 3-4 mm of their total length of 5-7 cm.

5. The needle-wire assembly is dry sterilized in a simple paper folder for one hour at 130°C. The electrodes are inserted with the needle and the needle withdrawn to leave the wires in the muscle (see Figure 26). A simple spring-wire coil connector (Basmajian, Forrest, and Shine, 1966 [13]), to which amplifier input wires have been soldered connects the fine wires
Karma wire looped through hypodermic needle

Nylon insulation removed from looped end and free ends

Loop cut and bare ends of wire staggered

Wire ends bent back over needle to form barbs

Figure 24. Steps in Making Bipolar Fine-Wire Electrodes [13].
Figure 25. EMG Equipment.

Figure 26. Insertion of Intramuscular Electrodes.
to the amplifiers. These spring connectors are brass springs 4 mm in diameter and 12 mm in length. The fine wire is clamped between the spring coils.

For these experiments an eight-channel multiwire cable was devised by Mr. Shine to connect the amplifier bank with the electrodes. Once the fine wires were clamped in the springs, the springs were taped to a tape surface on the skin, without directly contacting the skin itself.
CHAPTER V

EXPERIMENTAL METHODS

5.1 Subjects

A total of ten healthy subjects, between the ages of 23 and 29, volunteered for the experiment. Of the nine males, eight were engineering graduate students. The other male was the orthopedic resident who did the needle (electrode) insertions. One subject (SST) had a five inch incision scar longitudinally along the anterior side of the forearm in the vicinity of the insertions for Pronator Teres and Flexor Carpi Radialis. In 1968, this subject had broken both radius and ulna bones, and a metal plate and rod, which have since been removed, were inserted.

The subject was seated in a modified dental chair next to a small table to which the loading device was clamped, as shown in Figures 27 and 28. The chair height was adjusted to allow the subject's right arm to rest comfortably on the padded arm rest of the loading device. Further adjustments of the subject's position were postponed until all electrodes were inserted.

5.2 Electrode Placement

The subject was first grounded to the shield of the coaxial cable by means of a silver disc, about 3/4 inch in diameter, which was applied to the skin at the middle of the forearm with an electrode jelly electrolytic paste (EKG-SOL) and held in place with
Figure 27. Dental Chair and Equipment.

Figure 28. Subject and Loading Device.
a velcro strap. Seven intramuscular electrodes were inserted into the following muscles by Dr. Scharf according to surface anatomy: Biceps Brachii, short and long heads, Brachialis, Brachioradialis, Extensor Carpi Radialis, Pronator Teres, and Flexor Carpi Radialis. Surface electrodes were used to monitor EMG activity in Triceps medial head. Electrode positions were checked and confirmed by observing the EMG activity on the oscilloscope while appropriate muscle functions were performed. The belly of the muscle was the goal for every insertion, and the sharpness of EMG spikes was one basic criterion for determining whether electrodes were indeed in the belly. One-inch needles were used for all muscles except Brachialis which required a 1 1/4 in. needle. For each muscle, the procedure for insertion and checking electrode position is described in detail below.

1. Biceps Brachii, Short and Long Heads: The two heads of Biceps Brachii originate separately. About half way down the arm the two bellies fuse and the muscle is inserted by a common tendon on radius bone of the forearm. Before fusing, the two heads are separated by a septa running down the middle of the muscle proximally separating the two heads. On the basis of this information, the Biceps muscle mass was grasped. Insertions were done medially and laterally for the short and long heads respectively. To check electrode positions, EMG activity was expected during elbow flexion against resistance and during supination of the forearm against resistance. This test was used for both heads, and no reliable
method was conceived to distinguish between the two.

(2) Brachialis: Knowing that Brachialis is posterior to Biceps, rather than attempt to insert through Biceps to Brachialis, Dr. Scharf inserted the needle laterally on the upper arm at the junction of the middle and distal thirds (see footnote p.37). The needle was inserted to humerus bone and then moved anterior into the belly of Brachialis. To assure proper position of the electrodes, EMG activity was expected during elbow flexion, both with and without resistance; no activity was expected during supination of the forearm.

(3) Brachioradialis and Extensor Carpi Radialis Longus: Henry's Mobile Wad of Three was grasped. Insertion for Brachioradialis was made in the most medial and ventral portion, and insertion of Extensor Carpi Radialis Longus was made in the most lateral and dorsal portion. Wrist extension, both with and without resistance, was used to confirm electrode position in Extensor Carpi Radialis. Confirmation for electrode position in Brachioradialis involved flexion of the elbow against resistance with the forearm in a semi-prone position.

(4) Pronator Teres and Flexor Carpi Radialis: With both muscles originating from the medial epicondyle on humerus, Pronator Teres was recognized to be ventral and lateral to Flexor Carpi Radialis. Consequently, insertion for Flexor Carpi Radialis was made at a point approximately two inches distal and one inch anterior to the medial epicondyle, and insertion for Pronator Teres was made at a point slightly lateral to the flexor insertion point. The EMG
activity test to distinguish these two muscles was pronation, both with and without resistance, which should result in activity in Pronator Teres, but no activity in Flexor Carpi Radialis. Due to the close proximity of these two muscles, proper electrode positioning required reinsertions in several experiments.

(5) Triceps: In order to verify expected inactivity in antagonist muscles during elbow flexion, the medial head of Triceps was monitored with surface electrodes placed approximately 1.5 inches apart midway down on the upper arm.

5.3 Experimental Procedure

After the intramuscular electrodes had been inserted and surface electrodes applied, the EMG activity of each muscle was again monitored during appropriate arm movements to confirm proper electrode positioning. The chair back and position of the loading device were adjusted to establish an elbow angle of either 90° or 150°, approximately. The lateral epicondyle of the humerus, considered to be the center of rotation of the elbow joint, was aligned with the moment vector of the loading device. With forearm either prone or supine, the subject's wrist was then secured to the loading device with a two inch wide leather strap, as shown in Figure 28.

As mentioned previously, the output from the strain gages on the loading device was displayed on a small oscilloscope, the screen of which was a grid, vertically graduated in eight divisions. The subject was asked to view the scope and attempt to vertically deflect the scope beam by flexing his elbow against the loading device.
The subject was allowed to familiarize himself with the equipment for a few minutes, and then was given the following instructions:

You will be asked to deflect the beam on the scope 0.5, 1.0, 1.5, 2.0, 2.5, 3.0, 3.5, 4.0, 5.0, 6.0, 7.0, and 8.0 divisions; at each load you should hold that load until instructed to proceed to the next load. You are asked to relax only after the entire 12-load sequence is completed. With the same arm position, you will be asked to repeat the experiment with two other loading sequences: (1) loading up to the maximum load and progressively unloading, so that the sequence is 8.0, 7.0, 6.0, etc. divisions, and (2) random loading in which the 12 loads are randomly selected. At the beginning of each loading sequence, you are to completely relax. This relaxation period is important as it allows monitoring and recording of the interference level in each EMG channel.

With the subject ready to begin the experiment, interference levels in the EMG channels were checked and minimized by passing a current through the electrodes at the electrode spring connectors with an ohmmeter. Also, the overhead fluorescent lights were turned off and the door shut. The eight-channel EMG cable to the amplifiers was quite motion sensitive, but this potential problem was minimized by supporting the cable.

The Hewlett-Packard recorder was turned on and the subject asked to relax. When a relaxed condition had been achieved, as evidenced by no EMG activity on the monitor scope, the experimenter recorded the one-volt pulse. Approximately two-three seconds later, the subject was instructed to achieve the first load. The one-volt pulse was recorded approximately two seconds after the load had been achieved. After the pulse, the subject maintained the load for approximately two seconds, and then he was instructed to achieve the next load. The 12-load sequence required about a minute.

The entire experiment involved:
(1) three different loading methods: sequential loading, sequential unloading, and random loading, and

(2) four different arm positions: forearm prone, elbow angle 90°; forearm supine, elbow angle 90°; forearm prone, elbow angle 150°; forearm supine, elbow angle 150°.

The experiment required 1 1/2 to 2 hours, including electrode insertions.

5.4 Reproducibility Experiments

Considering the range of parameters, both controllable and uncontrollable, involved in an experiment of this nature, it was deemed not unnecessarily prudent to check experimental reproducibility. For this purpose, one subject (SST), who took part in the first set of experiments, plus one other subject (DIL), repeated a portion of the experiment. The repeat experiment was done with sequential loading method only and with only two arm positions (forearm prone, elbow angle 90° and forearm supine, elbow angle 90°). For each arm position, the experiment was repeated four times.

Electrode insertions were done by Mr. Steve Wolf, doctoral student in anatomy. Since with a pair of surface electrodes, it is quite possible to pick up EMG activity from several muscles, it seemed conceivable that the activity observed at higher loads from Triceps in the first set of experiments was actually activity from Biceps or Brachialis. For this reason, intramuscular instead of surface electrodes were inserted into medial head of Triceps. Needle length was constant for each subject: 1 1/4 inch for.
Brachialis for both subjects, one inch for SST and $\frac{3}{4}$ inch for DTL for remaining muscles. This experiment was repeated on two different occasions, with two days of rest between experimental days. At the second experiment both subjects complained of soreness and slight cramping, resulting from previous experiment.
CHAPTER VI

DATA REDUCTION

The data acquired from these experiments consists of ten channels of analog information (Output Torque Channel, Sample Pulse Channel, and eight EMG Channels) simultaneously recorded on FM tape. The data reduction scheme involved:

(1) digital integration with respect to time of both the Torque and EMG signals, and

(2) plotting the results as Integrated EMG versus Integrated Torque.

Data reduction was accomplished with a DEC PDP 8/I Laboratory Computer, a Hewlett-Packard X-Y Plotter, Burroughs 5500 Digital Computer, and a Cal-Comp Plotter. A diagram summarizing the Data Acquisition and Reduction Process is shown in Figure 29.

Analog-to-digital conversion was accomplished with the DEC PDP 8/I computer, which had maximum of four analog input channels. Thus, three EMG Channels and the Torque Channel were input simultaneously to the computer from the FM tape. The Sample Pulse Channel was input to the computer's Pulse input channel. The four analog signals were digitally sampled at one KHz. This sampling rate was selected upon completion of a Sampling Rate Determination presented in Appendix B. The sampling routine was triggered by the one-volt Sample Pulse after which one second of data was digitally sampled.
Figure 29. Diagram of Data Acquisition and Reduction Process.
Thus, for a one kHz sampling rate, 1000 digital samples were obtained for each one-second record for each analog channel. By summing successive samples in lots of 100, ten data points were obtained for each one-second record. A numerical average of these ten points was calculated to represent the integrated value of the one-second record.

For each of the 12 load points in the loading sequence, as well as the initial relaxed point, one second of data was sampled in the manner described above, and an integrated value was obtained for the Torque Channel and each of the eight EMG Channels. These integrated values were output from the computer in arbitrary units on the Teletype. For each EMG channel, the X-Y plotter was used to plot the Torque values on the abscissa and the EMG values on the ordinate. These EMG-Torque graphs in arbitrary units served as a first look at the data. However, since the graphs were in arbitrary units and thus not comparable among experiments, the data was further reduced by means of a calibration technique.

