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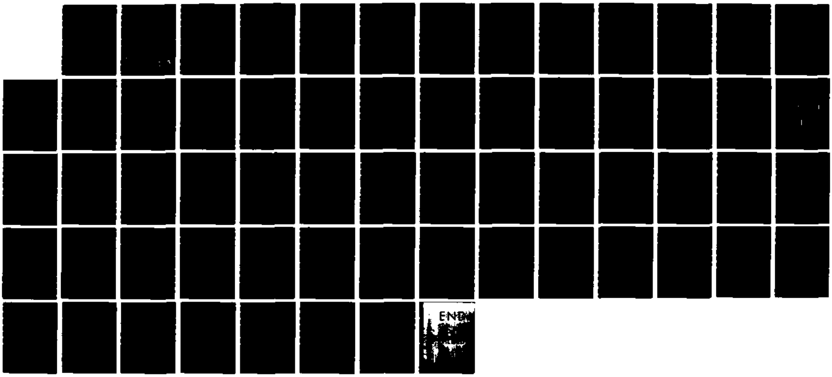

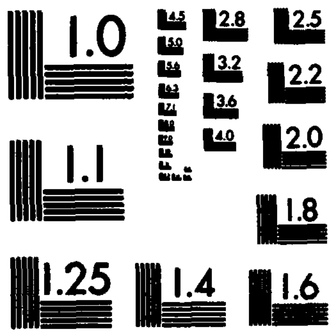

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ELECTROMYOGRAPHIC ANALYSIS OF THE PERONEOUS LONGUS DURING BICYCLE ERGOMETRY ACROSS WORK LOAD AND PEDAL TYPE

by

Danny Lee Holt

A Thesis submitted to the Faculty of The University of North Carolina at Chapel Hill in partial fulfillment of the requirements for the degree of Master of Science in the department of Medical Allied Health Professions in the Division of Physical Therapy.

Chapel Hill, 1983

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ABSTRACT

Title: Electromyographic Analysis of the Peroneous Longus During Bicycle Ergometry Across Work Load and Pedal Type

Investigator: Danny L. Holt, Captain, USAF

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Lateral ankle injuries often result in residual disability. Increasing the endurance of the peroneous longus may reduce this problem. Bicycle ergometry may increase the endurance of the peroneous longus, but the activity of the peroneous longus during pedaling is not known. The purpose of this study was to analyze the electromyographic (EMG) activity of the peroneous longus across work load (1, 2, and 3 Kp) and pedal type (standard and medial support only) during pedaling. The analysis included total, peak, and phasic EMG activity per cycle. EMG activity was monitored with bipolar surface electrodes arranged in a longitudinal configuration over the motor points. The pedaling cycle was monitored with a photo electric cell and the gait cycle was monitored with a heel switch. Gait data were used to normalize the pedaling data.

Data from nineteen muscles were collected from eleven subjects, all adult males. Two factor ANOVA's with repeated measures over both factors were used to test for significance as each subject was tested in all six pedaling conditions. When significance was found, the source was located with a protected least significant difference test. Results indicated that: 1) increasing work load significantly increases the total, peak, and phasic EMG activity; 2) modified pedals significantly increase the total, peak, and phasic EMG activity; 3) the interaction between work load and pedal type significantly increases the total and peak EMG activity; and 4) only pedaling with modified pedals results in EMG activity comparable to gait.

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Chapter I

INTRODUCTION

Ankle injuries are one of the most common problems encountered in almost all sports and activities requiring locomotion over land. A two year study of high school athletes revealed that 14% of all injuries involved the ankle, and 85% of these injuries were sprains and strains of the lateral ankle ligaments and musculature.[19] This supported an earlier study by Thorndike. Thorndike documented the body part involved in injuries incurred in sports by students at Harvard University from 1932 to, 1954. He found that 12% of all injuries recorded involved the ankle.[45]

Many ankle injuries result in residual disability and repeated injuries. Bosien and co-workers reported that 40% of the ankle injuries in their study became chronic problems with recurring injuries.[7] They also reported that the only statistically significant factor shared by the cases of residual disability was peroneal muscle weakness. Many other factors thought to cause repeated ankle injuries have been explored. These include varus instability of the talus in the ankle mortise,[14] anterior-posterior instability of the talus in the ankle mortise,[1] instability at the subtalar joint,[7] inferior tibio-fibular disastasis,[36] and a "weak spot" in the lateral ligaments.[45] Proprioceptive deficit has emerged most recently as a possible cause of recurring ankle injuries.[17,18] However, the importance of peroneal muscle strength in the rehabilitation of ankle injuries and the prevention of residual disability continues to be emphasized. Commenting on his own personal experience in this regard, Bostrom wrote, "graduated exercise to strengthen the limb, particularly the peroneous musculature, often produced improvement which rendered operation unnecessary."[9]

Rehabilitation programs for ankle injuries described in the literature usually include peroneal muscle strengthening exercises.[20,33,48] These programs, however, are often not well described and do not include many endurance exercises for the peroneal musculature until after weight bearing has begun. Although bicycle ergometry may be specific for the peroneal muscles in terms of endurance because of its similarity to gait, it is not included in these rehabilitation programs.[2] The absence of bicycle ergometry from traditional rehabilitation programs may be explained by the small number of studies that have reported the activity of the lower extremity musculature during pedaling. In addition, one of the most important muscle groups involved in ankle injuries, the peroneals, have not been included in the few studies that have been done.

The purpose of this study was to analyze the electrical activity of the peroneous longus during bicycle ergometry across work load and pedal type. This analysis included the total electromyographic (EMG) activity per cycle, the peak EMG activity per cycle, and the phasic EMG activity per cycle. An additional purpose of this study was to compare the total and peak EMG activity per cycle of the peroneous longus during pedaling with that found during gait.

Hypotheses

Nine specific null hypotheses were tested by this study. The first three are the primary hypotheses and concern the effect of work load and pedal type on the total EMG activity per cycle of the peroneous longus during bicycle ergometry. The last six are secondary hypotheses concerning the effect of work load and pedal type on the peak and phasic EMG activity per cycle of the peroneous longus during bicycle ergometry. They were tested to help explain the acceptance or rejection of the primary hypotheses. All nine hypotheses are listed below:

1. Ho: I1=I2=I3. There is no significant difference in total EMG per cycle because of the main effect of work load, controlling for pedal type effects.

2. Ho: Is=Im. There is no significant difference in the total EMG per cycle because of the main effect of pedal type, controlling for work load effects.

