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The Relationship Between Integrated Electromyography and Torque at Varying Joint Angles and Intensities of The Static Contraction of the Rectus Femoris Muscle in Normal Female Subjects

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THE RELATIONSHIP BETWEEN INTEGRATED ELECTROMYOGRAPHY AND TORQUE AT VARYING JOINT ANGLES AND INTENSITIES OF THE STATIC CONTRACTION OF THE RECTUS FEMORIS MUSCLE IN NORMAL FEMALE SUBJECTS

by

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This Thesis submitted by Mary Lee Metelak Leikas in partial fulfillment of the requirements for the Degree of Master of Science from the University of North Dakota is hereby approved by the Faculty Advisory Committee under whom the work has been done.

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This Thesis meets the standards for appearance and conforms to the style and format requirements of the Graduate School of the University of North Dakota, and is hereby approved.

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Title: The Relationship Between Integrated Electromyography and Torque at Varying Joint Angles and Intensities of the Static Contraction of the Rectus Femoris Muscle in Normal Female Subjects

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Mary Lee Lukas

Signature

April 19, 1983

Date

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ABSTRACT

Sports medicine specialists, researchers, and rehabilitation specialists are frequently concerned with quantifying muscular activity for the purpose of studying muscle function. Two instruments that are currently used are the Cybex II Isokinetic Unit and integrated electromyography (IEMG). The Cybex measures torque and IEMG measures the neuromuscular activity associated with a contracting muscle.

When measuring muscle function on the Cybex and IEMG, one would not expect a linear relationship to exist between the two values if biomechanical, anatomical, and physiological factors are allowed to change. One would expect a linear relationship to exist if these factors were held constant.

The hypotheses were tested in a two part study using thirty female subjects. For the first phase, the subject was required to perform a maximal static exercise of knee extension for 10 seconds at 90^{0}, 60^{0}, 30^{0}, and 10^{0} of knee flexion. Simultaneous readings of torque and IEMG were recorded for each contraction. There was no linear relationship between torque and IEMG at varying joint angles ($r = -0.101$).

For the second phase, the subject was required to perform a submaximal static exercise of knee extension for 10 seconds at 20 percent, 40 percent, 60 percent, and 80 percent of their maximal torque produced at 60^{0} of knee flexion in phase one. All measurements were taken with the knee at 60^{0} of flexion. A linear relationship was found to exist

between IEMG and torque of an isometric contraction when all biomechanical, anatomical, and physiological factors were constant.

The Cybex II Isokinetic Unit and IEMG have both proven to be useful tools in studies of human skeletal muscle. Under certain restricted conditions, the two evaluative tools tell us relatively the same thing about muscle activity. Under changing conditions, the two evaluative tools cannot be related.

CHAPTER ONE

THE PROBLEM

Quantification of muscular activity is essential to researchers, rehabilitation specialists, and sports medicine specialists. Results of muscle testing provide information that: 1) enables the study of muscle function in various activities; 2) enables one to assess the degree of muscle injury; and 3) enables one to measure the amount of progress of an exercise program. Two instruments that are currently used are the Cybex II Isokinetic Unit and integrated electromyography (IEMG). The Cybex measures torque (foot-pounds). Torque is defined as the force times the length of the arm on which it acts (Brunnstrom 1972). The IEMG measures the amount of tension (micro-volts) produced in a given muscle. The EMG signal is the electrical manifestation of the neuromuscular activation associated with a contracting muscle.

A motor unit consists of the anterior horn cell of the spinal cord, the nerve fiber (axon) and the muscle fibers it innervates (Brunnstrom 1972). Muscles vary in innervation ratios (number of muscle fibers innervated by one nerve fiber) from 1:2000 (large limb muscles) to 1:6 (extraocular muscles). Muscles which function to control fine movements generally have small motor units while muscles responsible for gross motor activities have large motor units (Basmajian 1978). Motor units function according to the all or none law. The nerve fiber transmits a nerve impulse to the group of muscle fibers it innervates and the stimulated fibers contract simultaneously and completely. A stronger stimulus would not cause the muscle fibers to contract any further. The all or none law applies to the motor unit but not to the muscle as a whole (Mathews and Fox 1976). The strength of a muscle contraction is graded by the size of the active motor units, the firing frequency of each active motor unit, and the number of motor units that are active. Basmajian reports that in the normal pattern of recruitment, the small motor units are recruited initially followed by large motor units and as the force increases further, frequency of firing of all motor units occurs (Basmajian 1978).

The EMG signal is dependent on the number of active motor units, frequency of firing, and spike shape.

Torque is dependent on this neuromuscular activity plus biomechanical, anatomical, and physiological factors. These factors are line of action of the muscle over the joint (this changes as movement occurs), angle of insertion, cross section of the muscle, functional excursion, relative length of the muscle, and speed of contraction.

When measuring muscle function on the Cybex and integrated electromyography, one would not expect a linear relationship to exist between the two values if biomechanical, anatomical, and physiological factors are allowed to change. One would expect a linear relationship to exist if these factors were held constant.

This paper is a report on the study of the relationship between integrated electromyography (IEMG) and torque as measured on the Cybex. Part one deals with the relationship when biomechanical, anatomical, and physiological factors are manipulated. Part two reports on the relationship when these factors are constant.

CHAPTER II

REVIEW OF THE LITERATURE

There is a vast amount of literature on electromyography and torque. This paper deals only with the study of muscle function during isometric contractions at varying joint angles and at varying intensities of force. Therefore, the review of the literature will be limited to the effect of varying the joint angle on EMG and torque and the relationship of muscle tension (force) and EMG. The literature cited here were chosen because they were studies done using similar equipment or the same exercise (knee extension). Some pieces described the methodology of EMG definitively and established what EMG measures. Others explained biomechanical, anatomical, and physiological factors and how the EMG signal and torque are affected by them.

Effect of Varying the Joint Angle on EMG and Torque

Several investigators have concluded that no constant relation exists between force and EMG in a muscle when it is allowed to change in length. Given a maximal contraction of a muscle in a shortened position, the muscle force is small, while the electrical activity is high. The opposite is true of a muscle in the stretched position; i.e., the force is greater, while the EMG is reduced. Leverage factors change the muscle force as the joint angle changes and further complicates the relationship between force and EMG.