The equipment was calibrated for each experiment to account for day-to-day equipment variations. The Load Device was calibrated by applying a known torque (4.4 ft-lbs) to the beam and recording the output signal from the strain gages on the Torque Channel on FM tape. Each of the EMG Argonaut amplifiers was calibrated with a 200 Hz input signal of one millivolt. The output was also recorded on FM tape. These calibration signals were processed in the same manner as the data. The integrated values for the calibrated torque
signals were output from the computer on the Teletype in arbitrary units per 4.4 ft-lbs. By dividing the data torque values by the appropriate calibrated torque values, that is, arbitrary units divided by arbitrary units per 4.4 ft-lbs, torque values in units of foot-pounds were obtained. Similarly, the integrated values for the calibrated EMG signals were output from the computer in arbitrary units per millivolt. By dividing the data EMG values by the calibrated EMG values, EMG values in units of millivolts were acquired. Using the Burroughs 5500 digital computer and the Cal-Comp plotter, the data was graphed in the final form shown in Appendix A. That is, the EMG values in millivolts were plotted on the ordinate versus the Torque values in foot-pounds on the abscissa. In this form the data is easily and accurately compared among the different experiments.
7.1 Graphs: Integrated EMG versus Integrated Torque

The data are presented in graphical form with Average* Integrated EMG in millivolts plotted on the ordinate and Average* Integrated Torque in foot-pounds plotted along the abscissa. Each muscle is represented by 13 data points, including the initial resting data point and 12 load data points. The exceptions, which have only 12 data points, are so indicated on the graphs. The first data point should indicate the base line noise level on the EMG channel. However, in some instances, the subject was not completely relaxed for the first data point, indicated by the decrease in EMG activity for the second and/or third data points. Thus, the base line noise level is estimated as that point having least EMG activity. To correctly compare EMG magnitudes from different graphs, the base line noise level must be subtracted from the EMG

* As explained in Chapter VI, the analog signal was digitally integrated thusly: \[ \sum_{i=1}^{N} (\text{EMG}_i \Delta t) \] where \( N \) = number of \( \Delta t \) increments.

This results in a data point with units (millivolts-second). To obtain units (millivolts), an average value is computed:

\[ \frac{1}{N} \sum_{i=1}^{N} (\text{EMG}_i \Delta t) \] The torque signal was likewise integrated.
values. With a few exceptions, the noise level remained constant for each channel throughout each experiment. The data and discussion are presented in four parts.

7.1.1 Data Set I

This consists of 40 graph pages, each presenting EMG-Torque data for each of the eight muscles monitored. For each of Experiments 1-8, data for four arm positions were taken:

<table>
<thead>
<tr>
<th>Arm Positions</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Forearm Position</strong></td>
</tr>
<tr>
<td>1. Prone</td>
</tr>
<tr>
<td>2. Supine</td>
</tr>
<tr>
<td>3. Prone</td>
</tr>
<tr>
<td>4. Supine</td>
</tr>
</tbody>
</table>

Data for only the first two arm positions listed were taken for Experiments 9-12. The graphs and a brief discussion of each experiment's graphs are found in Appendix A on pages 152-212.

Insight into the information content of the EMG signal is gained by inter-subject examination of the data and careful consideration of the similarities and differences observed from such examination. The manner in which EMG activity in the eight-muscle system responds to an isometric torque definitely varies from person to person. Variation among subjects, however, is certainly expected, as indicated by Basmajian and Latif [1] (1957) who stated
that they observed no set pattern of the order in which the muscles’ (Biceps, Brachialis and Brachioradialis) activity began or ended. They also observed a wide range of response from each muscle. However, in spite of the inter-subject differences, there are similarities, which indicate trends that agree quite well qualitatively with the anatomical model previously described. Inter-subject comparison is made in several ways and with respect to a variety of parameters.

One of the simplest parameters to observe and compare among subjects is that muscle which appears to be the most prominent elbow flexor. This muscle is evaluated for large torques as that muscle which generates maximum EMG at the maximum torque. For small torques, the prominent elbow flexor (PEF) muscle is determined as that muscle which first* exhibits EMG activity at the lower torques. In two-thirds of the experiments, the PEF muscle is the same for both large and small torques, as shown in Table 2. At large torques, for nine of the 12 experiments, the PEF muscle is either Biceps (short or long head) or Brachialis; in the three exceptions, Brachioradialis is the PEF muscle for one of the arm positions with either Brachialis or Biceps the PEF muscle for the other arm position. For small torques, the PEF muscle is Brachialis in eight cases, and Biceps (short or long) in four cases. In only four experiments, all at large torques, is the PEF muscle variant with arm position. As

---

* As indicated by the slope of the line drawn from the origin through the first two data points.
Table 2. Prominent Elbow Flexor Muscles (PEF).

<table>
<thead>
<tr>
<th>Experiment Number</th>
<th>Large Torque - PEF</th>
<th>Small Torque - PEF</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>BS</td>
<td>BS</td>
</tr>
<tr>
<td>2</td>
<td>BL</td>
<td>BL</td>
</tr>
<tr>
<td>3</td>
<td>BL</td>
<td>BL</td>
</tr>
<tr>
<td>4</td>
<td>EA BS(90°) BR(150°)</td>
<td>BL</td>
</tr>
<tr>
<td>5</td>
<td>B</td>
<td>B</td>
</tr>
<tr>
<td>6</td>
<td>FP B(P) BL(S)</td>
<td>B</td>
</tr>
<tr>
<td>7</td>
<td>B</td>
<td>B</td>
</tr>
<tr>
<td>8</td>
<td>BS</td>
<td>B</td>
</tr>
<tr>
<td>9</td>
<td>BS</td>
<td>B</td>
</tr>
<tr>
<td>10</td>
<td>B</td>
<td>B</td>
</tr>
<tr>
<td>11</td>
<td>FP BS(P) BR(S)</td>
<td>B</td>
</tr>
<tr>
<td>12</td>
<td>FP BR(P) B(S)</td>
<td>B</td>
</tr>
</tbody>
</table>

Except where indicated (Experiments 4, 6, 11, 12), the PEF muscle listed is the same for all arm positions. For Experiment 4 the PEF varies with elbow angle (EA) and in Experiments 6, 11, and 12 the PEF varies with forearm position (FP) where S = supine and P = prone.
stated previously from Primary Anatomy [16], Brachialis is the chief elbow flexor muscle, while Biceps and Brachioradialis act to help flex the elbow against strong resistance. The sole function of Brachialis is elbow flexion whereas Biceps and Brachioradialis act in other capacities in addition to elbow flexion. Thus the observation of Brachialis as the FEF muscle at small torques in two-thirds of the experiments and Brachialis as well as Biceps and Brachioradialis as the PEF muscles at large torques, appear most agreeable with elbow joint anatomy.

The ordering of the muscles according to the values of IEMG at maximum torque provides useful inter-subject information which also agrees with elbow joint anatomy. An examination of muscle ordering reveals the manner in which the muscles interact with each other to sustain the torque as a muscle system. Muscle ordering for arm positions of Prone 90° and Supine 90° are presented in Tables 3 and 4, respectively. Since the significance of Extensor Carpi Radialis (WE) and Flexor Carpi Radialis (WF) muscles is questionable, they are excluded from these two tables but are discussed in detail on page 120. In both tables, inter-subject variation is emphasized by the fact that none of the 12 experiments have the exact same ordering of muscles. However, again for both tables, the first four muscles (listed as No. 1-4 in the tables) are Brachialis, Biceps Short, Biceps Long, and Brachioradialis, in different combinations, for seven of the 12 experiments. In the other five cases, Pronator Teres replaces one of these four muscles in the ordering (PT designated as No. 4 or less). Correlation of these observations with the geometric anatomical model
Table 3. IEMG at Maximum Torque (Prone 90°).

<table>
<thead>
<tr>
<th>Experiment Number</th>
<th>Maximum Torque</th>
<th>BS</th>
<th>BL</th>
<th>B</th>
<th>BR</th>
<th>PT</th>
<th>T</th>
</tr>
</thead>
<tbody>
<tr>
<td>8</td>
<td>35</td>
<td>1</td>
<td>2</td>
<td>3</td>
<td>4</td>
<td>5</td>
<td>6</td>
</tr>
<tr>
<td>9</td>
<td>42</td>
<td>1</td>
<td>3</td>
<td>2</td>
<td>4</td>
<td>5</td>
<td>6</td>
</tr>
<tr>
<td>4</td>
<td>37</td>
<td>1</td>
<td>3</td>
<td>4</td>
<td>2</td>
<td>6</td>
<td>5</td>
</tr>
<tr>
<td>11</td>
<td>37</td>
<td>1</td>
<td>4</td>
<td>3</td>
<td>2</td>
<td>5</td>
<td>6</td>
</tr>
<tr>
<td>1</td>
<td>34</td>
<td>1</td>
<td>6</td>
<td>3</td>
<td>2</td>
<td>4</td>
<td>5</td>
</tr>
<tr>
<td>3</td>
<td>40</td>
<td>2</td>
<td>1</td>
<td>4</td>
<td>3</td>
<td>5</td>
<td>6</td>
</tr>
<tr>
<td>2</td>
<td>38</td>
<td>4</td>
<td>1</td>
<td>2</td>
<td>5</td>
<td>3</td>
<td>6</td>
</tr>
<tr>
<td>6</td>
<td>40</td>
<td>2</td>
<td>4</td>
<td>1</td>
<td>3</td>
<td>6</td>
<td>5</td>
</tr>
<tr>
<td>10</td>
<td>42</td>
<td>2</td>
<td>3</td>
<td>1</td>
<td>4</td>
<td>5</td>
<td>6</td>
</tr>
<tr>
<td>7</td>
<td>42</td>
<td>5</td>
<td>4</td>
<td>1</td>
<td>2</td>
<td>3</td>
<td>6</td>
</tr>
<tr>
<td>5</td>
<td>40</td>
<td>4</td>
<td>5</td>
<td>1</td>
<td>3</td>
<td>2</td>
<td>6</td>
</tr>
<tr>
<td>12</td>
<td>37</td>
<td>4</td>
<td>5</td>
<td>2</td>
<td>1</td>
<td>3</td>
<td>6</td>
</tr>
</tbody>
</table>

BS = Biceps Short Head  
BL = Biceps Long Head  
B  = Brachialis  
BR = Brachioradialis  
PT = Pronator Teres  
T  = Triceps
Table 4. IEMG at Maximum Torque (Supine 90°).

<table>
<thead>
<tr>
<th>Experiment Number</th>
<th>Maximum Torque ft-lbs</th>
<th>BS</th>
<th>BL</th>
<th>B</th>
<th>BR</th>
<th>PT</th>
<th>T</th>
</tr>
</thead>
<tbody>
<tr>
<td>8</td>
<td>35</td>
<td>1</td>
<td>2</td>
<td>3</td>
<td>4</td>
<td>5</td>
<td>6</td>
</tr>
<tr>
<td>9</td>
<td>42</td>
<td>1</td>
<td>2</td>
<td>3</td>
<td>4</td>
<td>5</td>
<td>6</td>
</tr>
<tr>
<td>4</td>
<td>37</td>
<td>1</td>
<td>3</td>
<td>4</td>
<td>2</td>
<td>6</td>
<td>5</td>
</tr>
<tr>
<td>1</td>
<td>34</td>
<td>1</td>
<td>5</td>
<td>3</td>
<td>2</td>
<td>4</td>
<td>6</td>
</tr>
<tr>
<td>3</td>
<td>40</td>
<td>2</td>
<td>1</td>
<td>3</td>
<td>4</td>
<td>5</td>
<td>6</td>
</tr>
<tr>
<td>2</td>
<td>38</td>
<td>2</td>
<td>1</td>
<td>3</td>
<td>5</td>
<td>4</td>
<td>6</td>
</tr>
<tr>
<td>6</td>
<td>40</td>
<td>4</td>
<td>1</td>
<td>2</td>
<td>3</td>
<td>6</td>
<td>5</td>
</tr>
<tr>
<td>10</td>
<td>42</td>
<td>2</td>
<td>3</td>
<td>1</td>
<td>4</td>
<td>5</td>
<td>6</td>
</tr>
<tr>
<td>7</td>
<td>42</td>
<td>6</td>
<td>4</td>
<td>1</td>
<td>3</td>
<td>2</td>
<td>5</td>
</tr>
<tr>
<td>5</td>
<td>40</td>
<td>3</td>
<td>4</td>
<td>1</td>
<td>5</td>
<td>2</td>
<td>6</td>
</tr>
<tr>
<td>12</td>
<td>37</td>
<td>4</td>
<td>5</td>
<td>1</td>
<td>2</td>
<td>3</td>
<td>6</td>
</tr>
<tr>
<td>11</td>
<td>37</td>
<td>3</td>
<td>4</td>
<td>2</td>
<td>1</td>
<td>5</td>
<td>6</td>
</tr>
</tbody>
</table>

BS = Biceps Short Head  
BL = Biceps Long Head  
B  = Brachialis  
BR = Brachioradialis  
PT = Pronator Teres  
T  = Triceps
for 90° elbow angle is evidenced by the fact that the muscles in order of decreasing moment arm length are Biceps Short, Biceps Long, Brachioradialis, Brachialis, and Pronator Teres. In view of the above remarks from *Primary Anatomy* and the additional textbook statement that Pronator Teres is chiefly a forearm pronator but aids elbow flexion when great force is required, the data seem to correlate also with elbow joint anatomy.