3. Ho: (I1m-I1s)=(I2m-I2s)=(I3m-I3s). There is no significant interaction effect between pedals and work loads in terms of total EMG per cycle.

4. Ho: A1=A2=A3. There is no significant difference in peak EMG amplitude per cycle because of the main effect of work load, controlling for pedal type effects.

5. Ho: As=Am. There is no significant difference in the peak EMG amplitude per cycle because of the main effect of pedal type, controlling for work load effects.

6. Ho: (A1m-A1s)=(A2m-A2s)=(A3m-A3s). There is no significant

interaction effect between pedals and work loads in terms of peak EMG amplitude per cycle.

7. Ho: P1=P2=P3. There is no significant difference in the percentage of the pedal cycle during which the peroneous longus was active because of the main effect of work load, controlling for pedal effects.

8. Ho: Ps=Pm. There is no significant difference in the percentage of the pedal cycle during which the peroneous longus was active because of the main effect of pedal type, controlling for work load effects.

9. Ho: (P1m-P1s)=(P2m-P2s)=(P3m-P3s). There is no significant interaction effect between pedals and work loads in terms of the percentage of the pedal cycle during which the peroneous longus is active.

In addition to these nine hypotheses, two other hypotheses concerning the relative intensity of peroneous longus muscle activity during pedaling were tested by this study:

1. Total and peak EMG activity of the peroneous longus muscle during pedaling with standard pedals is less than that found during gait.

2. Total and peak EMG activity of the peroneous longus muscle during pedaling with modified pedals is equal to or greater than that found during gait.

Limitations

This study was limited by the small number of subjects (11), their sex (all male), and their age (all young adults). A further limitation of this study was the nature of the population from which the subjects were drawn. All the subjects came from a collegiate population. This special population may differ from the general public in terms of overall health and fitness. Finally, the assumption is made that treadmill gait is no different than over ground walking in terms of peroneous longus function. A pilot study by this investigator found little difference in peroneal function when the subject had experience with walking on a treadmill. In addition, the results of a study by Walmsley of the phasic activity of the peroneous longus during gait on a treadmill were similar to those from studies that used over ground walking.[49,41,44]

Definitions

The following terms and their definitions are important to the understanding of this study:

1. Electromyography (EMG): The detection, recording, and study of electrical potentials produced by muscle.

2. Integrated EMG (IEMG): A method of processing the EMG signal. This technique allows the determination of the total amount of activity generated by a muscle.

3. Linear envelope (RC): A method of processing the EMG signal. This technique allows the determination of the instantaneous amplitude

of the activity generated by a muscle. It can be used to find the peak activity of a muscle.

4. Metabolic specificity: This refers to the particular physiological process employed by a muscle to produce the necessary energy for contraction. The particular process employed depends on the duration and intensity of the work performed by the muscle.

5. EMG Normalization: The mathematical process of converting the absolute EMG data collected during a testing condition to a percentage of the absolute EMG data collected from a standard contraction or movement using the same electrode setup. This technique lessens the influence of inconsistencies in electrode setup and allows comparisons to be made more accurately.

Chapter II

LITERATURE REVIEW

Several important concepts are presented in this literature review to provide the background and foundation on which this study is built. In addition, this review supports the methods used in this study by presenting methods used in similar studies.

The principle of metabolic specificity is well documented in the literature.[15,10,46,23] However, the effective application of this principle requires a knowledge of the normal function of the muscle being rehabilitated. For this reason, and to establish the need for an endurance exercise for the peroneous longus during rehabilitation, studies investigating the normal activity of the peroneous longus muscle are reviewed initially. Following this, EMG studies that have observed the lower leg muscles during bicycle ergometry are presented with emphasis on their methods and conclusions. Finally, the present status of EMG is described in terms of the relationship between the EMG signal of a muscle and the force or tension developed by that muscle. The variables that affect this relationship are also discussed.

Normal Function of the Peroneous Longus

The activity of the peroneous longus has been studied by many investigators. The primary tool of these investigators has been EMG. Needle stylet, fine wire, and surface electrodes have been among the types of electrodes used to monitor the peroneous longus muscle during gait.[41,22,49] Walmsley even contrasted the data collected using fine wire electrodes with that collected using surface electrodes and reported essentially no difference.[49]

The placement of the electrodes has been based either on anatomical landmarks or on direct current stimulation of the muscle motor points.[41,49] The exact location of the electrodes has included the area directly over the motor point and the area over the middle of the muscle belly.[41,22] Sometimes the exact location has not been reported.[49,44]

Over ground gait and treadmill gait have both been investigated as well as the effect of "flat feet" on the activity of the peroneous longus muscle during gait.[41,49,22] Gait cycle has been monitored with the use of heel switches or with cinematography.[41,44] Several of the investigators also reported using a standard or "control" contraction to normalize their data. They indicated that this procedure reduced the effects of inconsistencies in electrode setup and allowed more accurate comparisons to be made between subjects.[13,49]

Although the methods used in past studies have varied, their results are similar. All investigators reported peroneous longus muscle activity beginning shortly after heel strike, peaking during heel off, and ceasing just before toe off.[41,44] This was reported for both over ground and treadmill gait.[49] In addition, Gray and Basmajian indicated that "flat feet" caused slightly more activity in the peroneous longus muscle than "normal feet."[22] Based on the results of other muscles of the lower leg, the activity of the peroneous longus during gait is probably less than the maximum activity possible but increases with activities such as running or jumping.[13]

An article by Scheller, Kasser, and Quigley, in 1980, considered the biomechanical aspects of tendon injuries about the ankle.[39] The peroneous longus and posterior tibialis muscles were credited with providing 15% of the plantarflexion force of the ankle. Both muscles were also said to be important in dynamic stability of the ankle, leading the authors to conclude that peroneal rehabilitation after ankle injury is vitally important in the prevention of recurring injuries.

In summary, the normal activity of the peroneous longus muscle during gait is cyclic in nature and provides some plantarflexion force as well as lateral dynamic stability during the stance phase of gait. The intensity of this activity is probably less than the maximum activity possible from the peroneous longus muscle. The normal function of the peroneous longus during gait is clearly one demanding endurance more than strength.

EMG Studies of Bicycle Ergometry

Human locomotion and pedaling are very similar. Basmajian compared the two and indicated that human locomotion is efficient because of its similarity to pedaling.[2] He described each foot as a short segment of an invisible wheel resulting in two offset wheels slightly separated for stability. Because of this similarity, the stationary bicycle has been included in many rehabilitation programs for the lower extremities. As early as 1939 the stationary bicycle was indicated in the literature as a means of increasing range of motion and strength of the lower extremities.[4,5,12,42] However, the value of the stationary bicycle has been based primarily on empiricism because of the relatively few studies that have investigated the activity of the lower leg muscles during pedaling.