Lunnen, et al. examined the relationship between integrated EMG activity, torque, and length of human biceps femoris muscle during the isometric contraction while the knee joint angle remained constant (Lunnen, Yack and LeVeau 1981). Changes in muscle length were produced by changes in hip joint angle. Twelve females and four males were involved. Torque measurements were provided by a force dynamometer incorporated within the Cybex II System. The EMG activity was processed by telemeter EMG equipment designed and constructed at the University of North Carolina School of Medicine. Surface electrodes were used to pick up the signal. Their results indicated: (1) EMG activity decreased as the muscle lengthened and (2) the torque increased as the muscle length increased.

Moritz, et al. studied nineteen healthy subjects in an attempt to analyze the relative influence of motor unit activity, muscle force lever, and muscle length on the torque of the knee extensor muscle group at different joint angles (Moritz, Svantesson and Haffajee 1973). They report that the relative muscle length of the quadriceps is directly proportional to the flexion angle. The IEMG, the strain gauge tension, and the joint angle were recorded simultaneously by a 4-channel Mingograph. Measurements were taken at 10°, 30°, 50°, 70°, and 90° of knee flexion. In the first part of their study, knee extensor torque and EMG were recorded during maximum voluntary isometric contraction. They observed peak torque at 50° of flexion, a reduction of 50 percent of maximum value at 10° and a scattering of values at 90°. They found the curve of the IEMG to show an increase from 10° to 90°. In the second part, a measurement of torque was taken at a constant, submaximal EMG output. They found that the average torque curve differed little from that of maximum voluntary effort. Since the myoelectrical output was constant, the shape of the torque curve was attributed to variations of the lever and of the muscle length. This group conducted the same experiment on sixteen subjects who had the patella removed on one side. They found the torque these subjects produced on knee extension was considerably reduced between 20^0 and 60^0. They concluded that the patella has its greatest importance for the lever length of the muscle force within this range.

Williams and Stutzman measured the torque curves of ten subjects for knee extension at 30^0 increments. The mean peak torque was at 60^0 followed by 90^0 (Williams and Stutzman 1959). They concluded that the length-tension factor was far more important than muscle lever length for the quadriceps.

A group of investigators have studied the relationship between muscle tension and EMG in amputees having cineplastic muscle tunnels using a strain-gauge dynamometer and integrated electromyography (Inman, Ralston, Saunders, Feinstein and Wright 1952). This procedure enabled them to observe the length-tension phenomenon of muscle without leverage complications. They stated that because the tension developed by a muscle varies with the length of the muscle, one should not expect any parallelism between EMG and force with varying lengths of muscle. They found that the amplitude of EMG decreases and the tension is greater when the muscle is stretched. When the muscle is short, the force is small and the EMG is maximal. They stated that this lack of a constant relationship between tension and EMG in a muscle changing in length is due to the existence of the length-tension relationship in muscle and to factors not yet explained.

In 1959, a group of researchers reported on a series of experiments they conducted to test the hypothesis that a major factor in producing the lower EMG in the lengthened muscle is the initiation of inhibitory afferent impulses from tendon organs of the stretched muscle (Libet, Feinstein and Wright 1959). They examined the effect of the muscle length on the maximal voluntary EMG in normal subjects before and after anesthetizing the tendon below the musculo-tendinous junction. They also compared the maximum voluntary EMG's of patients with degeneration of afferent fibers from the muscle (tabes dorsalis) with the EMG's of normal subjects. They found that tendon anesthesia eliminated the decrease in EMG that normally occurs in a stretched muscle, therefore supporting their hypothesis. The patients with tabes dorsalis further supported the role of afferents in inhibiting impulses from a stretched muscle. The EMG in the lengthened position was much closer to the EMG in the intermediate and short position in these patients.

Relation of Isometric Contraction (Tension) and EMG

Several investigators have concluded from their research that the electrical output as revealed by IEMG closely parallels the tension in a given muscle contracting isometrically.

Moritani and deVries examined the relationship between the surface integrated electromyogram (IEMG) and force of isometric contraction (Moritani and deVries 1978). They were concerned with these four problems of previous research using IEMG. (1) lack of standardization of experimental methods and procedures, (2) inadequate control of the fatigue factor, (3) small movements at the joint allowing for systematic shortening of the muscle during "isometric" contraction, and (4) co-

contraction. By controlling these factors and clearly defining the experimental conditions, they observed a linear relationship between IEMG and isometric force. They tested twenty-six young healthy males across the entire domain of forces in the elbow flexor group. The force of contraction was established by a hydraulic dynamometer. EMG activity was monitored using unipolar surface electrodes with an integration period of one second.

Inman, et al. observed a parallelism between the integrated EMG and tension during isometric contractions of a muscle (Inman, et al. 1952). They repeatedly observed this phenomenon in the pectoral, triceps, and forearm muscles of eleven subjects with cineplastic tunnels. The muscle had been freed from its insertion and attached to a dynamometer which measured the tension of voluntary isometric contractions. They concluded that the integrated EMG may be used as an index of developed tension when all electrical factors are kept constant and the contraction is isometric.

Lippold conducted thirty experiments on different subjects and found the relationship between isometric tension of a voluntarily-contracting human muscle and its integrated electromyogram to always be directly linear (Lippold 1952). He studied the gastrocnemius-soleus group. Muscle strength was measured with a dynamometer while simultaneous electromyograms were recorded by means of an amplifier and cathode-ray oscillograph. Recordings were taken at ten different strengths of contractions. Integrated electrical output and the strength of contraction of a muscle both increase as the number of active motor units increase and as the frequency at which these units repetitively contract increases.

CHAPTER III

METHODOLOGY

Subjects

Thirty female subjects ranging in age from 18 to 28 with an average age of 21.8 years participated in the study. Most of the subjects were involved in collegiate track, cross-country, or field hockey teams. All of them exercised routinely. The mean height of the subjects was 65.8 inches with a range of 62 inches to 72.5 inches. The mean weight was 127.9 pounds with a range of 106 pounds to 157 pounds. Anthropometric data for the subjects are included in table 6 of the Appendix. They had no demonstrable defects in muscle function.

Instrumentation

The torque measurements were provided and recorded using the Cybex II Isokinetic Unit. The Cybex II measures dynamic or isometric strength at every point in the range of motion by providing an automatic accommodating resistance against a lever moving at a controlled velocity (degrees/second). Resistance is equal to the force exerted by the subject at every point in the range of motion. A dual channel recorder consisting of a dynamometer and electrogoniometer provides a continuous printout of the torque (foot-pounds) and the joint angle of the limb being tested. The Cybex II was calibrated according to manufacturer's instruction (Isolated 1980). Calibration was done before testing, after every five subjects, and randomly throughout the study. Pre-calibrated lead weights were used to calibrate the torque scale.