A third parameter significant in inter-subject comparisons is the effect of the elbow angle (EA) and forearm position (FP) on IEMG-Torque relationships. In 1969, Bankov and Jorgensen [39] using surface electrodes, concluded the following points from their data:

1. Isometric maximum torque of the elbow flexors (Biceps and Brachioradialis) depends on the EA and FP.
2. IEMG from Biceps for prone forearm was 50 per cent of that obtained with supine forearm.
3. No significant differences in IEMG for Biceps for different EA or for Brachioradialis for different EA and/or FP.

In this thesis study, the effect of EA and FP has been evaluated in two ways:

1. Variation of IEMG at maximum torque with EA and FP, and
2. Effect of EA and FP on the ordering of the muscles according to the value of IEMG at maximum torque.

Results of the first method are presented in Table 5, in which the IEMG at maximum torque is evaluated for the two forearm positions and the two elbow angles. For most of the muscles, one of the two forearm positions results in substantially more EMG activity than the
Table 5. Effect of Forearm Position and Elbow Angle on IEMG at Maximum Torque for Each Muscle

<table>
<thead>
<tr>
<th>Muscle</th>
<th>%P&gt;S</th>
<th>%S&gt;P</th>
<th>% Equal</th>
<th>%90&gt;150</th>
<th>%150&gt;90</th>
<th>% Equal</th>
</tr>
</thead>
<tbody>
<tr>
<td>BL</td>
<td>20</td>
<td>60</td>
<td>20</td>
<td>12.5</td>
<td>50</td>
<td>37.5</td>
</tr>
<tr>
<td>BS</td>
<td>25</td>
<td>35</td>
<td>40</td>
<td>31.2</td>
<td>96.2</td>
<td>12.6</td>
</tr>
<tr>
<td>B</td>
<td>20.8</td>
<td>54.2</td>
<td>25</td>
<td>31.2</td>
<td>43.8</td>
<td>25</td>
</tr>
<tr>
<td>BR</td>
<td>12.5</td>
<td>37.5</td>
<td>50</td>
<td>31.25</td>
<td>31.25</td>
<td>37.5</td>
</tr>
<tr>
<td>WE</td>
<td>20.8</td>
<td>50</td>
<td>29.2</td>
<td>37.5</td>
<td>37.5</td>
<td>25</td>
</tr>
<tr>
<td>FT</td>
<td>37.5</td>
<td>25</td>
<td>37.5</td>
<td>43.8</td>
<td>25</td>
<td>31.2</td>
</tr>
<tr>
<td>WF</td>
<td>37.5</td>
<td>41.7</td>
<td>20.8</td>
<td>31.2</td>
<td>43.8</td>
<td>25</td>
</tr>
<tr>
<td>BS + BL</td>
<td>25</td>
<td>55</td>
<td>20</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

Explanation of Column Titles:

1. %P>S: The percentage of the 12 experiments in which the IEMG at maximum torque was greater for prone forearm than supine.
2. %S>P: The percentage of the 12 experiments in which the IEMG at maximum torque was greater for supine forearm than prone.
3. % Equal: The percentage of 12 experiments in which the IEMG for prone forearm was equal (± 0.03 mv) to that of supine forearm.
4. %90>150: The percentage of 8 experiments in which IEMG at maximum torque was greater for elbow angle of 90° than 150°.
5. %150>90: The percentage of 8 experiments in which IEMG at maximum torque was greater for elbow angle of 150° than 90°.
6. % Equal: Percentage of 8 experiments in which IEMG equal for both FA.
other. The same is true for elbow angles. For BL, BS, B, and WF the IEMG is definitely greater for elbow angle of 150° than 90°. Reference to the geometric anatomical model shows that for all muscles except Triceps the moment arm length decreases as the elbow angle increases from 90° to 150°. For the same torque to be sustained, the force sustained by each muscle must increase as the elbow angle increases. The increase in IEMG for increased elbow angle for BL, BS, B and WF muscles thus agrees with the anatomical model. Only for PT are results such that the IEMG is greater for 90° than 150° contradicting model predictions, assuming that the model basically predicts direct proportionality between integrated EMG and integrated Torque. Also, the percentages for BR and WE indicate that both elbow angles result in approximately equal EMG activity in these two muscles.

The other method by which the effect of EA and FP is evaluated is the effect of EA and FP on the ordering of the muscles according to IEMG values at maximum torque. Table 6 shows, for each experiment and each FP and EA, the muscles in decreasing order to the right according to IEMG at maximum torque. Again, as in Tables 3 and 4, WF and WE are excluded from Table 6. In six cases (No. 5, 6, 7, 9, 11, and 12) the muscle order seems established by the forearm position, while in two cases (No. 1, 4) the order is a function of elbow angle. Consider Experiment 6 as an example of forearm position ordering. For both elbow angles, for a prone forearm the muscles in decreasing order of IEMG at maximum torque are: B, BS, BR, BL, T, PT; for a supine forearm the muscles are, in order; BL, B, BR, BS, T, PT, again for both elbow angles. In three cases (No. 3, 8, 10) the muscle
Table 6. Effect of Forearm Position and Elbow Angle on Ordering of Muscles According to IEMG at Maximum Torque.

<table>
<thead>
<tr>
<th>Experiment 1</th>
<th>Experiment 7</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>P</strong></td>
<td>BS</td>
</tr>
<tr>
<td><strong>S</strong></td>
<td>BR</td>
</tr>
<tr>
<td><strong>P</strong></td>
<td>BS</td>
</tr>
<tr>
<td><strong>S</strong></td>
<td>BS</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Experiment 2</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>P</strong></td>
</tr>
<tr>
<td><strong>S</strong></td>
</tr>
<tr>
<td><strong>P</strong></td>
</tr>
<tr>
<td><strong>S</strong></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Experiment 3</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>P</strong></td>
</tr>
<tr>
<td><strong>S</strong></td>
</tr>
<tr>
<td><strong>P</strong></td>
</tr>
<tr>
<td><strong>S</strong></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Experiment 4</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>P</strong></td>
</tr>
<tr>
<td><strong>S</strong></td>
</tr>
<tr>
<td><strong>P</strong></td>
</tr>
<tr>
<td><strong>S</strong></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Experiment 5</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>P</strong></td>
</tr>
<tr>
<td><strong>S</strong></td>
</tr>
<tr>
<td><strong>P</strong></td>
</tr>
<tr>
<td><strong>S</strong></td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Experiment 6</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>P</strong></td>
</tr>
<tr>
<td><strong>S</strong></td>
</tr>
<tr>
<td><strong>P</strong></td>
</tr>
<tr>
<td><strong>S</strong></td>
</tr>
</tbody>
</table>

Note: First column indicates elbow angle (90° or 150°) and second column designates forearm position (P = prone, S = supine).
order is virtually unchanged by varying FP and/or EA, and in one case (No. 2) the order is definitely affected, but in an indeterminant manner.

Another topic pertaining to the significance of FP and EA on EMG activity, refers specifically to the effect of FP on EMG activity in Biceps Brachii. Basmajian and Latif [1] reported activity in Biceps is definitely greater with the forearm supine. Bankov and Jorgensen confirmed this statement and further stated that IEMG in Biceps while the forearm was supine was twice that for a prone forearm. In only one case of 12 of this thesis data, is the IEMG from Biceps (sum of short and long heads) for supine forearm greater than that for prone forearm by a factor of two. In none of the other cases is this factor even 1.5. As shown in Table 5, the percentages for BS are somewhat inconclusive. The ratio \( \%S > \%P / \%S > \%P > 3 \) is 4.0 for BL and 2.2 for (BL + BS) indicating that supine forearm does result in more EMG activity in Biceps Brachii than prone forearm. Also, it is recognized that these percentages are based on twelve experiments, which hardly constitutes a statistical study. However, the different results for the two heads of Biceps suggest differing anatomical functions for Biceps Long and Biceps Short.

Recall from elbow joint anatomy, that the two heads of Biceps originate from two different places on the scapula, but insert as a common tendon on the radius bone. Basmajian and Latif found that BL was more active than BS during slow flexion and supination of the forearm against resistance. They observed little difference in the
activity of the two heads during isometric contraction. A good explanation for the data showing supine IEMG greater than prone 60 per cent of the cases for BL and 35 per cent for BS, involves sagittal plane elbow flexion. Okamoto found the direction of load application a crucial parameter in his studies (see pp. 75-76). Shown in Figure 30 are two front views of the upper limb including elbow and shoulder joints. The two views differ in the orientation of the humerus bone, as View A shows humerus parallel to the body, i.e., in a sagittal plane defined on p. 22 while View B shows humerus abducted from the body 30 degrees, and hence not in a sagittal plane. The radius bone is shown perpendicular to the plane of the paper. The lines connecting the origin points and insertion point for BL and BS are also shown. Since the origin points of the two heads are horizontally separated by the distance Ax, abducting the humerus away from the body decreases the angle, \( \theta \), between the lines of action of the two heads. Since, the humerus during these experiments was not rigidly restricted to the sagittal plane, slight abduction was possible. If this phenomenon did occur, it would explain the differing data regarding forearm position for the two heads of Biceps.

One of the primary reasons that the elbow joint is such a complex structure is the fact that most of the muscles which cross the elbow joint also cross either the shoulder joint or wrist joint as well. A muscle "crosses" a joint if the joint lies between the muscle's origin and insertion points. Of the eight muscles examined in this study only PT, BR and B are "one-joint" muscles, as they cross only the elbow joint. Triceps medial and lateral heads are also one-joint
Figure 30. Biceps Brachii and Orientation of Humerus Bone.
muscles; however, the long head of Triceps originates on the scapula, hence crossing the shoulder joint in addition to the elbow joint. Both heads of Biceps Brachii originate above the shoulder joint and are also two-joint muscles. In the experiments, the shoulder joint was not loaded so that all activity generated in Biceps Short, Biceps Long, and Triceps Long was the result solely of torque sustained at the elbow joint.