One of the first studies to investigate pedaling was performed in 1959 by Houtz and Fischer.[24] Three healthy adult women served as subjects. The phasic activity of fourteen muscles of the lower extremity was monitored with EMG. These muscles consisted of flexors and extensors of the hip, knee, and ankle, but did not include the peroneous longus. The subjects were tested at two seat heights and six work loads. Two speeds were also tested, but the exact speeds were not given. Surface electrodes were placed 2.4 centimeters apart, equidistant from the motor point, and parallel with the muscle fibers. The motor points were located with a small direct current stimulator.

The results indicated that increasing work load caused more muscles to become active and caused the activity of any single muscle to increase. Muscle activity was described simply as minimal, moderate, or maximal. No change in the basic timing of the muscular activity occurred with increasing work load. Increasing speed also increased the number of muscles participating and the total activity, but not to the same degree as increasing work load. The authors concluded that pedaling resulted in a pattern of muscle activity that was orderly and coordinated and that increasing load or speed increased the number of muscles active and the total activity present. However, the timing of the muscles' phasic activity was not changed.

Goto and co-workers used bicycle ergometry with three male subjects to study the relationship between IEMG and both work load and speed.[21] The gluteus maximus, vastus lateralis, gastrocnemius, and anterior tibialis muscles were monitored with bipolar surface electrodes during pedaling. The placement of the electrodes was not described. Loads of 0, 1, 2, and 3 kilograms and rates of 40, 60, 80, and 100 revolutions per minute (rpm) were assigned to different testing conditions. Rest periods were given to eliminate any fatigue effects.

A positive linear relationship between IEMG and rate was reported for the vastus lateralis and gastrocnemius muscles with load held constant. However, for the tibialis anterior and gluteus maximus muscles this relationship was curvilinear. The relationship between IEMG and work load was positive and linear for the vastus lateralis and gastrocnemius, but curvilinear for the gluteus maximus and anterior tibialis. The other muscles displayed changing relationships between IEMG and work load with changing speeds also.

One of the most recent studies on pedaling was performed by Mohr and co-workers in 1981.[34] Six males between 28 and 47 years of age served as subjects. Speeds of 40, 60, 80, and 100 rmp were tested, but the data from the tests made at 40 and 100 rpm were discarded because of poor speed control by the subjects. Work loads of 0, 300, 600, and 1200 kilograms per minute were tested. However, work loads of 0 and 300 kilograms did not elicit enough activity for analysis and were discarded. Six muscles were studied including a flexor and extensor of the hip, knee, and ankle, however, the peroneous longus was not included. Bipolar surface electrodes were placed one centimeter apart, equidistant from the motor point, and parallel with the muscle fibers. Motor points were located with a direct current stimulator. The subjects pedaled thirty seconds at each test condition. An EMG recording was made during the last five seconds of this period. The muscles were considered active when their EMG signal rose above the baseline.

EMG amplitude was reported as increasing with increasing work load in all muscles. The nature of this positive relationship was not described. The effect of increasing speed was quite variable. The gastrocnemius was reported as active from the top of the cycle to 225 degrees through the cycle. This timing was not affected by load or speed.

All of these studies reported coordinated, rhythmic activity of the lower leg muscles during pedaling. Both load and speed have been reported to effect the intensity of the muscle activity but not the timing. Obviously, bicycle ergometry could be used to provide endurance training for the muscles of the lower extremity which are active during pedaling. Unfortunately, the peroneous longus has never been studied during pedaling, and its activity remains unknown.

EMG/Force Relationship

In order to accurately interpret the EMG signal in terms of muscle force, an understanding of the relationship between EMG and force and the factors that influence that relationship is necessary. This relationship has been studied during both isometric and nonisometric tasks.

In the case of isometric tasks, Lippold reported a linear relationship between IEMG and tension developed in the gastrocnemius and soleus muscles during plantarflexion of the foot.[30] Other investigators have also reported a linear relationship.[11,23,40] However, curvilinear relationships have also been reported by Komi and Buskirk[27] as well as Kuroda and co-workers.[28]

The discrepancies in the results of these studies have been blamed on several factors. Moritani and DeVries reported a linear relationship between IEMG and force in elbow flexion with monopolar electrodes but a curvilinear relationship with bipolar electrodes.[35] Vigreux and associates reported stronger EMG signals with longitudinal placement of the electrodes than with transverse placement.[47] Lippold also observed that the relationship between EMG and tension varied considerably unless precautions were made to reproduce the same electrode placement with all subjects.[30] He also reported much higher reproducibility with surface electrodes than with wire electrodes. These reports indicate that electrode configuration, placement, and type must be kept constant to reduce possible variations in data resulting from electrode differences. The process of normalization has been cited by several investigators as also contributing to less electrode variation effect.[50,51] Generally, the EMG of a maximum isometric contraction has been used as the normalizing factor.[38] The configuration, placement, and type of electrodes, and the type of normalizing contraction should be reported in all studies.

Joint angle and the resulting muscle length and moment arm have also been cited as influencing the relationship between EMG and force. Liberson and associates reported a drop in muscle activity as a muscle shortens.[29] However, Inman and associates reported an increase in

EMG activity in the shortened muscle.[25] Lunnen and co-workers also reported increasing EMG and decreasing torque as the muscle shortened. The opposite was reported as the muscle lengthened.[31] This variable should be controlled by keeping the joint angle and resulting muscle length and moment arm constant for all subjects.

Fatigue has been shown to affect the EMG/force relationship. Maton reported an increase in EMG with the onset of fatigue while force remained constant.[32] Basmajian also cited many authors who have found that fatigue affects the EMG/force relationship.[3] In order to control this variable, subjects should be given adequate rest periods between testing sessions.

Although a relationship between EMG and force exists during isometric tasks, this relationship becomes much more complicated during nonisometric tasks. All the previous factors discussed for isometric tasks still apply plus several others. Velocity of movement has been shown by Bigland and Lippold to affect the EMG activity regardless of the force generated.[6] They reported a linear relationship between EMG and force during nonisometric tasks when velocity was controlled. When tension was constant, they reported a direct relationship between the velocity of movement and the level of EMG activity.