The EMG signal was amplified by a Cyborg J33 biofeedback unit. The J33 served as a pre-amplifier for the BL900 Processor. The BL900 was used to integrate the raw EMG. This process provides a value for the total area under the EMG curve and plots this area against time (McLeod 1973). The IEMG is a good index to the raw EMG, but it does not differentiate between increased frequency and increased amplitude (Hall 1970). A meter provided a direct reading of the integrated EMG in micro-volts. The shortest integration period on the BL900 is ten seconds. The BL900 integrator was calibrated using a ten micro-volt peak-to-peak sine wave current. Calibration was done before initial testing, after every five subjects, and randomly throughout the study.

Beckman silver-silver chloride ten millimeter surface electrodes were used to pick up the electrical changes in the muscle. Bouisset and Maton demonstrated that surface electrodes correlate well with intramuscular electrodes when measuring muscle tension (Bouisset and Maton 1972).

Procedure

A pilot study was done to check if fatigue was a factor in this study. Several investigators have found that the relation between EMG and isometric tension changes during fatigue (Edwards and Lippold 1956, Basmajian 1978). Although it remains linear, the slope of the line increases. Using shorter rest periods in the pilot study than were used in the experiment, there was no evidence that fatigue affected the results.

Each subject was introduced to the equipment and the testing procedure was explained before subject preparation began. Informed consent was obtained. The subjects were dressed in gym shorts and shoes were removed. The procedure was standardized by taking all the measurements on the subject's right knee. This also eliminated the need to move the equipment between subjects. The subject was positioned on the Cybex according to protocol for knee extension exercises (Isolated 1980). The axis of the knee joint was aligned with the dynamometer input shaft and the shin pad was attached above the subject's ankle joint. The position angle channel was then synchronized with the subject's anatomical position. Pelvic, torso, and thigh stabilization straps were applied. The experimental set-up is shown in figure 1.

The motor point of the rectus femoris muscle was located using an electrical galvanic stimulator. The skin was prepped using sand paper and acetone to decrease skin resistance. Impedance was reduced to less than 4,200 ohms in all cases. The surface electrodes were filled with redux paste and attached to the skin with electrode adhesive discs longitudinally over the muscle belly 2 centimeters apart and equidistant from the motor point. The ground electrode was placed medial to them. The electrodes were left in place for the entire procedure. Electrode placement is pictured in figure 2.

For the first phase of the experiment, the subject was required to perform a maximal static exercise of knee extension for 10 seconds at 90^0, 60^0, 30^0, and 10^0 of knee flexion. The lever arm was adjusted to the appropriate knee angle by reading the joint position scale on the printout. The speed selector was set at 0^0 per second to provide an immovable resistance to knee extension. The subject was given two

Fig. 1. The experimental set-up

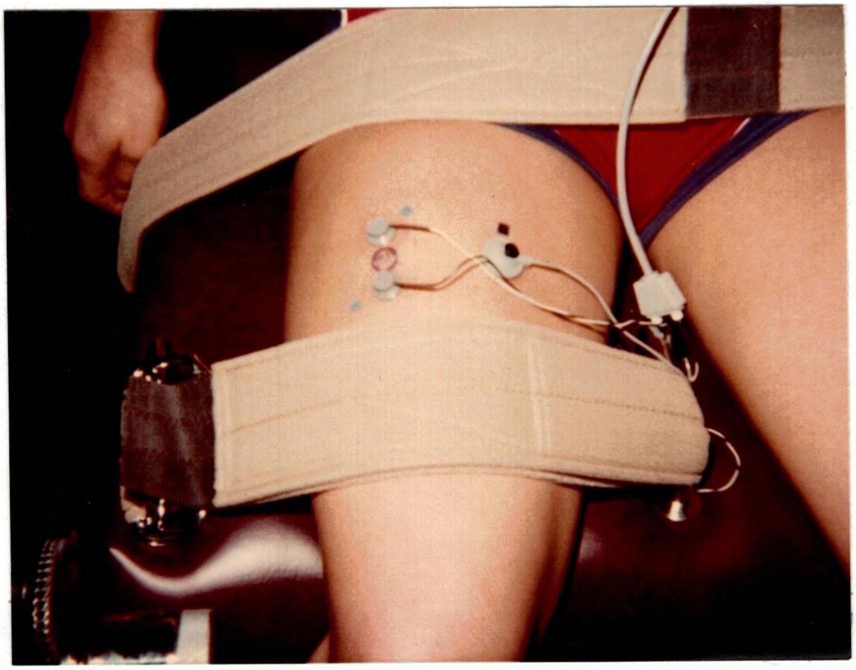

Fig. 2. Position of recording electrodes on the right thigh

submaximal trials at each angle to familiarize themselves with the equipment and procedure. The subject was then instructed to push as hard as possible until told to stop. There was a three-minute rest between contractions and a ten-minute rest between the two phases of the experiment to control the fatigue factor. Simultaneous readings of torque and IEMG were taken and recorded for each contraction.

For the second phase, the subject was required to perform a submaximal static exercise of knee extension for 10 seconds at 20 percent, 40 percent, 60 percent, and 80 percent of their maximal torque produced at 60^{0} of knee flexion in phase one. The equipment set-up remained the same. The subject's position was unchanged with the exception that all measurements were taken with the knee at 60^{0} of flexion. Previous studies have demonstrated that the isometric torque curve peaks at 60^{0} of flexion for the quadriceps (Williams and Stutzman 1959 and Mendler 1967). The percentage of torque for each intensity was calculated. The subject was instructed to watch the dynamometer and to keep the needle deflection at the indicated level until told to relax. The dynamometer dial was marked at 12 foot-pound increments. This created a margin of error when the subject was asked to produce a specific force. Again, there was a three-minute rest period between contractions. The IEMG was read and recorded for each contraction.

CHAPTER IV

RESULTS

The null hypotheses state:

1. There is no linear relationship between IEMG (micro-volts) and torque (foot-pounds) of the static contractions of the rectus femoris muscle at varying joint angles

2. In the variation of range of intensity of contraction of the rectus femoris muscle, there is no linear relationship between IEMG and torque

The .05 level of significance was established.

The raw data for each subject are given in tables 7 and 8 of the Appendix. Phase one and phase two of the experiment were analyzed separately. Descriptive and inferential statistics were computed to aid in interpretation of the data (means, standard deviations, Pearson's product-moment correlation) (SAS 1982).

The mean torque and IEMG values recorded during phase one from the static knee extension exercise at four different knee angles are presented in table 1.