The other two muscles of the eight, Flexor Carpi Radialis (WF) and the Extensor Carpi Radialis Longus (WE), are also two-joint muscles, as they originate above the elbow joint and insert below the wrist joint, crossing both joints. These two muscles, though generally associated primarily with the wrist joint, could sustain torque at the elbow joint. In addition to these two muscles, there are seven other muscles crossing both the wrist and elbow joints, which also could sustain elbow joint torque.* Thus the significance of WF and WE as elbow flexor muscles in the eight-muscle system studied is of considerable importance. This question will now be considered in light of several observations of the experimental data and the anatomical model.

In Tables 3, 4, and 5, as mentioned previously, the muscle ordering patterns would be obscured by including WF and WE, as in over 50 per cent of the cases, WF and/or WE would be at least fifth in the ordering of the eight muscles. At maximum torque, the sum of the IEMG

* Extensor Carpi Radialis Brevis, Extensor Digitorum, Extensor Digitii Minimi, Flexor Carpi Radialis, Palmaris Longus, Flexor Carpi Ulnaris, and Flexor Digitorum Superficialis.
of WF and WE is 66 per cent of the IEMG of the prominent elbow flexor muscle for Prone 90° arm position and 77 per cent for Supine 90°.

Similarly, at maximum torque, the sum of the IEMG of WF and WE is 17 per cent of the sum of IEMG of all eight muscles for Prone 90° and 18.5 per cent for Supine 90°. These four percentages are average values for the 12 experiments. If all eight muscles contributed equally to sustaining the elbow torque, the sum of the IEMG of WF and WE would be 25 per cent of the sum of IEMG of all eight muscles.

Compared to this figure the percentages of 17 per cent and 18.5 per cent appear quite substantial. Again, IEMG is assumed to be directly proportional to the load sustained by a muscle. Thus, this high degree of activity in WF and WE indicates considerable load sustained by these muscles. This load could be sustained at either the wrist or elbow joint. However, the subject was asked to maintain the wrist in a neutral (not flexed or extended) position to eliminate wrist joint loading. Also a hand rest was provided so that the weight of the hand could not contribute to torque sustained at the wrist joint. Hence wrist joint torque is concluded to be negligible so that virtually all EMG activity results from the elbow joint torque. Reference to the geometric anatomical model also demonstrates the considerable significance of WF and WE in elbow flexion. For 90° elbow angle, the sum of the moment arm lengths of WF and WE is 14.4 per cent of the sum of the moment arm lengths of all eight muscles. For 150° elbow angle, this figure increases slightly to 16.2 per cent.

The experimental data shows notable EMG activity in Extensor
Carpi Radialis Longus and Flexor Carpi Radialis. The anatomical model indicates that non-negligible torque could be sustained by these two muscles at the elbow joint. These observations along with the fact that seven other muscles similar to WF and WE exist, suggest that the two-joint muscles thought to be primarily associated with the wrist joint are quite possibly highly influential at the elbow joint as well. If so, then the simple eight-muscle system studied in this thesis is not a complete picture of the elbow joint muscular system. The following observation of Data Set I also suggests this incompleteness.

Presented in Figures 31 and 32 for Prone $90^\circ$ and Supine $90^\circ$ respectively, is the Total IEMG ($BL + BS + B + BR + PT + WF + WE - T$) at maximum Torque plotted versus Maximum Torque. There are 12 points, one for each experiment, on each graph. The most striking feature of these graphs is that IEMG appears to be a decreasing function of torque. For both graphs, the lowest value of torque corresponds to the highest value of IEMG and the highest value of torque corresponds to the lowest value of IEMG. This phenomenon is particularly peculiar in view of the fact that all of the individual graphs show IEMG to be an increasing function of torque. One possible explanation is incompleteness. That is, EMG activity was generated in muscles other than those monitored, muscles such as the group of wrist joint muscles. Of course, other explanations exist, but incompleteness is as plausible as the others. The concept of the inadequacy of the eight muscles studied to fully describe the entire elbow joint muscle system is discussed again in Section 7.2.1.
Figure 31. Total IEMG Magnitudes versus Maximum Torque Prone 90°.
Figure 32. Total IEMG Magnitudes versus Maximum Torque Supine 90°.
7.1.2 Data Set II

This set includes repetitive run data for Experiment 9 (Prone 90° and Supine 90°) and Experiment 10 (Prone 90°). Presented data consists of two graph pages for each experiment, with four graphs per page, and one muscle per graph. As discussed in detail on page 100, the 13-load sequence was repeated four times within approximately 30 minutes. For each muscle, the four repetitive runs are designated on the graphs as "1", "2", "3", and "4". The graphs are found on pages 213 - 220 in Appendix A.

As mentioned before, during an experiment, electrodes tend to migrate into the tissue with repeated muscle contraction. Jonsson and Reichman [51] (1967) found with intramuscular electrodes, that the difference in the degree of activity in two electromyographic recordings made on the same occasion was slight. The results of this thesis study agree. As shown in the graphs, the difference in the four repeated runs is slight, for both Experiments 9 and 10. There is an unusual shift in noise level for the third run, Experiment 9, Prone 90°, on WF muscle, attributed to equipment difficulties. Of more interest, is the increase in data scatter for Supine 90° as opposed to Prone 90° for Experiment 9, noticeable on all eight muscles. The supine position was run after the prone position, introducing the possibility of fatigue as cause of the scatter increase. However, even with data scatter, the similarities in the repetitive runs is not obscured. It is concluded that electrode migration is negligible and that consistency within an
experiment is definitely achieved.

7.1.3 Data Set III

This set consists of data for Subjects DTL and SST. DTL performed the experiment on two separate occasions (Experiments 10 and 12) three days apart, and SST was a subject three times (Experiments 6, 9, and 11) with five months between Experiment 6 and 9 and three days between Experiment 9 and 11. For DTL, data are presented as two graph pages for each of the two arm positions (Prone 90° and Supine 90°). The two Experiments 10 and 12, are designated "1" and "2" respectively, on the graphs. Data for SST is also presented for the same two arm positions, and the three Experiments 6, 9, and 11 are designated "1", "2", and "3" respectively. The graphs are included in Appendix A on pages 221-229.

Consider first the two experiments with subject DTL. Activity in Triceps is nil, always. The three major elbow flexors, Brachialis, Biceps Long and Biceps Short, are very similar on the two experimental occasions. Differing noticeably, however, are Brachioradialis, Pronator Teres, Extensor Carpi Radialis, and Flexor Carpi Radialis. In terms of electrode placement, it is reasonable to suspect that distinguishing among the larger muscles, such as Biceps and Brachialis, is more accurate than for the smaller muscles. For subject DTL, particular difficulty was encountered in distinguishing between Pronator Teres and Flexor Carpi Radialis.

Considerably more variation among experiments on different occasions is evident for subject SST. Note that for Triceps, the
first experiment indicates more activity (at the higher loads) than the second and third experiments. Surface electrodes, capable of picking up EMG activity in other muscles, were used in the first experiment on Triceps, while intramuscular electrodes with restricted local pick-up, were used in the second and third experiments. In the case of Prone 90\(^\circ\), the variation among the three experiments is more prominent for the larger muscles (B, BL, BS) than for the smaller muscles (BR, PT, WF, WE) though the variation for all is noticeable. These results contradict the results for subject DFL and indicate that the significance of electrode placement is independent of muscle size. Tables 7 and 8 demonstrate this variation. Table 7 presents the values of IEMG for all muscles except Triceps, for all three experiments, as measured at the maximum torque of Experiment 11. Also shown is the value of IEMG as a percentage of the total IEMG of the seven muscles. The percentages for each muscle vary considerably among the three experiments. Only BR, BL and WF are at all similar. Table 7 presents the ordering of the muscles according to the IEMG at maximum torque. Again for all experiments, a different muscle order is the case. Reference to Table 2 reveals that at large torques, the PEF muscle also varies for Experiments 6, 9, and 11. In general, the data presented for the experiments performed by the same subject on different occasions seriously questions day-to-day consistency.

As previously discussed in Section 3.3 the three studies cited [51,52,53] also report poor day-to-day experimental reproducibility in quantitative EMG studies. These studies are quite recent (1968
Table 7. Experiments No. 6, 9, and 11 for Subject SST
Values of IEMG at Maximum Torque of Experiment 11.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>IEMG values at maximum torque of Experiments 11 - 35 ft-lbs</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Total EMG = 2.05 mv</td>
</tr>
<tr>
<td></td>
<td>Experiment No. 6</td>
</tr>
<tr>
<td></td>
<td>value (% total)</td>
</tr>
<tr>
<td>BL</td>
<td>.26 (12.7)</td>
</tr>
<tr>
<td>BS</td>
<td>.44 (21.5)</td>
</tr>
<tr>
<td>F</td>
<td>.65 (31.7)</td>
</tr>
<tr>
<td>BR</td>
<td>.42 (20.4)</td>
</tr>
<tr>
<td>WE</td>
<td>.07 (3.4)</td>
</tr>
<tr>
<td>PT</td>
<td>.04 (1.95)</td>
</tr>
<tr>
<td>WF</td>
<td>.17 (6.3)</td>
</tr>
</tbody>
</table>
Table 8. Experiments No. 6, 9, and 11 for Subject SST
Muscle Ordering According to IEMG at Maximum Torque.

<table>
<thead>
<tr>
<th>Experiment Number</th>
<th>Prone 90°</th>
</tr>
</thead>
<tbody>
<tr>
<td>6</td>
<td>B</td>
</tr>
<tr>
<td>9</td>
<td>BS</td>
</tr>
<tr>
<td>11</td>
<td>BS</td>
</tr>
</tbody>
</table>

| Supine 90° |  
|------------|----------|
| 6          | BL       | B  | BR | BS | T | PT |
| 9          | BS       | BL | B  | BR | PT | T  |
| 11         | BR       | B  | BS | PT | BL | T  |
and 1970) and, as stated by Komi and Buskirk [53] repeatability of measurements is "perhaps an area, which has been most neglected in electromyography." Poor reproducibility is due to several factors, all of which are difficult to control and one of which is difficult to precisely define as well. The concept of the latter factor is expressed nicely by Komi and Buskirk: "It seems illogical to expect excellent repeatability of measurement in such a complex biological phenomenon as whole muscle activity." Any number of known and unknown parameters from the psychological state of the subject to his physical condition might alter the EMG activity in a muscle (see pp. 63 - 64). A well-defined but difficult-to-control parameter is electrode placement within a muscle. In spite of attempts to maintain insertion point consistency, such as using constant length needles and having the same anatomist insert all electrodes, the electrode position in a muscle certainly varies on different experimental occasions. Komi and Buskirk cite this variation as a reason for poor experimental reproducibility. Also, inconsistency in electrodes will cause day-to-day inconsistency in EMG data. The length of stripped wire might vary slightly, although all electrodes were hand-made and the length of wire measured under a microscope. More critical as well as difficult to control is the electrode geometries relative to one another and relative to the myoneural junctions (see Section 2.1.4). Since the EMG signal measured with bipolar intramuscular electrodes is the potential difference between the two electrodes, the distance between the electrodes is significant, and of course, the positions of the electrodes
relative to myoneural junctions is most important. Both of these geometric parameters will vary considerably for different experimental occasions.

In any case, day-to-day inconsistency is a definite problem in quantitative electromyography studies and is demonstrated in Data Set III. Whether attributed to electrode placement or physiological variation of biological phenomenon, poor experimental reproducibility is an extremely difficult problem to solve.

7.1.4 Data Set IV

This set consists of data for Experiments 3 and 8 for which the effect of three different loading methods on the IEMG-Torque relationship is shown graphically. Graphs are on pages 230-234 in Appendix A. The three methods of loading the elbow joint are:

1. sequential loading of 13 loads from Load 1 up to Load 13,
2. sequential unloading from Load 13 down to Load 1, and
3. random loading.