Bigland and Lippold also investigated the EMG/force relationship during an eccentric contraction. They found a positive linear relationship during eccentric contractions but with a smaller slope than the relationship found during concentric contractions. In addition, they found that at constant tension, the EMG activity of a concentric contraction was linearly related to velocity while that of an eccentric contraction was almost independent of velocity.[6] Komi demonstrated that the level of EMG activity is lower for eccentric contractions than for concentric contractions when tension is constant.[27]

Obviously, any EMG study performed on nonisometric tasks must strive to control these factors. Even with extreme care, any statements made about the force generated by a muscle based on its EMG activity must be very limited in scope.

Summary:

The principle of metabolic specificity dictates that a rehabilitation program must consider a muscle's normal function and the energy production system most utilized to perform that function. The peroneous longus is primarily a dynamic stabilizer of the lateral ankle during locomotion and must perform this function for extended periods of time. Endurance is essential, and a rehabilitation program for the peroneous longus should include endurance activities.

Bicycle ergometry provides an endurance activity for all muscles of the lower extremity which are active during pedaling. However, the activity of the peroneous longus during pedaling is not known. Because the standard pedal on stationary bicycles is stable in the inversion/eversion plane, very little activity may be required from the peroneous longus during pedaling.

Chapter III

METHOD

Introduction

This study was experimental in nature and included two independent variables, work load and pedal type, and one dependent variable, EMG activity. Three work loads and two pedal types were tested resulting in six different testing conditions. In each testing condition three aspects of the independent variable, EMG, were observed. These included total EMG activity per cycle, peak EMG activity per cycle, and phasic EMG activity per cycle. Each subject was tested in all six conditions. The functional activity of walking was used to normalize all test data and a maximum isokinetic eversion contraction was used to determine relative intensity levels.

Subjects

Eleven male subjects between the ages of 21 and 34 were recruited from the graduate and undergraduate population of the University of North Carolina at Chapel Hill. The subjects were all free of any current pathology of the lower extremities, and none of them had ever been diagnosed as having "flat feet" or "fallen arches." Four of the subjects reported having moderate to extensive experience walking on a treadmill, while seven reported little or no experience. All of the subjects stated that they participated in some form of exercise at least two to three times per week, and all reported having some experience with pedaling. However, none of the subjects were competitive cyclists. The age, height, weight, and ankle eversion strength of each subject are presented in table 1. Both right and left peroneous longus muscles were tested in eight of the subjects. The remaining three subjects only had one of their peroneous longus muscles tested because of time or equipment constraints.

Instrumentation

EMG activity of the peroneous longus muscle was recorded with Beckman* 8mm, recessed cup, silver-silver chloride surface electrodes. Boisset and Maton have shown that surface electrodes provide a valid representation of superficial muscles when compared with fine wire or needle electrodes.[8] In addition, Walmsley found, specifically, that the peroneous longus was reliably monitored using surface electrodes as compared to fine wire electrodes.[49]

The raw EMG signal was amplified and processed by equipment designed and constructed by the Biomedical Engineering Department of the University of North Carolina School of Medicine. The frequency response of the amplifier was 10 to 10,000 Hz. The raw EMG signal was first amplified and then rectified. The result was then processed by two separate means: (1) the level reset integration (IEMG) and (2) a

* Beckman Instruments Inc., Schillert Park, Illinois

TABLE 1

INDIVIDUAL SUBJECT DATA

SUBJECT NUMBER	AGE[a]	HEIGHT[b]	WEIGHT[c]	STRENGTH[d] RIGHT	LEFT
1	21	5'8"	160	10.4	10.1
2	30	5'11 1/2"	140	e	9.8
3	27	5'11"	176	12.0	12.5
4	31	5'7 1/2"	145	8.7	7.3
5	24	5'10 1/2 "	169	11.6	9.1
6	22	6'7"	185	14.0	12.5
7	34	5'7"	157	e	8.5
8	33	6'0"	150	12.3	7.0
9	22	6'2"	175	12.0	10.5
10	25	5'8"	158	8.5	7.5
11	23	6'0"	175	9.0	e

a = age in years; b = height in feet and inches; c = weight in pounds; d = strength in foot-pounds; e = not included in study linear envelope (RC). The raw, IEMG, and RC signals were recorded on a model 906 C Honeywell Visicorder† oscillograph at a paper speed of 50 mm per second.

A Collins Treadmill‡ was used during the gait portion of the study. The treadmill speed was adjusted for each subject to 60 cycles per minute using a metronome. Gait cycle was determined using an electrical pressure switch attached to the subjects' heels. The output of the heel switch was recorded on the oscillograph.

A Schwinn Ergometric Ex-2** bicycle ergometer with electrically generated resistance was used for the pedaling portion of the study. The standard pedals on the ergometer were equipped with toe straps. Pedaling rate was determined by the tachometer attached to the ergometer. Pedal position was monitored using a photoelectric cell activated when the pedals were in the full down (180 degree) position by signature cards attached to each pedal. A light beam was supplied by a special "needle hole" slide inserted in a slide projector. Because each signature card was slightly different, the light beam was interrupted in a unique way by each card. As a result, pedal position was easily determined. The output of the photoelectric cell was recorded on the oscillograph along with the heel switch and EMG signals.

A specially built medial support platform was used to modify the

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† Honeywell Test Instrument Division, Denver, Colorado

‡ Warren E. Collins, Inc., 220 Wood Road, Brain Tree, Massachusetts

**Excelsior Fitness Equipment Co., 613 Academy Drive, Northbrook, Illinois standard pedals. This platform was one inch high, one and a quarter inches wide, and three and a half inches long. The platform laid across the standard pedal against the inside edge providing support only for the head of the first metatarsal (figure 1). Toe straps were also included on the modified pedals.

Strength measurements were provided by a force dynamometer incorporated within a Cybex II†† system. This system utilized the eversion/inversion apparatus for testing the maximum eversion strength of the subjects. A twelve inch platform was used to support the subject's lower leg during the strength tests.