TABLE 1--Descriptive Statistics for Torque and IEMG of a Maximal Static Knee Extension Exercise for Thirty Subjects at 90°, 60°, 30°, and 10° of Knee Flexion

Angle	Foot-Pounds			Micro-Volts		
	$\overline{X}$	S	Range	$\overline{X}$	S	Range
90°	100.20	21.37	71-170	148.30	36.03	90-218
60°	108.93	17.43	66-144	167.37	48.64	55-255
30°	65.80	13.88	48-117	180.53	53.61	72-310
10°	35.60	9.00	21- 63	186.60	51.65	88-310

The greatest torque was produced at 60° (108.93 ft-lbs) followed by 90° (100.20 ft-lbs) and 30° (65.80 ft-lbs) respectively. The lowest mean torque was recorded at 10° (35.60 ft-lbs), while the highest mean electrical activity was measured at 10° (186.60 μV). The smallest amount of electrical activity was produced at 90° (148.30 V) followed by 60° (167.37 μV) and 30° (180.53 μV) respectively. A graph of these mean values is depicted in figure 3.

A Pearson product-moment correlation was employed for analysis of the relationship between the torque and the IEMG at four different joint angles and between torque and angle and between IEMG and angle. The results of this analysis are presented in tables 2 and 3.

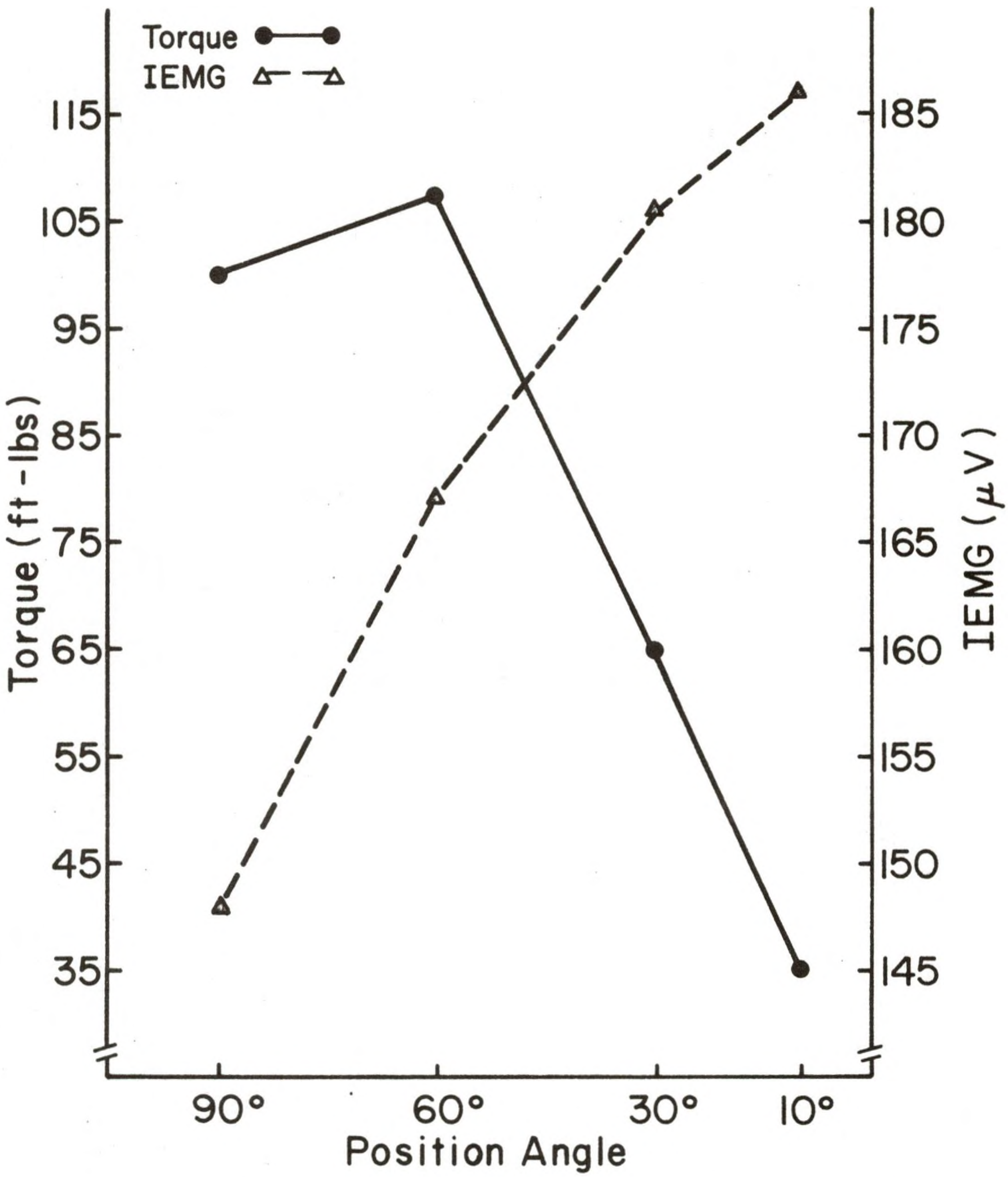

Fig. 3. Mean torque and IEMG values of a maximal static knee extension exercise for thirty subjects at four different joint angles

TABLE 2--Results of the Pearson Product-Moment Statistical Analysis of the Torque and IEMG Measurements of the Static Contraction of the Rectus Femoris Muscle at Four Different Knee Angles

	All Angles	90^o	60^o	30^o	10^o
r	-0.10105	0.09521	0.36664	0.19845	0.26012
r^2	0.01021	0.00906	0.1344	0.03938	0.06766
p	0.2721	0.6167	0.0463	0.2931	0.1651
Significance	NS	NS	Sig	NS	NS

TABLE 3--Matrix Demonstrating the Correlation (r) Between Angle and Torque, Angle and Micro-Volts, and Torque and Micro-Volts

	Angle	Torque	Micro-Volts
Angle	1.0	r = 0.779 p = .0001	r = -0.291 p = .0012
Torque		1.0	r = 0.101 p = .2721
Micro-Volts			1.0

For the four measured points (90^o, 60^o, 30^o, 10^o) through the range of motion, the r = -0.10105 and the coefficient of determination (r^2) was .01021, thus 1.02 percent of the variance in the torque was accountable from the variance in the IEMG (p = .2721). The correlation between torque and joint angle was calculated to be 0.779 (p = .0001). Sixty-one percent of the variance in the torque was accountable from the variance in the joint angle. The correlation between IEMG and joint angle was calculated to be -0.291 ($p = .0012$). Thus the null hypothesis of no relationship between torque and IEMG of the isometric contractions of the rectus femoris muscle at varying knee joint angles was accepted.