All loads, of course, were applied by the subject as isometric torque to his elbow joint. For each muscle, the three loading methods are plotted on one graph for comparison.

The IEMG-Torque relationship is definitely affected by the different loading methods. These results contradict those of DeVries [4] who in 1965 with surface EMG from Biceps Brachii observed no hysteresis from loading and unloading (see p. 51). Of particular interest is the comparison of sequential loading and sequential unloading methods. In sequential loading, the elbow is isometrically flexed, while when the elbow joint is sequentially unloaded, it is
isometrically extended. Comparison of these two methods reveals a

type of muscle hysteresis. A somewhat analogous demonstration of

hysteresis is the magnetization and subsequent demagnetization

of an iron ring. In this case, the hysteresis effect is attributed to

residual magnetization and is considered an energy loss in the form

of internal friction dissipation of heat. However, this analogy is

incomplete in that the muscle is allowed to return to its unloaded

state between loading and unloading, while the magnet was immediately
demagnetized from the magnetized state.

The hysteresis could have originated in the load device,
but in calibration of the device (see page 87) this possibility was
checked and virtually no hysteresis was observed. The hysteresis
effect, as shown in the graphs, varies for each muscle. There is

some internal friction generated as the individual muscle fibers
contract, however, the variation of the hysteresis effect with muscles
suggests that the functional aspects of the muscles are involved.
Certainly the function of the muscles is different for elbow flexion
(sequential loading) than for elbow extension (sequential unloading).
For example, Brachialis is the active flexor of the elbow; but as
the flexed elbow is extended, Brachialis contracts to oppose and
regulate the extension of the elbow. In elbow flexion, Brachialis is
thus a prime mover, while in elbow extension, it is an antagonist.
Such variation in function of a muscle with the two loading methods
is a probable explanation for the observed muscle hysteresis.
7.2 Numerical Techniques of Data Analysis

7.2.1 Coefficient Matrix System

Consider the eight muscles, Biceps Long, Biceps Short, Brachialis, Triceps, Brachioradialis, Extensor Carpi Radialis, Pronator Teres and Flexor Carpi Radialis as the complete muscular system sustaining the isometric torque about the elbow joint. Further, assuming that IEMG is a linear function of the torque, the following system equation can be written:

\[ a_1 BL + a_2 BS + a_3 B + a_4 T + a_5 BR + a_6 WE + a_7 PT + a_8 WF = \text{Torque} \]

where: BL, BS, B, T, BR, WE, PT, WF are values of IEMG (mv) in the appropriate muscles at a particular value of Torque. T and WE are signed as negative. Torque is the particular value of applied torque in ft-lbs. \( a_1, a_2, \text{etc.} \) are coefficients (ft-lbs/mv) constants.

Both the values of IEMG and the Torque are known experimentally determined values, while the eight constant coefficients are unknown. By assuming that inter-subject differences are tolerable, eight such equations, one for each of eight subjects, can be considered as a system of eight equations and eight unknowns, namely the coefficients. Expressing such a system in matrix notation:
where the subscripts on the $a$ coefficients are indicative of the muscles (1-8), and the subscripts on the IEMG and Torque values indicate the subject number (1-8).

Comparison of the results from this technique with the geometric anatomical model is obvious when the following equations are compared:

$$M \ (\text{ft-lbs}) = F_1L_1 + F_2L_2 + \ldots + F_8L_8$$

$$\text{Torque} \ (\text{ft-lbs}) = BL_1a_1 + BS a_2 + \ldots + WFa_8$$

The first equation is the sum of moments equation developed from the geometric anatomical model. For each muscle, the terms $F_iL_i$ and
BL_{a, f} for Biceps Long, for example, should be comparable.

Several variations of this basic method were explored, in order to investigate the effect of eliminating certain muscles and the effect of nonlinearity of the EMG-Torque relationship. The following combinations were tested:

1. All eight muscles with Torque squared.
2. All eight muscles with Torque cubed.
3. All eight muscles with IEMG values squared.
4. Exclusion of PT.
5. Exclusion of PT with Torque squared.
6. Exclusion of PT with IEMG values squared.
7. Exclusion of WE and T.
8. Exclusion of WE and T with Torque squared.
9. Exclusion of WE and T with IEMG values squared.
10. Exclusion of WF and WE.
11. Exclusion of WF and WE with Torque squared.
12. Exclusion of WF and WE with IEMG values squared.
13. Exclusion of T, WF, and WE.
14. Exclusion of T, WF, and WE with Torque squared.
15. Exclusion of T, WF, and WE with IEMG values squared.

Also, sets of coefficients were obtained for eight values of torque for each of the four arm positions. This work was done on the Burroughs 5500 at Georgia Institute of Technology.

Basically, the results indicate that something is definitely amiss with this numerical model of the muscular system. Not only are
the magnitudes of the coefficients questionable, but also in many cases, the signs of the coefficients are completely contrary to anatomical facts, that are unquestionably correct. For example, the coefficient of the IEMG of Biceps Long is in some instances negative, suggesting that this muscle is an antagonist to elbow flexion and is effectively a negative contributor to sustaining the applied torque. This is simply not true. As it is quite probable that inter-subject differences are in fact, so great as to be intolerable in this analysis, the technique was applied to one subject using EMG values from the muscles at different values of torque as input. However, results for the one subject analysis are similar to those for the inter-subject analysis.

Two possible explanations for these results are suggested. Summing the coefficient-weighted values of IEMG and equating the sum to the total torque sustained by the elbow joint musculature, inherently assumes that the complete elbow joint muscle system is described by the eight muscles studied in this thesis. Such gross failure of this model suggests that the model is incomplete, specifically that the eight muscles are only a part of the entire elbow joint musculature. In the discussion of Data Set I, Extensor Carpi Radialis and Flexor Carpi Radialis, were shown to be quite significant in elbow joint movements. Recall that these two muscles are two of a group of nine two-joint muscles generally associated with the wrist joint. The seven remaining muscles of this group might very well be the missing part of the elbow joint muscle system.

The other basic assumption of this model is that integrated
EMG can be expressed as some function of the load sustained by a muscle. Again, the complete failure of the model perpetrates questioning of the validity of the model basic assumptions. Qualitative analysis of integrated EMG-load information indicates that basically as the load sustained by a muscle increases, the electromyographic activity in the muscle also increases. Anatomy and physiology of the motor unit and muscle in general, indicate this type of relationship. However, a quantitative analysis such as have been presented suggests that, though information regarding the load sustained by a muscle may be contained in the electromyographic signals, this information may be obscured by integration of the EMG. In fact, other ways of reducing EMG may reveal the load information in the signal.

7.2.2 Least Squares Curve Fitting

The functional relationship between the electromyographic activity in a muscle and the load sustained by the muscle has been the subject of several investigations. In the 19 papers discussed in detail in Section 3.2 a variety of muscles, experimental techniques, and data handling methods were involved. In general, however, two schools of thought prevailed: one supporting a linear relationship and the other quadratic. In light of this extensive investigation of the topic and the large amount of EMG-Torque data obtained for eight muscles in this thesis study, it was deemed worthwhile to examine the functional relationship between IEMG and Integrated Torque.

Using a Least Squares Curve Fitting program on the Burroughs
5500, the approximate degree of each IEMG-Torque curve was determined. The general functional relationship between IEMG and Torque is given by the following:

\[ a_0 + a_1 X + a_2 X^2 + a_3 X^3 + \ldots + a_n X^n = Y \]

where:  
\( X = \text{Integrated Torque} \)
\( Y = \text{Integrated EMG} \)
\( a's = \text{Coefficients with } n \text{ designating the equation degree.} \)

For each IEMG-Torque graph, the 13 \((X,Y)\) points were input to the Least Squares program and the equation degree specified. Program output was the values of the specified number of coefficients. Statistical parameters to determine goodness of fit were unavailable, thus the equation degree was estimated in the following manner. For each set of 13 points, several sets of coefficients were calculated for different specified equation degrees. By comparing the values of the coefficients for the different degrees, the significance of a higher degree term could be determined. Consider the following example of Experiment 2, Pronator Teres, prone forearm, and elbow angle of \(90^\circ\). Two sets of coefficients were calculated:

\[
\begin{array}{cccc}
  a_0 & a_1 & a_2 & a_3 \\
  3.1 \times 10^{-2} & 1.66 \times 10^{-2} & 7.51 \times 10^{-5} & - \\
  2.43 \times 10^{-2} & 1.99 \times 10^{-2} & 3.85 \times 10^{-4} & 7.2 \times 10^{-6} \\
  \text{n = 2} & & & \text{n = 3}
\end{array}
\]
In both sets of coefficients, the terms of higher degree than one are insignificant relative to the terms of degree zero and one. A simple arithmetic check reveals that this is indeed the case. For the thirteenth load of 39 ft-lbs, the contribution of the second degree \( (a_2) \) term is 14 per cent of the total (sum of the three terms on the left-hand side of equation (1)); at the twelfth load of 34 ft-lbs, the contribution is 13 per cent, at eleventh load 28 ft-lbs, 10.6 per cent, and for the first ten loads, the contribution is less than 10 per cent. The degree of this particular IEMG-Torque function is concluded to be one \( (a_1) \); that is, the curve is fit quite well by a straight line. Reference to the appropriate graph on page 160 in Appendix A will confirm the linearity.

Obviously, the higher degree equation will provide more accurate curve fit; in fact, for 13 points input, an equation of degree 13 would provide perfect fit to the data. However, accuracy, in this study is not the ultimate goal, but rather accuracy while maintaining a reasonable degree of workable simplicity. This study aimed at exploration of the linear-quadratic question, and determination of the effect of such parameters as muscle, subject, forearm position, elbow angle, and electrode position on the functional relationship of integrated EMG and integrated torque applied to a system of muscles. The somewhat qualitative method of estimating the degree of the functional relationship and the resulting accuracy are considered quite adequate to investigate these questions.

Presented in Table 9 are the results of the Least Squares Curve Fitting technique. For each of the eight muscles, 12 experiments,
Table 9. Results of Least Squares Curve Fitting.

<table>
<thead>
<tr>
<th>Experiment Number</th>
<th>Forearm Position: Prone (P) Supine (S); Elbow Angle: 90° or 150°</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>1 2 3 4 5 6</td>
</tr>
<tr>
<td></td>
<td>Q L Q L C Q L Q Q Q L L L L Q Q Q Q L Q Q Q Q L Q Q Q Q L Q</td>
</tr>
<tr>
<td></td>
<td>90 90 150 150</td>
</tr>
<tr>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>7 8 9 10 11 12 13</td>
</tr>
<tr>
<td></td>
<td>Biceps L.</td>
</tr>
<tr>
<td></td>
<td>C Q C Q L Q Q Q Q L L Q Q C Q L L Q Q Q Q L Q Q Q Q L Q</td>
</tr>
<tr>
<td></td>
<td>90 90 150 150</td>
</tr>
<tr>
<td>K = Constant</td>
<td>L = Linear</td>
</tr>
</tbody>
</table>

K = Constant  
L = Linear  
Q = Quadratic  
C = Cubic
and four arm positions, is listed the estimated degree of the best fit equation. Symbols used are

- \( K \) = constant horizontal straight line (degree 0),
- \( L \) = linear (degree 1),
- \( Q \) = quadratic (degree 2), and
- \( C \) = cubic (degree 3).