Procedure

The proposal for this study and the human subject's consent form were previously reviewed and approved by the Committee on the Protection of the Rights of Human Subjects of the University of North Carolina at Chapel Hill. Before testing, the subject read and signed the human subject's consent form (appendix B). The subject was then interviewed, and information was collected concerning age, height, weight, fitness level, treadmill experience, and pedaling experience. The procedure was then reviewed with the subject, and the subject was given several minutes to become acquainted with the equipment. After the period of familiarization, surface electrodes were applied to either the right or left peroneous longus muscle. The determination of which muscle to test first was made in random fashion. The motor

†† Lumex Inc., Bay Shore, New York

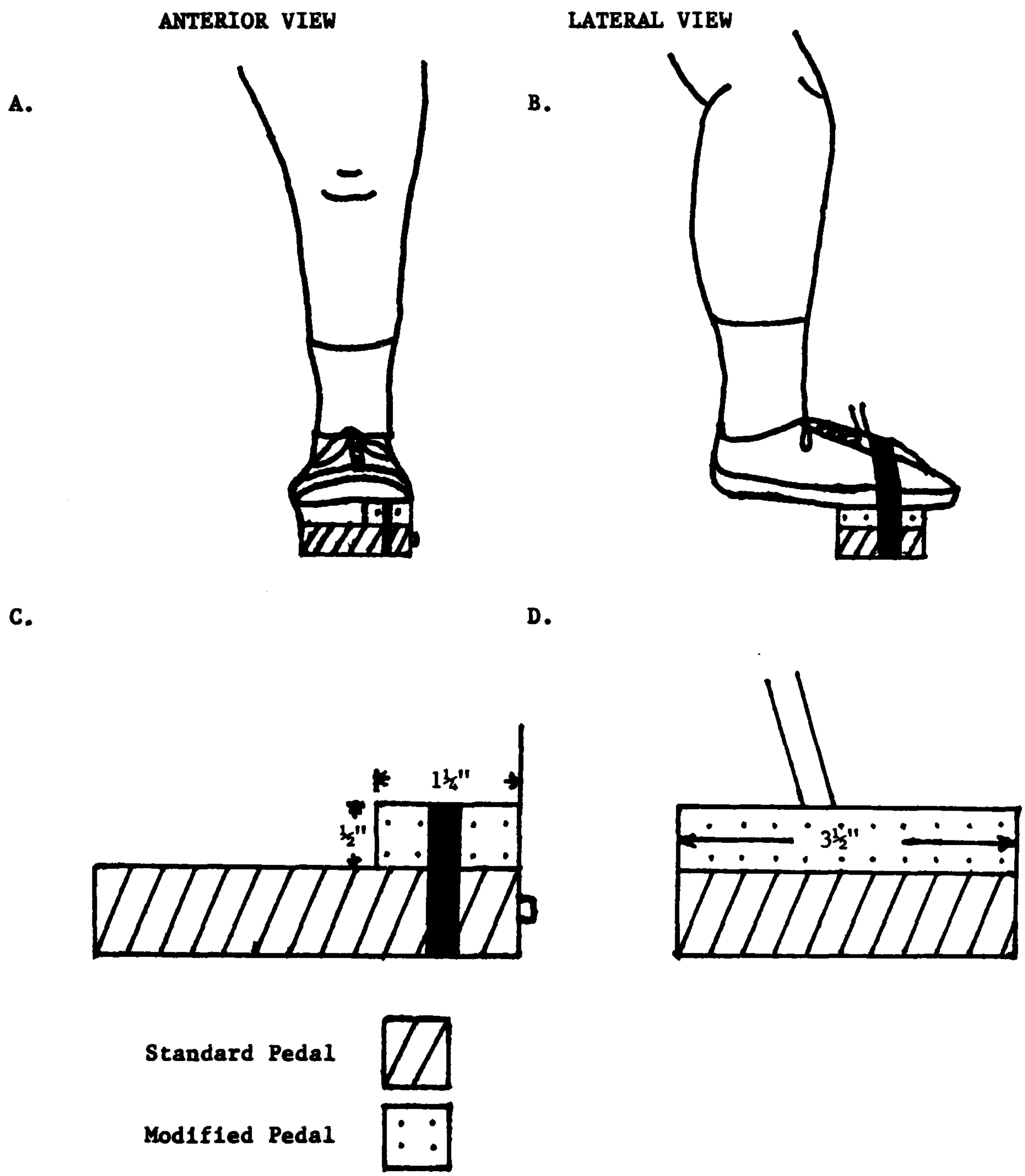

Figure 1: Modified Pedal. A = anterior view of subjects foot on modified pedal; B = lateral view of subjects foot on modified pedal; C = enlarged anterior view of the modified pedal strapped to the standard pedal; D = enlarged lateral view of the modified pedal strapped to the standard pedal

point was located by palpation and direct current stimulation. In all cases, the motor point was found on the lateral proximal leg just below the head of the fibula. A bipolar longitudinal configuration of electrodes was used on all the subjects with one active electrode over the motor point and the other active electrode one and a half centimeters distal to the motor point, and parallel with the muscle fibers. The ground electrode was placed posterior to the two active electrodes forming an equilateral triangle (figure 2). This electrode configuration provided an intrasubject reliability coefficient of 0.96 in pilot studies performed by this investigator. Skin resistance was lowered to 20,000 ohms by shaving the electrode sites and cleaning each site with alcohol. When necessary, the sites were slightly abraided with a dental burr. Electrodes were filled with conducting gel and held in place with double adhesive rings.

An electrical pressure switch was taped to the subject's heel on the same side as the electrodes. The wires from the electrodes and heel switch ran up the subject's leg to his waist where a small pre-amp amplified the EMG signal. The amplified EMG signal was carried to the main processing equipment by a flexible multiwire cable. The signal from the pressure switch was carried on a separate wire to the recording equipment. All wires were held tightly against the subject's leg to reduce movement artifact.

The subject then walked on the treadmill at a rate of sixty cycles per minute. A metronome was used to set and maintain the proper pace. As the subject walked, the EMG equipment was adjusted to provide proper amplification for accurate analysis. After approximately one minute, a ten second recording was made of both the EMG and pressure switch