The correlation between torque and IEMG was calculated for each angle separately. At 90^{0} of knee flexion, the $r = 0.09521$ and $r^2 =$ .90 percent. At 60^{0}, the $r = 0.36664$ and $r^2 = 13.44$ percent. At 30^{0}, the $r = 0.19845$ and $r^2 = 3.94$ percent. At 10^{0}, the $r = .26012$ and $r^2 =$ 6.77 percent. Even though the correlation is statistically significant at 60^{0} ($p = 0.0463$), 86.56 percent of the variance is unrelated to changes in either variable and for practical purposes a relationship between torque and IEMG does not exist at this angle.

The mean torque, IEMG, and percent of maximum IEMG recorded during phase two from the rectus femoris muscle while contracting isometrically at five different intensities of force are presented in table 4.

TABLE 4--Descriptive Statistics for Torque and IEMG at Five Intensities of Isometric Contractions of the Rectus Femoris Muscle at 60^{o} of Knee Flexion for Thirty Subjects

% Max Torque	Torque* (ft-lbs)	S	Range	IEMG+ (μV)	S	Range	% Max* IEMG	S	Range
20%	21.77	3.45	13.20-28.60	42.60	16.80	19.00-79.00	25.69	6.41	11.67-37.62
40%	43.57	6.97	26.40-57.60	63.63	19.95	30-105	38.76	7.32	22.78-54.55
60%	65.36	10.46	39.60-86.40	96.37	29.84	42-145	58.51	10.19	36.11-77.01
80%	87.15	13.94	52.80-115.20	129.97	37.11	53-185	78.96	11.61	51.11-96.88
100%	108.93	17.43	66 - 144	167.37	48.64	55-255	100.00	0.00	100 - 100

*calculated
+measured

At 20 percent of maximum torque, the mean percent IEMG was 25.69 percent. At 40 percent of maximum torque, the mean percent IEMG was 38.76 percent. At 60 percent of maximum torque, IEMG was 58.51 percent. And at 80 percent of maximum torque, the mean percent IEMG was 78.96 percent. The 100 percent values were taken from the maximum effort at 60^{o} of knee flexion position in phase one of the experiment. These values were used to calculate the other percentages. This relationship is depicted in figure 4.

A Pearson product-moment correlation was employed for analysis of the relationship between the torque (independent variable) and the IEMG

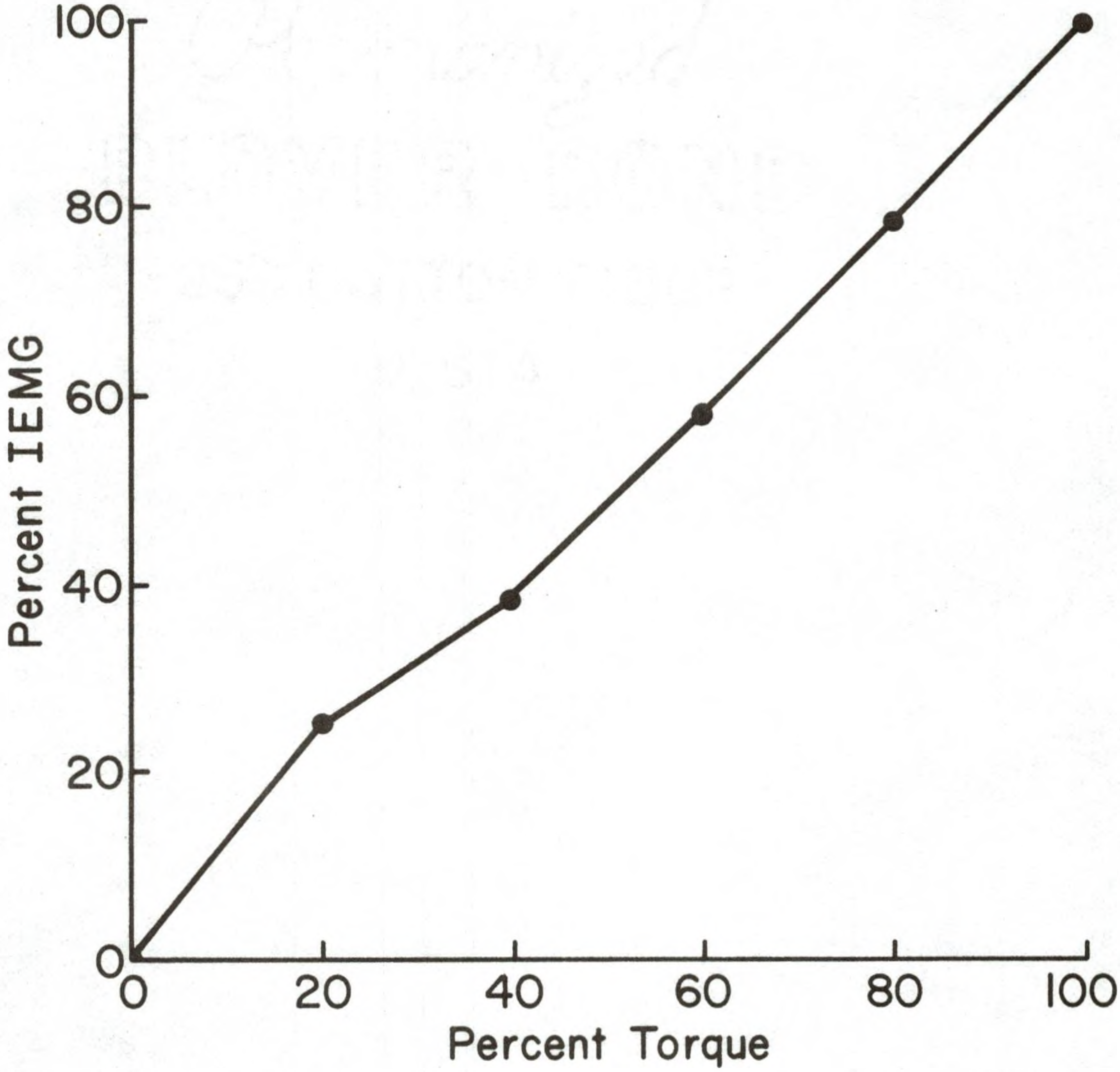

Fig. 4. Relationship of mean percent torque to mean percent IEMG at five intensities of static contractions of the rectus femoris muscle at 60^{0} of knee flexion for thirty subjects

(dependent variable) at five different intensities of isometric contraction. The results of this analysis are presented in table 5.