Regardless of muscle, experiment, forearm position or elbow angle, IEMG appears to be predominantly a quadratic function of Integrated Torque. Each of the muscles, except Triceps and the Wrist Extensor, is also, in general, quadratic. In the majority of cases, IEMG in Triceps is a constant function of Torque, which is obvious from the graphs and the lack of activity in Triceps. Wrist Extensor is about half the time constant and the other half quadratic, with a few exceptions. It is interesting to note that for every muscle, there is at least one case of each of the three degrees, linear, quadratic, and cubic. Terms of higher order than three were in all cases definitely negligible (less than 3 per cent significant for all load values). Although for a particular muscle, a certain degree predominates, the degree definitely varies among the subjects. For example, consider the Biceps Long, for which the degree is Quadratic for Experiment 3, yet Linear for Experiment 4, and finally Cubic for Experiment 7. With respect to the question of variation of degree with forearm position and elbow angle, for Experiments 1 - 10, each of which has four arm positions, 64 per cent of the graphs have the same degree for at least three of the four arm positions. For Experiments 9 - 12, for which each has two arm positions, 62.5 per cent of
the graphs have the same degree for both arm positions. Of the remaining 37.5 per cent, 12.5 per cent have degrees for the two arm positions that differ by two degrees, e.g., L and C, and 25 per cent have degrees differing by only one degree, e.g., L and Q.
CHAPTER VIII

CONCLUSIONS

The purpose of this study as stated in the Introduction was to make a quantitative electromyographic study of the elbow joint, and to investigate the functional relationship between time integrated EMG and time integrated load. The conclusions are based on both experimental data and analytical model information, which was discussed in detail in the previous chapter. The first two conclusions pertain directly to the purpose of the thesis.

Conclusion Number 1

As justified in Section 2.2, eight muscles were chosen to model the elbow joint musculature. EMG activity from these muscles was recorded simultaneously with the torque output of the elbow joint. The EMG signal for each muscle was integrated and graphed as a function of the integrated torque output from the joint. The integrity of this electromyographic model of the elbow joint can now be evaluated.

(a) Three observations demonstrate validity of the model.

First, the major elbow flexor muscles, as shown by the data and the geometric model, agree with elbow joint anatomy presented in Primary Anatomy [16]. Brachialis is shown to be the prominent elbow flexor muscle at small torques. Brachialis, Biceps Brachii and Brachioradialis are the prominent elbow flexor muscles at large torques.
(see p. 108). Secondly, the muscle ordering according to IEMG at maximum torque (excluding wrist muscles) shown Biceps Long, Biceps Short, Brachialis and Brachioradialis as the top muscles for seven of 12 cases, with Pronator Teres replacing one of these four muscles in the remaining five cases. With exception of Pronator Teres, the moment arm lengths of the geometric model support these data results (see p. 110). Third, for Biceps Brachii, Brachialis and Brachioradialis, the IEMG is greater for elbow angle of 150° than 90°, a data result which is also supported by the moment arm lengths of the anatomical model (see p. 115). Thus, regarding the major elbow joint flexor muscles, the eight-muscle model is concluded to be quite adequate.

(b) Those areas in which the model is inadequate are shown by three observations. Each will be discussed separately along with possible explanations for these observations.

1. The experimental data shows considerable IEMG activity in Extensor Carpi Radialis (WE) and Flexor Carpi Radialis (WF) as a result of the torque sustained by the elbow. The geometric model also demonstrated that substantial elbow joint torque could be sustained by these two muscles. This evidence, along with the fact that there are seven other muscles similar to WF and WE, suggest that the two-joint muscles thought to be primarily associated with the wrist joint should be included in an elbow joint model. This explanation, however, assumes that integrated EMG is proportional to the load sustained by a muscle, and that IEMG activity in WF and WE is indicative of con-
siderable load sustained by these muscles. This assumption may not be valid, a possibility that is discussed in relation to the next two observations as well.

2. In Figures 31 and 32, the Total IEMG (sum of eight muscles) appears to be a decreasing function of Maximum Torque. Three possible explanations for this phenomenon are presented. One conclusion is that unmeasured EMG in muscles other than those monitored is non-negligible. These unmonitored muscles could very well be the seven two-joint muscles primarily associated with the wrist joint, as discussed above. A second possible conclusion, also discussed above, is that the assumption that integrated EMG is proportional to the load sustained by a muscle, is not completely valid. This explanation is also applicable to the next observation and will be discussed in detail in that regard. A third possible explanation for Figures 31 and 32, involves the basic assumption that a single sample of the EMG activity in the muscle is representative of the total activity in the muscle. If this assumption is invalid, it would be necessary to include a weighting factor on the single EMG sample to reflect muscle size in some manner such as motor units per muscle. This factor, however, would be most difficult to estimate.

3. The third observation demonstrating model failure is the results of the Coefficient Matrix System (see p.133). The three explanations for the observation just discussed might also account for the inaccurate results of the Coefficient Matrix analysis. Summing the coefficient-weighted values of IEMG and equating the sum to the total torque sustained by the elbow joint, assumes
that the complete elbow joint musculature is described by the eight muscles, i.e., that the total EMG generated by the torque was measured in these eight muscles. Quite possibly, the total EMG generated included EMG in unmonitored muscles, such as the seven two-joint "wrist" muscles. The Coefficient Matrix analysis also assumes that the integrated EMG activity in a muscle is proportional to load sustained by the muscle. Once again, the results question this assumption. Quantitative analysis of integrated EMG-load information indicates that basically as the load sustained by a muscle increases, the EMG activity in the muscle also increases. However, these quantitative observations suggest that, though information regarding the load sustained by a muscle may be contained in the EMG signal, this information might be obscured by integration of the EMG. Other ways of reducing EMG may better reveal the load information in the signal. A third possible explanation for the Coefficient Matrix results is that the assumption that the single sample of the EMG activity in the muscle is representative of the total activity in the muscle, is invalid. Again, in that case, appropriate muscle size factors would have to be included as coefficients of the IEMG values in the Coefficient Matrix System.

Conclusion Number 2

This conclusion pertains to the second part of the thesis purpose. The functional relationship between time integrated EMG and time integrated Torque is dependent on the muscle, subject, elbow angle, forearm position, etc. and varies from a proportional to a
cubic function (see p. 137). It must be recognized that, in this study, the functional relationship investigated is that between integrated EMG and the integrated Total Torque sustained by the elbow joint, not the load sustained by each individual muscle. This distinction is important since the assumption that integrated EMG is proportional to the load sustained by a muscle was questioned in evaluation of electromyographic integrity.

The following four conclusions are extraneous to the stated purpose of the thesis but nevertheless represent worthwhile information.

**Conclusion Number 3**

Elbow angle and forearm position are definitely important parameters in evaluating elbow joint performance. For most of the muscles, one of the two forearm positions and/or elbow angles resulted in more EMG activity than the other at comparable torques. The manner in which the muscles interact varies with elbow angle and forearm position; definite patterns are established by the elbow angle and/or forearm position in the ordering of the muscles according to IEMG at maximum torque (see p. 113).

**Conclusion Number 4**

Hysteresis was demonstrated in muscle, and it was shown to vary for each muscle, which suggests that functional aspects of the muscles are involved. The two methods of loading the elbow joint required the muscles to act in two functionally different ways, antagonistically and agonistically, resulting in the hysteresis
Electrode migration was found to be negligible, as indicated by achievement of consistency within an experiment (see p. 125).

Day-to-day reproducibility of data was poor and concluded to be almost impossible to achieve. Proposed causes for the lack of data consistency for one subject tested on different days are electrode placement variation and electrode inconsistencies. Also recognized as a factor in such reproducibility is the variation in any number of known and unknown parameters, from the psychological state of the subject to his physical condition, which might alter the EMG activity in the muscles (see p. 126).
APPENDIX A

GRAPHS: AVERAGE INTEGRATED ENG VERSUS AVERAGE INTEGRATED TORQUE
APPENDIX A

GRAPHS: AVERAGE INTEGRATED EMG VERSUS AVERAGE INTEGRATED TORQUE

Data Set I

Data Set I consists of Figures 33 - 72, each presenting EMG-Torque data for each of the eight muscles monitored. For each of Experiments 1 - 8, data for four arm positions were taken:

Arm Position

<table>
<thead>
<tr>
<th>Forearm Position</th>
<th>Elbow Angle</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Prone</td>
<td>90°</td>
</tr>
<tr>
<td>2. Supine</td>
<td>90°</td>
</tr>
<tr>
<td>3. Prone</td>
<td>150°</td>
</tr>
<tr>
<td>4. Supine</td>
<td>150°</td>
</tr>
</tbody>
</table>

Data for only the first two arm positions listed were taken for Experiments 9 - 12.

Each figure consists of four graphs, and data for two muscles are plotted on each graph. The abbreviations used throughout this thesis for the muscles is used on the graphs: Biceps Long (BL), Biceps Short (BS), Brachialis (B), Triceps (T), Brachioradialis (BR), Extensor Carpi Radialis (ER), Pronator Teres (PT), and Flexor Carpi Radialis (WF).
Preceding the set of graphs for each experiment is a brief evaluation, presented in outline form, of the graphs.
## Data Set I

### Experiment 1 - Subject TN

1. Subject appeared ill at ease and found it difficult to relax completely. Subject tended to anticipate loads.

2. Noise levels in EMG channels:
   a. Levels reasonably constant throughout experiment
   b. Sometimes, noise level is not indicated by first data point, as subject was not relaxed. Examples: PT(2), PT(1), six muscles (4).
   c. WE high (.35), but there is no indication that the high noise level masked the EMG activity.

3. EMG values (millivolts) at Maximum Load (3/4 ft-lbs)
   - **Prone 90°**
     - BS  .92  .85  .67  .65  .55  .50  .37  .26
     - BR  .90  .89  .58  .50  .38  .34
     - B   .72  .56  .54  .27  .12
     - PT  .81  .72  .56  .38  .34
     - WF  .60  .60  .58  .54  .25
     - T   .36  .34  .25  .12
     - WE  .26  .24  .22  .10
     - BL  .10  .10  .10  .10  .10
   - **Supine 90°**
     - BS  1.12  .90  .89  .58  .52  .51  .38  .34
     - BR  .90  .89  .58  .50  .38  .34
     - B   .72  .56  .54  .27  .12
     - PT  .81  .72  .56  .38  .34
     - WE  .60  .60  .58  .54  .25
     - T   .36  .34  .25  .12
     - BL  .26  .24  .22  .10
     - BS  .10  .10  .10  .10  .10
   - **Prone 150°**
     - BS  1.04  .81  .72  .56  .38  .27  .18
     - B   .72  .56  .54  .27  .12
     - T   .36  .34  .25  .12
     - WE  .26  .24  .22  .10
     - BL  .10  .10  .10  .10  .10
   - **Supine 150°**
     - BS  1.0  .89  .60  .60  .58  .54  .25  .12
     - BR  .60  .60  .58  .54  .25  .12
     - WE  .26  .24  .22  .10
     - BL  .10  .10  .10  .10  .10

4. Initial Recruitment Patterns
   - (1) BS  B  BR  BL  PT  T  WF  WE
   - (2) BS  B  BL  BR  WE  PT  T  WF
   - (3) BS  B  BR  BL  PT  WF  WE  T
   - (4) BS  B  BL  BR  WE  PT  T  WF
5. Conclusions

a. Generally, BS is the PEF muscle.

b. Patterns in ordering of muscles appear to be established by elbow angle rather than forearm position.

c. A lot of activity in all muscles.
Figure 33. Experiment 1, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 34. Experiment 1, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 35. Experiment 1, Forearm Prone, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 36. Experiment 1, Forearm Supine, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Experiment 2 - Subject BB

1. Subject had difficulty achieving proper loads and errors were frequent. Maximum load was difficult for subject to hold and subject had tendency to anticipate the next load in sequence.