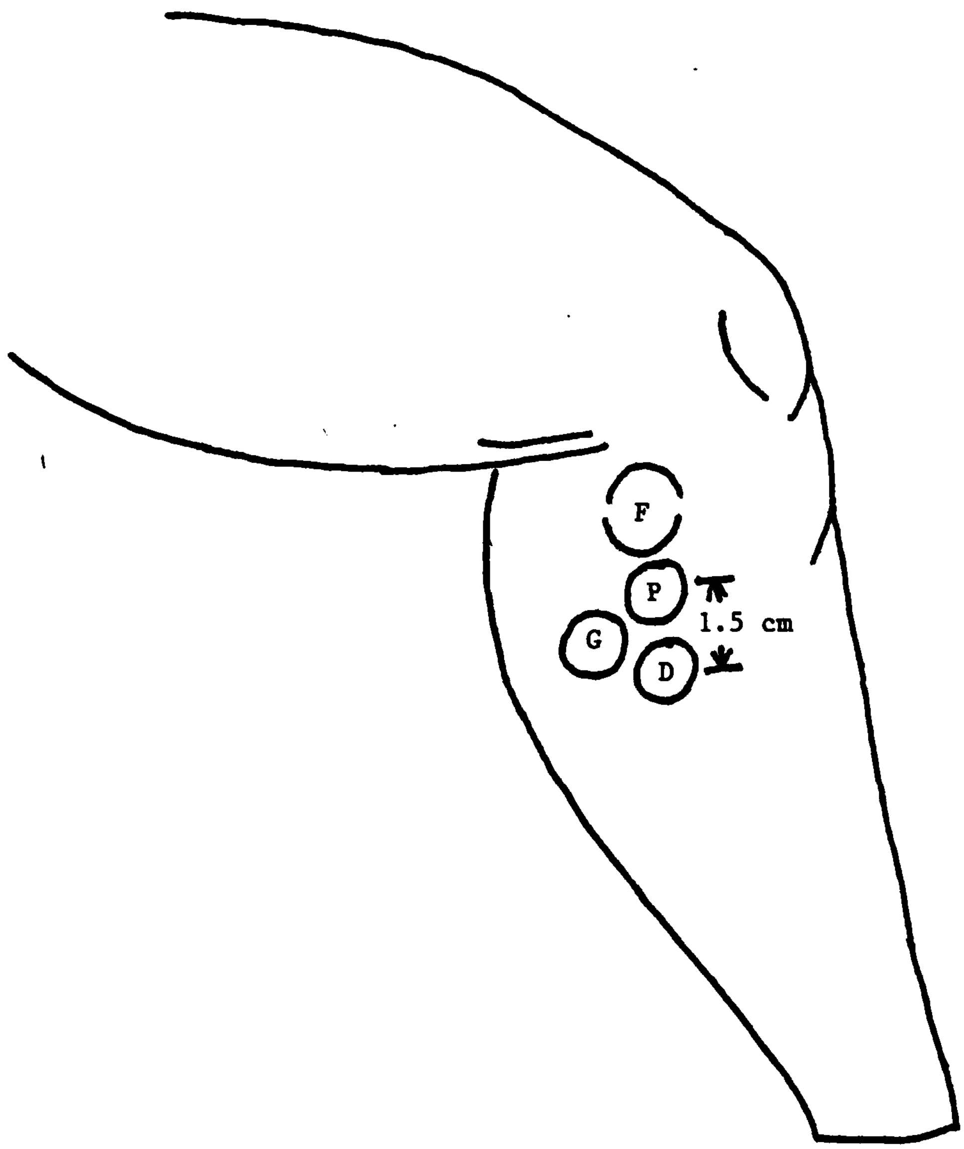

Figure 2: Electrode placement. F = head of fibula; P = proximal active electrode; D = distal active electrode; C = ground electrode

signals. The subject then stepped off the treadmill, and noise level recordings were made with the subject relaxed and the treadmill running. Calibration recordings were then made by passing a known current through the system in order to determine the exact level of amplification. This procedure was repeated whenever the amplification was adjusted during the tests.

The pressure switch was then removed, and the subject was seated on the bicycle ergometer. The seat height, measured from the center of the pedal spindle to the apex of the saddle along the seat tube, was adjusted to equal the subject's trochanteric leg length. This had been found to be the most efficient seat height in terms of oxygen consumption.[37] Work load was set at one, two, or three Kp resistance. These loads have produced significant EMG changes in other muscles of the lower extremity during pedaling.[21] The sequence in which the loads were tested was determined in random fashion. The subject pedaled at a rate of sixty rpm. This is a common speed for which exercise bicycles are calibrated, and it has been found to be easily maintained.[34] Each work load was tested with two different types of pedals. One type was the standard bicycle pedal, and the other was the modified pedal previously described (figure 1). The pedal sequence was determined randomly. When using the modified pedals, the subject was instructed to keep his feet level and not allow the outside unsupported portion of his feet to fall and rest on the standard pedals. Each test condition was maintained by the subject for approximately one minute. In the last ten seconds of this time, a recording was made of the EMG activity of the peroneous longus. The output of the photoelectric cell which had been placed at the bottom (180 degree) position of the pedal cycle was recorded along with the EMG signal. One minute rest periods were provided between tests. The subject remained seated with his elbows extended and his hands on the handle bars during all the tests. Noise level recordings were then made with the subject relaxed and seated on the ergometer.

The subject then laid supine with the test leg supported by a twelve inch platform and stabilized with velcro straps. The test foot was strapped into the eversion/inversion apparatus of the Cybex II system. The leg was leveled by raising or lowering the Cybex II foot plate. The speed control was set at sixty degrees per second, and the test ankle was placed in full inversion. The subject was then instructed to completely evert his ankle as forcefully as possible and to return to the full inversion position. This was repeated three times. orque measurements were recorded on the Cybex II system. EMG recordings were also made during the maximum strength eversion tests. Noise level recordings were then made with the subject relaxed. Eight of the subjects had the entire procedure repeated on their opposite leg.

Data Analysis

Data from nineteen peroneous longus muscles in eleven subjects were compiled and subjected to the statistical analysis. Data from the EMG recordings yielded total activity per cycle from the IEMG signal, peak activity per cycle from the RC signal, and phasic activity per cycle from the raw signal. The ten second recordings resulted in eight to ten complete cycles available for analysis for each bicycle test condition and for the gait condition. The cycles with the highest and lowest maximum amplitudes were not included in the analysis. Three of the remaining cycles were chosen for analysis based on their adherance to a consistent pattern over the recording period. This was determined by simple visual scanning of the recordings. The average value for the three cycles was then computed. The IEMG and RC data collected during pedaling were then normalized by converting them to a percentage of the IEMG and RC data collected during gait for each subject. These percentages were rounded off to the nearest whole number, and used in the statistical analysis.

The IEMG and RC recordings from the maximum eversion strength tests were used to normalize the gait data. This procedure provided some idea of the relative intensity of the electrical activity of the peroneous longus muscle during gait and pedaling. The recordings corresponding to the strongest eversion strength trial out of the three attempted trials were chosen for this purpose. The first half second of this recording was used for analysis. This period always included the maximum RC value for the entire trial. Limiting the analysis to the first half second allowed a more direct comparison of the intensity of electrical activity of the peroneous longus during gait because the average period that the peroneous longus muscle was active during gait was 0.46 seconds.

The phasic activity of the peroneous longus during bicycle ergometry and treadmill gait was determined using the raw signal recorded in each case. The peroneous longus was considered active when the raw signal was twice the height of the base line for at least three consecutive spikes. The phasic data from the pedaling recordings yielded the percentage of the pedal cycle during which the peroneous longus was active for each test condition. This percentage reflected both the onset and termination of EMG activity of the peroneous longus and was used in the statistical analysis. The phasic data from the gait recordings were compared with previous studies as a check of the accurate monitoring of the peroneous longus during this study.