TABLE 5--Results of the Pearson Product-Moment Statistical Analysis of the Torque and IEMG Measurements of Five Intensities of Force of the Static Contractions of the Rectus Femoris Muscle at 60° of Knee Flexion For Thirty Subjects

	All Ranges*	20%*	40%*	60%*	80%*	100%*	All Ranges (%)+
r	0.83037	0.28524	0.30092	0.33595	0.34187	0.36664	0.95440
r^2	.6895	.0814	.0906	.1129	.1169	.1344	.9109
p	0.0001	0.1265	0.1061	0.0695	0.0644	0.0463	0.0001
Significance	Sig	NS	NS	NS	NS	Sig	Sig

*results using raw data
+results using percent torque:percent IEMG

Over the range of intensities of force (20%-100%) for the calculated torque and the measured IEMG, the r = .83037 and r^2 was .6895, thus 68.95 percent of the variance in the IEMG was accountable from the variance in the torque (p = .0001). The analysis of the relationship between percent IEMG and percent torque over the range of force intensities presented an r = .95440 and r^2 = .9109. Therefore, 91.09 percent of the variance in the percent IEMG was accountable from the variance in the percent torque (p = .0001). The null hypothesis was rejected. The alternate hypothesis states that: In the variation of the range of intensity of contraction of the rectus femoris muscle, there is a linear relationship between IEMG and torque, percentages being a stronger indicator than raw values.

The correlation between torque and IEMG was calculated for each intensity (20%, 40%, 60%, 80%, and 100%) separately. The p values demonstrate that there is no significant relationship between IEMG and torque at 20, 40, 60, and 80 percent of intensity. The p value of .05 for 100 percent intensity showed statistical significance, but because this value was taken from the measurements done in part one and not actually measured in part two of the study, it is inappropriate to place significance on this value. Therefore, knowledge of the value of one measurement at one intensity is not a good predictor of the other value at that intensity.

CHAPTER V

DISCUSSION AND CONCLUSIONS

The torque curve of the knee extensors produced by a maximal isometric contraction at 90^{o}, 60^{o}, 30^{o}, and 10^{o} of knee flexion is typical of those found by other investigators (Moritz, et al. 1973 and Williams and Stutzman 1959 and Mendler 1967 and Smidt and Rogers 1982). The torque curve peaks at 60^{o} of flexion and is lowest at complete extension. Active contractile and passive elements of force production and biomechanical factors account for these results.

In the lengthened position, the actin-myosin relationship is optimal. As a muscle shortens, the myofilaments extent of movement becomes limited resulting in relatively less tension in the muscle (Poland, Hobart and Payton 1981). Shortening the muscle decreases the efficiency of the muscle fibers.

As a muscle is lengthened, the parallel elastic components are placed on stretch. As a muscle decreases in length, the passive tension is decreasing. This property of muscle passively produces increased force with stretch.

The force a muscle can generate is equal to the sum of the active contractile and passive elements. The length-tension relationship of muscle and the elastic component are both affected in a positive direction for the rectus femoris as the angle of the knee increases. The knee joint is unique because of the patellar mechanism. At 60^{o} of flexion, the patella lifts the tendon of insertion away from the joint axis, thereby improving the leverage. This factor takes away from a direct linear relationship between joint angle and torque at the knee joint. Regardless, the two factors have a fair correlation ($r = 0.78$).

The maximum IEMG curve of the rectus femoris showed the highest electrical record at 10^{0} of knee flexion and the lowest value at 90^{0}. The contributing factors of these results are the length-tension relationship and innervation from Golgi tendon organs.

As explained earlier, the muscle is less efficient in the shortened position. This may have caused increased recruitment and increased frequency of firing of the motor units. Even though all the contractions were maximal, the subjects routinely stated that the 30^{0} and 10^{0} positions were "more difficult" than the 60^{0} and 90^{0} positions.

The Golgi tendon organs have an inhibitory effect on muscle tension in a lengthened muscle. This afferent sensory ending responds to the stretch stimulus by causing autogenic inhibition accompanied by excitation of the antagonist (Poland, et al. 1981).

While the passive property of muscle causes increased torque in the lengthened position, it has no effect on the electrical record.

Because the factors affecting torque values and IEMG values are varied and sometimes opposing, one would not expect the two values to be correlated. This study was in agreement with other studies in that there was no significant relationship between IEMG and torque at different joint angles (Lunnen, et al. 1981 and Inman, et al. 1952 and Moritz, et al. 1973 and Libet, et al. 1959).

The low correlation between IEMG and torque at each angle suggests that one value cannot be used to predict the other value. The correlation between angle and IEMG was very low and negative. IEMG is not a constant value to be used for direct comparisons between subjects, or between muscles, or even of the same muscle on another day if the electrodes have been removed. IEMG values are affected by the size of the surface electrodes, distance between electrodes, exact placement over a large or small motor unit, and skin impedance (Hall 1970 and Vredenbregt and Rau 1973). The large standard deviations of the microvolt values reflect the variability of this value between subjects who are relatively similar in size and strength.

The results indicate that when all the biomechanical and physiological factors of torque are held constant, a linear relationship exists between torque and IEMG. Three things happen as the force of contraction increases: (1) recruitment of additional motor units, (2) increased frequency of firing of motor units, and (3) summation of contraction to produce some degree of tetanus (Poland, et al. 1981). Since the EMG signal is the electrical manifestation of the neuromuscular activation associated with a contracting muscle, one could expect that as the isometric force increases, the EMG would increase at a similar rate.

The study showed the greatest increase in IEMG from 0 percent to 20 percent of maximum torque, the least increase between 20 percent to 40 percent, and a steady moderate increase from 40 percent to 100 percent. As referred to in the methodology section, there was a margin of error (approximately $\pm$1-3 ft-lbs) in producing the percentage of maximum torque by reading the dynamometer dial. This would affect the relationship at the lower end of the scale more than the upper end. It has been explained that at low force levels, recruitment proved to be the major mechanism of increasing force. Increased frequency proved to be responsible for the coarser adjustments that are made at higher force levels (Milner-Brown, Stein and Yemm 1973).

If the raw values of both factors at a given intensity of contraction were known, the raw values for all other intensities could be predicted with reasonable accuracy.

However, knowledge of either the IEMG or torque value at one intensity is not a good predictor of the other value at that intensity. Again, this is due to the lack of a constant value for IEMG.