2. Noise levels in EMG channels:
   a. Levels constant throughout experiment.
   b. No relaxation problems.

3. EMG values (millivolts) at maximum load (38 ft-lbs)

   (1) Prone 90° BL BS PT BS WE BR WF T
       .73 .50 .43 .36 .26 .18 .05 .17
   (2) Supine 90° BL BS B WF PT WE BR T
       .37 .55 .54 .59 .32 .30 .15 .10
   (3) Prone 150° BL BS BS BS BS WE WE BR T
       .78 .60 .50 .45 .33 .18 .11 .06
   (4) Supine 150° BL BS BS WE WF WE WE BR T
       .36 .64 .60 .50 .45 .22 .18 .13

4. Initial Recruitment Patterns

   (1) IT BB BS WE B BR WF T
   (2) BR BS WE IT B BR WF T
   (3) BL BS PT B BR WE WF T
   (4) BR BS PT B BR WF WF T

5. Conclusions
   a. BL is the HIF muscle
   b. Definitely affected by forearm position and elbow angle variation.
but, the exact manner in which affected is indeterminate.

c. PT more active than BR throughout all four runs.
Figure 37. Experiment 2, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)

Ordinate: Average Integrated EMG (millivolts)
Figure 38. Experiment 2, Forearm Supine, Elbow Angle 90°
Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 39. Experiment 2, Forearm Prone, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 40. Experiment 2, Forearm Supine, Elbow Angle 150°
Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Experiment 3 - Subject SXI

1. SXI was an ideal subject. He was able to completely relax, achieve proper loads without errors, achieve maximum loads without difficulty, and did not anticipate loads.

2. Noise levels in EMG channels:
   a. B noise level is 0.2 on first three runs, then dropped to 0.1, indicating equipment variation. Otherwise, levels were constant throughout experiment.
   b. Not relaxed on first data point on BL on first run (1). Otherwise, relaxation is complete, and noise level is properly indicated by first data point.

3. EMG values (millivolts) at maximum load (40 ft-lbs).
   (1) Prone 90° BL BS BR B WE PT WF T
       1.16 1.08 .38 .30 .16 .08 .06 0
   (2) Supine 90° BL BS WE B BR PT WF T
       1.10 1.04 .54 .40 .38 .20 .11 0
   (3) Prone 150° BL BS B BR WE PT WF T
       1.13 1.02 .48 .47 .42 .30 .19 0
   (4) Supine 150° BL BS B BR WE PT WF T
       1.38 1.07 .53 .48 .23 .10 .06 0

4. Initial Recruitment Patterns
   (1) BL BS BR B WE PT WF T
   (2) BL B BS BR WE PT WF T
   (3) BL BS BR B WE PT WF T
   (4) BL BS BR B WE PT WF T
5. Conclusions

a. Biceps is FEF muscle, with BL slightly more activity than BS.

b. All four runs are quite similar with three IEMG magnitude domains with (BL,BS) in the first, (B,BR,WE) in the second, and (PT,WF,T) in the third.

c. Forearm position and elbow angle, have no obvious effects on either magnitudes or patterns of muscle activity.
Figure 41. Experiment 3, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 42. Experiment 3, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 43. Experiment 3, Forearm Prone, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 44. Experiment 3, Forearm Supine, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Data Set I

Experiment 4 - Subject EDD

1. Subject appeared quite sensitive to pain and had difficulty achieving higher loads.

2. Noise levels in EMG channels:
   a. Levels constant throughout experiment.
   b. Triceps never completely relaxed. Slight activity in BS (4) on first data point. Otherwise, no relaxation problems.

3. EMG values (millivolts) at maximum load (37 ft-lbs).
   (1) Prone 90° BS BR WF BL B WE T PT
       .61 .65 .45 .43 .26 .19 .10 .04
   (2) Supine 90° BS BR WF BL B WE T PT
       .76 .64 .60 .48 .32 .24 .05 .04
   (3) Prone 150° BR BS WF BL B WE T PT
       .68 .63 .62 .59 .33 .33 .05 .05
   (4) Supine 150° BR WS BL BS B WE T PT
       .62 .61 .51 .38 .36 .31 .07 .06

4. Initial Recruitment Patterns
   (1) BL BS BR B WF WE PT T
   (2) BL B BS BR WF WE PT T
   (3) BL BS B BR WF WE PT T
   (4) BL B BR BS WT WE PT T

5. Conclusions
   a. Generally, BS FEF muscle for 90° elbow angle and BR for 150°.
   b. BL definitely FEF muscle initially (first three torque values).
c. Last four muscles same throughout experiment:
   B  WE  T  PT  for Maximum IFMG
   WF  WE  PT  T  for initial recruitment.

d. High values of WF at high loads yet low initial recruitment
   while low values of B at high loads yet high initial recruit-
   ment suggests "replacement" of WF for B at higher loads.

e. Patterns appear slightly more elbow angle dependent than
   forearm position.
Figure 45. Experiment 4, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 46. Experiment 4, Forearm Supine, Elbow Angle 90°
Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 47. Experiment 4, Forearm Prone, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 48. Experiment 4, Forearm Supine, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Experiment 5 - Subject DSK

1. Subject was cooperative and had little difficulty achieving proper loads according to instructions.

2. Noise levels in EMG channels:
   a. Levels constant throughout experiment.
   b. No relaxation problems, except BS on (1) and (2) which was slight.

3. EMG values (millivolts) at maximum load (40 ft-lbs).
   (1) Prone 90° B PT WF BR BS BL WE T
       0.80 0.76 0.72 0.33 0.14 0.10 0.05 0.05
   (2) Supine 90° B PT BS WE BL WF BR T
       0.81 0.53 0.48 0.46 0.42 0.40 0.29 0.05
   (3) Prone 90° B WF PT BR BL T BS WE
       0.90 0.89 0.86 0.27 0.09 0.08 0.02 0.02
   (4) Supine 150° B PT BS WF BR BL WE T
       0.86 0.44 0.43 0.40 0.39 0.31 0.19 0.11

4. Initial Recruitment Patterns
   (1) B PT WF BS BL BR WE T
   (2) B PT WF BS BL BR WE T
   (3) B PT WF BR BS BL WE T
   (4) B PT WF BR BS BL WE T

5. Conclusions
   a. B is definitely PEF muscle.
   b. Patterns definitely forearm position dependent rather than
elbow angle:

(1) Prone: PT and WF almost as large as B

Supine: PT second behind B and only half of B in magnitude.

(2) General magnitude divisions:

Prone: (B, PT, WF) (BR) (BS, BL) (WE, T)

|     | .75 | .3 | .1 | .02 |

Supine: (B) (PT, BS, WE, BL, WF) (BR) (T)

|     | .85 | .4 | .3 | .1 |
Figure 49. Experiment 5, Forearm Prone, Elbow Angle 90°
Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 50. Experiment 5, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 51. Experiment 5, Forearm Prone, Elbow Angle 150°

- **Abscissa:** Average Integrated Torque (foot-pounds)
- **Ordinate:** Average Integrated EMG (millivolts)
Figure 52. Experiment 5, Forearm Supine, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Experiment 6 - Subject SST

1. No notable difficulties, except for scar on subject's forearm over PT and WF insertion area. See page 94 of text for further details.

2. Noise Levels in EMG channels:
   a. Strange change in noise level on BL: 0.1 for (1) and (3) Prone, 0.2 for (2) and (4) Supine

   Otherwise, constant.

   b. No relaxation problems.

3. EMG values (millivolts) at maximum load (40 ft-lbs)

   (1) Prone 90° B BS BR BL WF WE T PT
       .73  .50  .47  .29  .22  .10  .10  .06
   (2) Supine 90° BL B BR WF BS WE T PT
       .62  .56  .51  .51  .38  .17  .12  .08
   (3) Prone 150° B BS BR BL WF WE T PT
       .58  .50  .47  .33  .27  .18  .11  .05
   (4) Supine 150° WF BL B BR BS WE T PT
       .77  .75  .58  .46  .42  .13  .13  .00

4. Initial Recruitment Patterns

   (1) B = BS = BR = BL WF WE PT T
   (2) B BR BS = BL = WF WE PT T
   (3) B BR BS BL WF WE PT T
   (4) B BR BS BL WF WE PT T
5. Conclusions

a. Generally, B UEF muscle, with (4) exception.

b. Interesting interaction between WF and BS, with WF low on Prone while BS high; reverse on Supine.

c. Last three muscles always (WE, PT, T)

e. Appears to be forearm position dependent rather than elbow angle. That is, patterns in muscle ordering established by forearm position (Supine 90° and 150° similar patterns and likewise, Prone 90° and 150° similar).
Figure 53. Experiment 6, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 54. Experiment 6, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 55. Experiment 6, Forearm Prone, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 56. Experiment 6, Forearm Supine, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Data Set I

Experiment 7 - Subject ENT

1. Subject slightly impatient (fatigue possibilities), but otherwise cooperative. Subject, Dr. Scharf, did electrode insertion himself.

2. Noise Levels of EMG channels:
   a. Levels constant throughout experiment.
   b. No relaxation problems.

3. EMG values (millivolts) at maximum load (42 ft-lbs)
   (1) Prone 90° B WE PT BR BL WF T BS
       .79 .72 .39 .37 .17 .10 .03 .03
   (2) Supine 90° B PT BR BL T WF BS WE
       .83 .53 .37 .09 .06 .06 .04 .03
   (3) Prone 150° B BR PT WE BL BS T WF
       .69 .27 .25 .23 .18 .09 .03 .0
   (4) Supine 150° B PT BR BL BS WF T WE
       .80 .45 .30 .20 .10 .08 .04 .0

4. Initial Recruitment Patterns
   (1) B BR BS WE BL WF T PT
   (2) B BR PT BS BL WF WE T
   (3) B BR BS PT WE BL WF T
   (4) B BR PT BL BS WF WE T

5. Conclusions
   a. Definitely B PEF muscle throughout.
   b. Biceps, both BL and BS, very insignificant.
   c. Position patterning definitely indicated by initial recruitment patterns; also by similarity of Supine 90° and 150°.
Also WE significant on Prone, but very low on Supine.

Also, PT more active on Supine than Prone.
Figure 57. Experiment 7, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 58. Experiment 7, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 59. Experiment 7, Forearm Prone, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 60. Experiment 7, Forearm Supine, Elbow Angle 150°
Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
**Data Set I**

**Experiment 8 - Subject BST**

1. Subject could not achieve the last load, so it was dropped from the sequence. Thus only 12 data points for each muscle are presented on the graphs.

2. Noise levels on EMG channels
   - Levels all constant throughout experiment.
   - No relaxation problems, except for BR (4) and PT (4).