The nine null hypotheses as presented in the introduction (pages 3 and 4) were tested by analysing the appropriate data from the bicycle ergometry tests using a two way analysis of variance (ANOVA) with repeated measures.[16] Each hypothesis was tested at the 0.01 level.

When a null hypothesis was rejected, a protected least significant difference (LSD) method of multiple comparisons of cell means was used to locate the source of significance.[43]

The two additional hypotheses comparing the intensity of EMG activity of the peroneous longus muscle during pedaling with that found during gait as presented in the introduction (page 4) was tested simply by observing the normalized pedaling data. Any mean over 100% represented peroneous longus muscle EMG activity greater than that found during gait.

Chapter IV

RESULTS AND DISCUSSION

Overview of Data

The peroneous longus muscle was electrically active in all six of the pedaling conditions tested in this study. Work load, pedal type, and their interaction were found to cause significant effects on all except one of the characteristics of the EMG signal investigated. A detailed description of the specific results concerning the (1) total EMG activity per cycle; (2) peak EMG activity per cycle; and (3) phasic EMG activity per cycle of the peroneous longus is presented in the following sections. For clarity, the discussion of the results presented in each section immediately follows their presentation. Following these three sections, a general discussion of the relative intensity of the peroneous longus activity is presented. Next, the phasic activity of the peroneous longus during gait as found in this study is compared with previous investigations as a means of supporting the procedures used in this study. Finally, conclusions, clinical implications, and recommendations are offered. All the results presented in this chapter are expressed as a percentage of the gait activity unless stated otherwise.

Total EMG Activity Per Cycle

The mean total EMG activity per cycle for all pedaling conditions ranged from a low of 0.29 (standard pedals at 1 Kp work load) to 1.19 (modified pedals at 3 Kp work load). Table 2 presents the means and standard deviations for all six test conditions. Although the standard deviations are relatively large, an ANOVA resulted in F-ratios that were significant at the 0.01 level for both work load and pedal type as well as their interaction (table 3).

A protected LSD test performed on the differences between all combinations of cell means revealed all to be significant at the 0.05 level except one. The difference between the mean total EMG activity with standard pedals at 3 Kp work load and that found with modified pedals at 1 Kp work load was not found to be significant (appendix A).

A positive, linear relationship was found between work load and total activity per cycle for the peroneous longus during pedaling with standard pedals. With modified pedals, this relationship was slightly curvilinear (figure 3).

Discussion

The first three null hypotheses described in the introduction (page 3) were rejected by these results. These were the primary hypotheses tested by this study and their rejection indicates that increasing work load increases the total EMG activity of the peroneous longus and that the modified pedal elicits more total EMG activity than the standard pedal.

TABLE 2

TOTAL ACTIVITY PER CYCLE

WORK LOAD	STANDARD PEDAL MEAN	(SD)	MODIFIED PEDAL MEAN	(SD)
1 Kp	.29	(.19)	.70	(.26)
2 Kp	.48	(.25)	.91	(.36)
3 Kp	.64	(.26)	1.19	(.34)

All values are percentages of the total activity for gait

TABLE 3

TOTAL ACTIVITY PER CYCLE

ANOVA TABLE

SOURCE	D OF F	S OF S	VAR. EST.	F-RATIO
WORK LOAD	2	3.31	1.65	137.92*
PEDAL TYPE	1	6.18	6.18	44.46*
INTERACTION	36	0.23	0.006	10.00*