Since the maximum force level is accompanied by less integrated EMG activity than at other positions, this raises a clinical question. Can the desired training effect be accomplished most efficiently when the muscle develops the most torque (lengthened position) or where the muscle develops the most EMG activity (shortened position)? Further research is indicated to answer this question.

This project studied muscles contracting isometrically. The effect of properties of muscle such as isotonic contraction, speed of contraction, work, and power on the relationship between IEMG and torque would be of interest to study.

In situations where energy conservation is a factor (neurological patients, respiratory patients, industrial workers), modifying the task so that the person can produce the greatest torque with the least amount of muscular activity would be beneficial. Considering the results of this study, some principles for work simplification could be established.

The Cybex II Isokinetic Unit and IEMG have both proven to be useful tools in studies of human skeletal muscle. Under certain restricted conditions, the two evaluative tools tell us relatively the same thing about muscle activity. Under changing conditions, the two evaluative tools cannot be related.

SUMMARY

Torque and IEMG measurements were recorded from the rectus femoris muscle on thirty active female subjects. The Cybex II Isokinetic Unit monitored torque while a Cyborg J33 biofeedback unit and a BL900 Processor monitored electrical activity of the exercising muscle. A maximal static exercise at 90^{o}, 60^{o}, 30^{o}, and 10^{o} of knee flexion of the rectus femoris was tested. There was no linear relationship between torque and IEMG at varying joint angles ($r = -0.10105$).

The relationship between torque and IEMG was studied with the knee in 60^{o} of flexion--keeping all biomechanical, anatomical, and physiological factors constant. The IEMG was monitored over the range of intensity of contraction (20%, 40%, 60%, 80%, and 100% of maximum torque). A linear relationship was found to exist between IEMG and torque of an isometric contraction ($r = 0.95440$).

APPENDIX

TABLE 6

ANTHROPOMETRIC DATA OF THE SUBJECTS

	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10
Age	22	20	18	21	22	23	21	27	24	21
Height (inches)	64	62.5	65.5	68	70.5	62.5	65.5	65	67	72.5
Weight (pounds)	112	130	106	137	149	115	125	125	115	157

	S11	S12	S13	S14	S15	S16	S17	S18	S19	S20
Age	23	21	22	24	27	28	20	20	20	23
Height (inches)	68	70.5	62.5	64	66	63	69	67	65.5	65
Weight (pounds)	130	139	117	115	130	116	130	127	148	130

	S21	S22	S23	S24	S25	S26	S27	S28	S29	S30
Age	21	23	21	21	21	21	19	21	21	18
Height (inches)	68	63	65	65	65	62	68	65	66	64
Weight (pounds)	125	116	127	135	115	149	135	135	130	119

TABLE 7

RAW SUBJECT DATA FOR PHASE ONE OF THE EXPERIMENT

	90^0		60^0		30^0		10^0		
Subject	ft-lbs	μV	ft-lbs	μV	ft-lbs	μV	ft-lbs	μV	Impedence (ohms)
S 1	129	145	117	165	60	240	36	240	730
S 2	129	185	102	240	55	250	36	280	1320
S 3	71	100	92	138	80	160	42	180	1000
S 4	96	138	124	175	67	190	34	190	720
S 5	116	105	120	100	69	140	48	170	800
S 6	88	135	117	180	60	170	27	197	660
S 7	96	115	102	145	58	155	28	155	1400
S 8	77	90	100	87	67	110	36	115	800
S 9	81	90	66	55	48	72	23	88	1250
S10	170	150	144	155	85	150	36	150	2900
S11	104	155	141	185	117	190	63	175	1200
S12	98	190	129	255	79	260	46	265	2900
S13	92	155	102	215	60	285	37	295	1900
S14	102	137	113	143	60	170	30	160	2600
S15	92	195	94	200	52	210	21	200	1800
S16	99	138	80	110	64	152	40	160	775
S17	100	180	105	210	60	210	30	200	1700
S18	71	145	108	165	70	195	41	190	2400
S19	124	100	110	113	61	120	28	143	4200
S20	80	170	120	215	60	155	38	165	1100
S21	115	170	143	218	90	310	48	310	1200
S22	103	145	95	180	54	185	37	180	3500
S23	122	180	110	185	58	157	27	170	2200
S24	95	190	114	160	66	160	38	195	730
S25	81	175	90	217	55	220	25	195	830
S26	92	96	102	105	62	130	35	135	1200
S27	76	155	115	205	78	165	46	165	650
S28	110	218	113	190	67	215	37	220	1200
S29	117	105	107	110	59	100	31	110	4000
S30	80	197	93	200	53	190	24	200	830

TABLE 8

RAW SUBJECT DATA FOR PHASE TWO OF THE EXPERIMENT

		20%			40%			60%			80%			100%	
Subject	Torque	µV	%EMG	Torque	µV	%EMG	Torque	µV	%EMG	Torque	µV	%EMG	Torque	µV	%EMG
S 1	23.4	40	24.24	46.8	69	41.82	70.2	110	66.67	93.6	140	84.84	117	165	100
S 2	20.4	59	24.58	40.8	105	43.75	61.2	135	56.25	81.6	185	77.08	102	240	100
S 3	18.4	40	28.99	36.8	56	40.58	55.2	82	59.42	73.6	97	70.29	92	138	100
S 4	24.8	44	25.14	49.6	77	44.00	74.4	105	60.00	99.2	135	77.14	124	175	100
S 5	24.0	31	31.00	48.0	45	45.00	72.0	70	70.00	96.0	88	88.00	120	100	100
S 6	23.4	30	16.67	46.8	59	32.78	70.2	78	43.33	93.6	99	55.00	117	180	100
S 7	20.4	37	25.52	40.8	60	41.38	61.2	93	64.14	81.6	130	89.66	102	145	100
S 8	20.0	31	35.63	40.0	46	52.87	60.0	67	77.01	80.0	80	92.00	100	87	100
S 9	13.2	19	34.55	26.4	30	54.55	39.6	42	76.36	52.8	53	96.36	66	55	100
S10	28.2	28	18.06	57.6	48	30.97	86.4	72	46.45	115.2	99	63.87	144	155	100
S11	28.2	49	26.49	56.4	71	38.38	84.6	115	62.16	112.8	145	78.38	141	185	100
S12	25.8	65	25.49	51.6	90	35.29	77.4	130	50.98	103.2	185	72.55	129	255	100
S13	20.4	48	22.33	40.8	70	32.56	61.2	115	53.49	81.6	160	74.42	102	215	100
S14	22.6	28	19.58	45.2	59	41.26	67.8	86	60.14	90.4	135	94.41	113	143	100
S15	18.8	51	25.50	37.6	70	35.00	56.4	120	60.00	75.2	175	87.50	94	200	100
S16	16.0	27	24.55	32.0	49	44.55	48.0	61	55.45	64.0	89	80.91	80	110	100
S17	21.0	79	37.62	42.0	105	50.00	63.0	142	67.62	84.0	177	84.29	105	210	100
S18	21.6	37	22.42	43.2	52	31.52	64.8	90	54.55	86.4	137	83.03	108	165	100
S19	22.0	32	28.32	44.0	40	35.40	66.0	63	55.75	88.0	105	92.92	110	113	100
S20	24.0	39	18.14	48.0	62	28.84	72.0	88	40.93	96.0	142	66.05	120	215	100
S21	28.6	78	35.78	57.2	92	42.20	85.8	145	66.51	114.4	185	84.86	143	218	100
S22	19.0	21	11.67	38.0	41	22.78	57.0	65	36.11	76.0	92	51.11	95	180	100
S23	22.0	34	18.38	44.0	58	31.35	66.0	87	47.03	88.0	120	64.86	110	185	100
S24	22.8	43	26.88	45.6	68	42.50	68.4	105	65.63	91.2	155	96.88	114	160	100
S25	18.0	47	21.66	36.0	65	29.95	54.0	105	48.39	72.0	155	71.43	90	217	100
S26	20.4	22	20.95	40.8	36	34.29	61.2	59	56.19	81.6	72	68.57	102	105	100
S27	23.0	61	29.76	46.0	80	39.02	69.0	142	69.27	92.0	160	78.05	115	205	100
S28	22.6	61	32.11	45.2	80	42.11	67.8	127	66.84	90.4	150	78.95	113	190	100
S29	21.4	25	22.73	42.8	37	33.64	64.2	55	50.00	85.6	94	85.45	107	110	100
S30	18.6	72	36.00	37.2	89	44.50	55.8	137	68.50	74.4	160	80.00	93	200	100