3. EMG values (millivolts) at maximum load (35 ft-lbs)
   - **Prone 90° BS BL B WF BR PT T WE**
     - 0.83 .70 .62 .38 .35 .33 .03 .02
   - **Supine 90° BS BL B BR WF PT T WE**
     - 0.87 .85 .76 .50 .46 .33 .16 .10
   - **Prone 150° BS BL B WF BR PT T WE**
     - 0.90 .71 .58 .50 .42 .30 .10 .00
   - **Supine 150° BS BL B BR PT T WE WF**
     - 0.83 .79 .68 .47 .20 .14 .03 .00

4. Initial Recruitment Patterns
   - **(1) B PT WF BS BL BR WE T**
   - **(2) B BL BS BR PT WF WE T**
   - **(3) B BS BL PT BR WF WE T**
   - **(4) BL BS B PT BR WE T WF**

5. Conclusions
   - a. Definitely Biceps PEF muscle with BS slightly greater than BL.
   - b. Appears rather unaffected by forearm position and elbow
angle variations:
1. Shape of curves for each muscle very similar throughout.
2. Except for WF (4), ordering of IEMG at maximum load same for all four cases. Also suggests possible equipment problems, and actually activity on WF (4).
c. Effect of elbow angle: B is closer to BL at 90° than 150°.
d. Effect of forearm position: BL and BS closer together in magnitude both on curves and as seen on IEMG at maximum load when forearm supine than prone.
Figure 61. Experiment 8, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 62. Experiment 8, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 63. Experiment 8, Forearm Prone, Elbow Angle 150°
Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 64. Experiment 8, Forearm Supine, Elbow Angle 150°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Data Set I

Experiment 9 - Subject SST (again)

1. This experiment as well as Experiment 10 - 12 were done approximately four months after the first eight. Steve Wolf did insertions and triceps is now monitored with intramuscular rather than surface electrodes.

2. Noise levels on EMG channels:
   a. Levels constant throughout experiment.
   b. No relaxation problems, however, on second run, first data point was repeated erroneously.

3. EMG values (millivolts) at maximum load (42 ft-lbs)
   (1) Prone 90° BS B BL BR WF WE PT T
       1.23 .83 .76 .70 .34 .08 .06 0
   (2) Supine 90° BS BL B BR WF WE PT T
       1.23 .76 .77 .71 .50 .32 .14 .04

4. Initial Recruitment Patterns:
   (1) B BR BL BS WF PT WE T
   (2) B BR BL BS WF PT WE T

5. Conclusions:
   a. Definitely BS HGF muscle.
   b. On initial recruitment patterns, note that first load causes much activity in B but practically none in BS and BL.
   c. Only difference between prone and supine, except for greater IEMG for supine, is switch with B and BL in ordering sequence.
   d. Note, that though same load sustained in both cases, supine required total IEMG - 4.49 mv. while prone only 4.0 mv.
Figure 65. Experiment 9, Forearm Prone, Elbow Angle 90°
Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 66. Experiment 9, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Data Set I

Experiment 10 - Subject DTL

1. Subject became slightly ill, but was able to complete experiment. Had some difficulty distinguishing between PT and WF, and during Experiment 12 with this same subject.

2. Noise levels on EMG channels:
   a. Levels constant for each channel throughout experiment.
   b. No relaxation problems except for WF (1).

3. EMG values (millivolts) at maximum load (42 ft-lbs)
   (1) Prone 90° B BS BL BR PT WF T WE
       .63 .58 .28 .20 .16 .13 .01 0
   (2) Supine 90° B BS BL BR PT WF WE T
       .67 .55 .30 .23 .06 .04 .04 0

4. Initial Recruitment Patterns
   (1) B BS BL BR PT WF WE T
   (2) B BS BL BR PT WF WE T

5. Conclusions:
   a. Appears unaffected by change in forearm position; same pattern for both prone and supine by both maximum load EMG values and initial recruitment patterns. Magnitudes vary somewhat, however.
   b. Definitely B PEE muscle.
   c. Observe, increase in EMG on last load in both runs, when subject erroneously skipped next to the last load and resulting held last load for two data points. Fatigue possibly explains increase in EMG.
Figure 67. Experiment 10, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 68. Experiment 10, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)

Ordinate: Average Integrated EMG (millivolts)
Experiment II - Subject SST (again)

1. Subject's arm was quite sore and bruised from Experiment 9 conducted two days prior to Experiment II.

2. Noise levels on EMG channels:
   a. Levels for each channel constant throughout experiment.
   b. No relaxation problems. Again, however, on both runs, first data point is repeated erroneously.

3. EMG values (millivolts) at maximum load (37 ft-lbs)
   (1) Prone 90° BS BR B WE BL PT WF T
       .57 .53 .49 .47 .42 .34 .12 0
   (2) Supine 90° WE BR B BS PT BL WF T
       .73 .64 .54 .52 .50 .47 .18 0

4. Initial Recruitment Patterns
   (1) B BL BS BR WE PT WF T
   (2) B BL BR WE BS PT WF T

5. Conclusions:
   a. No definite PEF muscle; except for WF and T, all other muscles appear to be substantially contributing to sustaining the torque. Top four are always: BS BR B WE.
   b. Seems to be very definitely affected by changes in forearm position:
      1. EMG maximum values: BS and WE switch places in ordering.
      2. Initial recruitment: BS, WE, and BR move around.
   c. None of the eight muscles has much activity initially.
Figure 69. Experiment 11, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 70. Experiment 11, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)

Ordinate: Average Integrated EMG (millivolts)
Data Set I

Experiment 12 - Subject DTL (again)

1. Subject sore and bruised from Experiment 10, only three days previously. Again, had much difficulty distinguishing between WF and PT.

2. Noise levels on EMG channels:
   a. Levels constant throughout experiment.
   b. No relaxation problems.

3. EMG values (millivolts) at maximum load (37 ft-lbs)
   (1) Prone 90° BR B PT BS WF BL WE T
       .74 .71 .57 .50 .42 .35 .22 0
   (2) Supine 90° B BR PT BS BL WF WE T
       .59 .58 .59 .49 .23 .23 .10 0

4. Initial Recruitment Patterns:
   (1) B PT BR BL BS WF WE T
   (2) B PT BS BR BL WF WE T

5. Conclusions:
   a. FEF muscles are (almost equally) BR and B.
   b. Effect of forearm position
      1. Maximum EMG: switch of WF and BL; change in magnitudes.
      2. Initial Recruitment: three muscles (BR,BL,BS) move about.
   c. Relatively unaffected by forearm position, as curves are quite similar for both prone and supine.
   d. Initially, B, PT, and BR are definitely more active than Biceps.
   e. On first run (1), observe increase in EMG due to subject
skipping next to last load and two data points being recorded for last load. Possible fatigue.
Figure 71. Experiment 12, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 72. Experiment 12, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordone: Average Integrated EMG (millivolts)
Data Set II

Data Set II includes repetitive run data for Experiment 9 (Prone 90° and Supine 90°) and Experiment 10 (Prone 90°). Presented data consists of two figures for each experiment, with four graphs per page, and one muscle per graph. As discussed previously, the 13-load sequence was repeated four times within approximately 30 minutes. For each muscle, the four repetitive runs are designated on the graphs as "1", "2", "3", and "4". This Data Set consists of Figures 73 through 78.
Figure 73. Experiment 9, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 74. Experiment 9, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 75. Experiment 9, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 76. Experiment 9, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 77. Experiment 10, Forearm Prone, Elbow Angle 90°
Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 78. Experiment 10, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 79. Exp. 10 & 12, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Data Set III

Data Set III consists of data for subjects DTL and SST. DTL performed the experiment on two separate occasions (Experiments 10 and 12), and SST was a subject three times (Experiments 6, 9, and 11). For DTL, data are presented as Figures 79 through 82 for two arm positions (Prone 90° and Supine 90°). Experiments 10 and 12 are designated "1" and "2", respectively, on the graphs. Data for SST is also presented for the same two arm positions (Figures 83 through 86), and Experiments 6, 9, and 11 are designated "1", "2", and "3" respectively.
Figure 80. Exp. 10 & 12, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 81. Exp. 10 & 12, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 82. Exp. 10 & 12, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 83. Exp. 6, 9, & 11, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 84. Exp. 6, 9, & 11, Forearm Prone, Elbow Angle 90°
Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 85. Exp. 6, 9, & 11, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 86. Exp. 6, 9, & 11, Forearm Supine, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Data Set IV

Data Set IV consists of data for Experiments 3 and 8 for which the effect of three different loading methods on the EMG-Torque relationship is shown graphically. The three methods of loading the elbow joint are:

1. sequential loading of 13 loads from Load 1 up to Load 13,
2. sequential unloading from Load 13 down to Load 1, and
3. random loading.

These three methods are designated on the graphs as "L", "U", and "R" respectively. For each muscle, the three loading methods are plotted on one graph for comparison. This data set consists of Figures 87 through 90.
Figure 87. Experiment 3, Forearm Prone, Elbow Angle 90°
Abcissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 88. Experiment 3, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 89. Experiment 8, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
Figure 90. Experiment 8, Forearm Prone, Elbow Angle 90°

Abscissa: Average Integrated Torque (foot-pounds)
Ordinate: Average Integrated EMG (millivolts)
APPENDIX B

SAMPLING RATE DETERMINATION
The one Kilohertz sampling rate for the time integration data reduction technique was selected after completion of a two part study:

(1) a survey of literature pertaining to the significant energy levels of the electromyographic signal, and

(2) a sampling rate study performed on a selected portion of the data collected in this thesis study. The results of these two investigations are now presented.

The appropriate sampling rate is dependent on the upper limit of the frequency range of significant energy in the electromyographic signal. This upper limit is somewhat controversial, namely because it is defined by the key word "significant", which is in itself a somewhat opinionated term. Most investigators of the myoelectric energy spectra state an upper limit of 200 Hz. Walton [7] in 1952, used an audio-frequency spectrometer to find that the spectra of large limb muscles contains peaks at 100 and 200 Hz. Basmajian in Muscles Alive [13] references Hayes [72] (1960) as reporting the significant energy frequency band as 20-200 Hz for surface electrodes. Scott [9] in 1967, used a digital computer to determine the auto-correlation function and then the power density spectrum. He reports that the myoelectric signal has significant energy only in the frequency range
of approximately 30-200 Hz. Scott used intramuscular wire electrodes. Using needle electrodes in 1963, Kaiser and Petersen [21] reported peak frequencies at 50, 200, and 800 Hz. However, the latter figure of 800 Hz has not been confirmed. On the basis of the majority of opinions, for this thesis study it is concluded that the upper limit of the significant energy frequency range of the myoelectric signal is approximately 200 Hz.

The sampling rate study was performed on the data from Experiment 3 for three muscles, Biceps Long, Biceps Short and Brachioradialis. The data reduction technique of time integration described on page 102 of the text was used for various digital sampling rates. For all three muscles, rates (Hz) of 200, 500, 600, 700, 800, 900, 1000, 1500 and 2000 were used, and for Biceps Short, the additional rates of 25, 50, 75, 100, 125, and 250 were tried. For all sampling rates, ten data points were obtained for each one-second record, and a numerical average of these ten points was calculated to represent the integrated value of the one-second record. For example, for a 500 Hz sampling rate, 500 digital samples were obtained and by summing successive samples in lots of 50, ten data points were acquired. Similarly for a rate of 2000 Hz, the 2000 digital samples for the one-second record were summed in lots of 200 to obtain the ten data points.

Results of this sampling rate study were presented graphically (graphs not shown) as Integrated EMG plotted on the ordinate and Integrated Torque on the abscissa, both quantities in arbitrary units. These graphs show that below 500 Hz, the information in the myoelectric signal definitely deteriorates, with most obvious distortion beginning
at 200 Hz. At rates of 800 and 900 Hz, the graphs are quite similar to those at 1000 Hz, and at rates above 1000 Hz, the information appears unchanged. From this study, 1000 Hz was concluded to be an adequate sampling rate. In light of the 200 Hz upper limit the significant energy frequency range found in the literature survey, the sampling rate of 1 KHz also appears quite adequate.


12. Lovejoy, J. and Basmajian, J. V., "Functions of Popliteus Muscle in Man - a Multifactorial Electromyographic Study," J. Bone and


68. Okamoto, Tsutomu, "A Study of the Variation of Discharge Pattern During Flexion of the Upper Extremity," reprint, Department of Physical Education, Kansai Medical School, Osaka.