BIBLIOGRAPHY

BIBLIOGRAPHY

Basmajian, J. V. Muscles Alive. 4th ed. Baltimore: Williams & Wilkins Co., 1978, pp. 5-93.

Bouisset, S. "EMG and Muscle Force in Normal Motor Activities." In New Developments in Electromyography and Clinical Neurophysiology, Vol. 1, pp. 547-583. Edited by J. E. Desmedt. New York: S. Karger Co., 1973.

Bouisset, S., and Maton, B. "Quantitative Relationship Between Surface EMG and Intramuscular Electromyographic Activity in Voluntary Movement." American Journal of Physical Medicine 51 (December 1972): 285-295.

Brunnstrom, S. Clinical Kinesiology. 3rd ed. Philadelphia: F. A. Davis Co., 1972, pp. 35-54.

Edwards, R. G., and Lippold, O. C. J. "The Relation Between Force and Integrated Electrical Activity in Fatigued Muscle." Journal of Physiology 132 (June 1956): 677-681.

Hall, E. A. "Electromyographic Techniques: A Review." Physical Therapy 50 (May 1970): 651-659.

Inman, V. T.; Ralston, H. J.; Saunders, J. B. De C. M.; Feinstein, B.; and Wright, E. W. "Relation of Human Electromyogram to Muscular Tension." Electroencephalography and Clinical Neurophysiology 4 (May 1952): 187-194.

Isolated Joint Testing and Exercise. Bay Shore, NY: Lumex, Inc., 1980.

Libet, B.; Feinstein, B.; and Wright, E. W. "Tendon Afferents in Autogenetic Inhibition in Man." Electroencephalography and Clinical Neurophysiology 11 (February 1959): 129-140.

Lippold, O. C. J. "The Relation Between Integrated Action Potentials in a Human Muscle and Its Isometric Tension." Journal of Physiology 117 (August 1952): 492-499.

Lunnen, J. D.; Yack, J.; and LeVeau, B. F. "Relationship Between Muscle Length, Muscle Activity, and Torque of the Hamstring Muscles." Physical Therapy 61 (February 1961): 190-195.

McLeod, W. D. "EMG Instrumentation in Biomechanical Studies: Amplifiers, Recorders, and Integrators." In New Developments in Electromyography and Clinical Neurophysiology, Vol. 1, pp. 511-518. Edited by J. E. Desmedt. New York: S. Karger Co., 1973.

Mathews, D. K., and Fox, E. L. The Physiological Basis of Physical Education and Athletics. 2nd ed. Philadelphia: Saunders College Publishing, 1976, pp. 122-133.

Mendler, H. M. "Knee Extensor and Flexor Force Following Injury." Physical Therapy 47 (January 1967): 35-45.

Milner-Brown, H.S.; Stein, R. B.; and Yemm, R. "Changes in Firing Rate of Human Motor Units During Linearly Changing Voluntary Contractions." Journal of Physiology 230 (April 1973): 371-390.

Moritani, T., and deVries, H. "Reexamination of the Relationship Between the Surface Integrated Electromyogram (IEMG) and Force of Isometric Contraction." American Journal of Physical Medicine 57 (December 1978): 263-277.

Moritz, U.; Svantesson, G.; and Haffajee, D. "A Biomechanical Study of Muscle Torque as Affected by Motor Unit Activity, Length-Tension Relationship, and Muscle Force Lever." In New Developments in Electromyography and Clinical Neurophysiology, Vol. 1, pp. 675-682. Edited by J. E. Desmedt. New York: S. Karger Co., 1973.

Poland, J. L.; Hobart, D. J.; and Payton, O. D. The Musculoskeletal System. 2nd ed. Bern, Switzerland: Hans Huber Publishers, 1981.

Ralston, H. J. "Uses and Limitations of Electromyography in the Quantitative Study of Skeletal Muscle Function." American Journal of Orthodontics 47 (July 1961): 521-530.

SAS Release 79.5 at U.N.D. Cary, N.C.: SAS Institute Inc., November 2, 1982.

Smidt, G. L., and Rogers, M. W. "Factors Contributing to the Regulation and Clinical Assessment of Muscular Strength." Physical Therapy 62 (September 1982): 1283-1290.

Vredenbregt, J., and Rau, G. "Surface Electromyography in Relation to Force, Muscle Length, and Endurance." In New Developments in Electromyography and Clinical Neurophysiology, Vol. 1, pp. 607-622. Edited by J. E. Desmedt. New York: S. Karger Co., 1973.

Williams, M., and Stutzman, L. "Strength Variation Through the Range of Joint Motion." Physical Therapy Review 39 (March 1959): 145-152.