* Significant at the 0.01 level

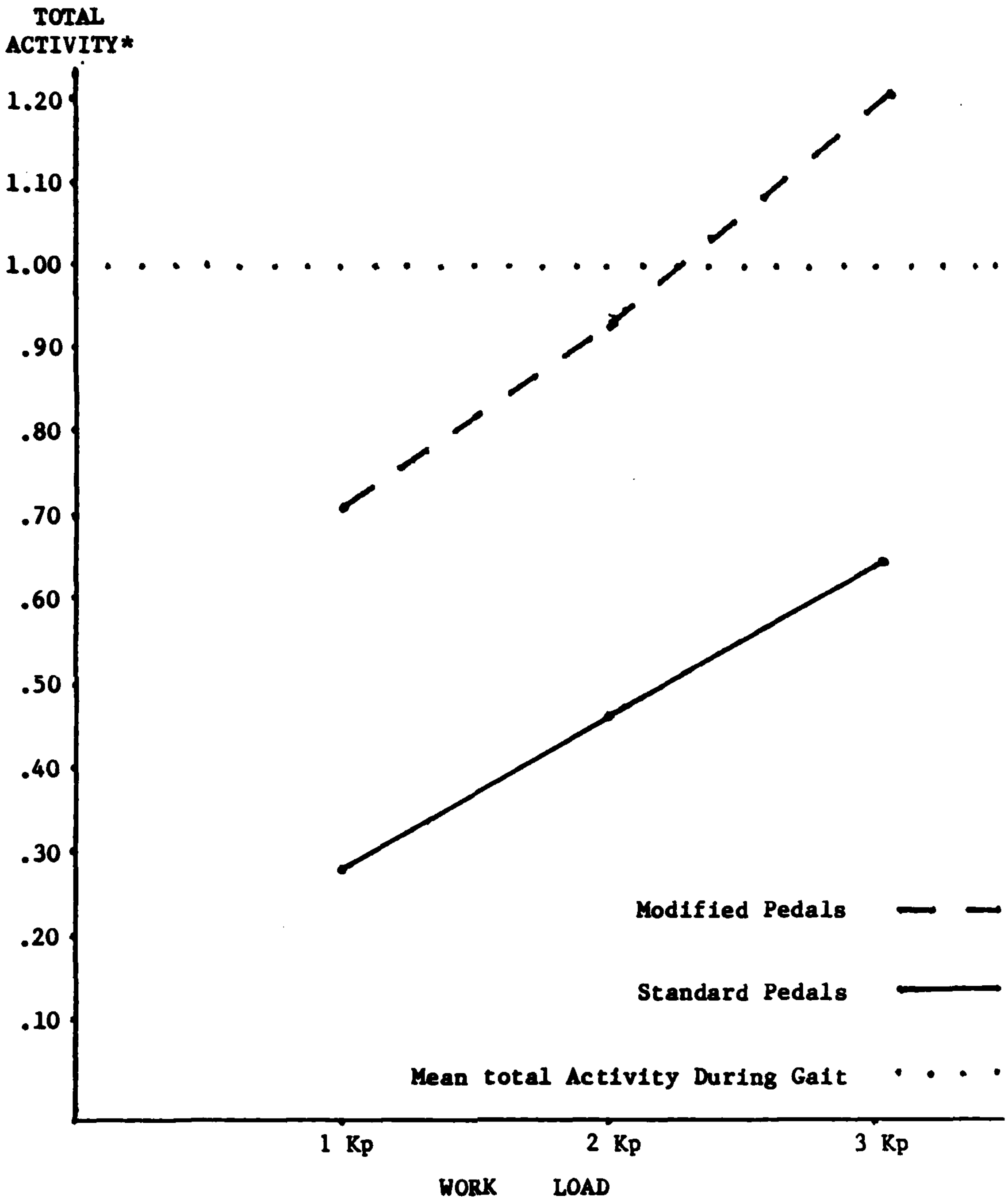

Figure 3: Plot of mean total activity values for standard and modified pedals across work load.

* All values are percentages of the total activity for gait

The standard pedal results are in complete agreement with the work of Goto and co-workers.[21] Although they did not observe the peroneous longus, Goto and co-workers did find a positive linear relationship between total activity and work load for the gastrocnemius, which, like the peroneous longus, is an ankle plantarflexor. As work load increases during pedaling, plantarflexion of the ankle has to become more forceful. All plantarflexors, including the peroneous longus, would increase their activity in response to increased work load. The increased muscle activity would result in increased EMG activity and would account for the increased EMG activity found in this study. Standard pedals are stable in the inversion/eversion plane, and the role of the peroneous longus as a lateral stabilizer during pedaling with standard pedals would be minimal. However, modified pedals provide support only for the medial portion of the foot, and lateral stability is lost. The increased activity of the peroneous longus during pedaling with modified pedals as found in this study may indicate that the peroneous longus provides the lateral stability absent with modified pedals.

An explanation of the activity of the peroneous longus during pedaling with modified pedals may begin with a balance of moments. The force vector for plantarflexion passes through the foot lateral to the support provided by the modified pedal. This results in an inversion moment calculated as follows:

$$IM=(PFF)(Di)$$

The inversion moment (IM) is equal to the plantarflexion force (PFF) times the distance (Di) from the point of application of the plantarflexion force to the outside edge of the modified pedal. This inversion moment must be countered by an eversion moment to stabilize the ankle in the inversion plane. The peroneous longus may provide the eversion moment required. The eversion moment would be calculated as follows:

$$EM=(PLF)(De)$$

The eversion moment (EM) is equal to the peroneous longus force (PLF) times the distance (De) from the point of application of the peroneous longus force to the outside edge of the modified pedal. This equation assumes that the peroneous longus is entirely responsible for the eversion force. Other muscles (peroneous brevis and tertius) are probably also involved, but the peroneous longus has been found to be representative of the whole peroneal group.[49] Figure 4 illustrates the relative location of these moments.

The balance between these moments would be lost if the distances or forces involved were altered. The distances cannot be altered unless the width of the modified pedal is changed. The forces can be altered by increasing the work load during pedaling as was done in this study. As work load is increased, plantarflexion force increases, and the inversion moment increases. Lateral stability would be lost unless the force of the peroneous longus also increased. The increased electrical activity of the peroneous longus during bicycle ergometry with modified pedals in response to increased work load, as found in this study, indicates that the force of the peroneous longus may indeed increase. The slightly curvilinear relationship between work load and total activity found with modified pedals may be a result of the significant interaction effect between work load and pedal type.

Any conclusions about the force of the peroneous longus based on

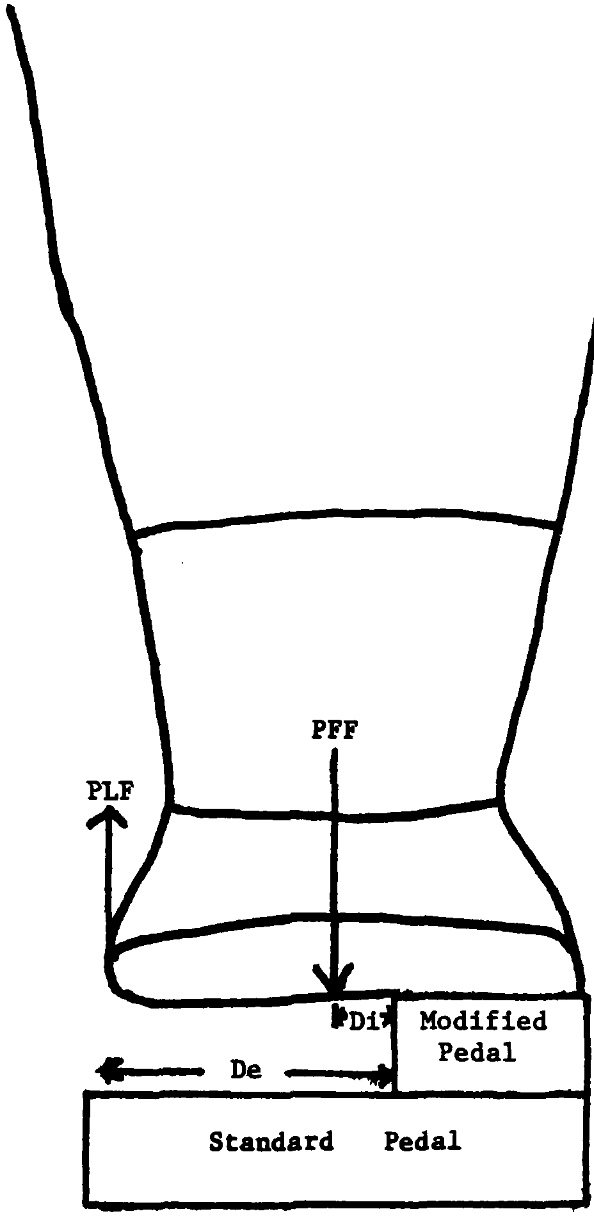

Figure 4: Relative location of moments. PFF = plantarflexion force; PLF = peroneous longus force; Di = inversion distance; De = eversion distance

electrical activity must be carefully made. All factors that affect the electrical activity of the muscle must be considered. In this study, these factors were carefully controlled. The same electrode setup was used in all testing conditions for each subject. The speed of pedaling was maintained at 60 rpm. The fatigue factor was controlled by giving frequent rest periods. The same type of contraction (concentric) was tested in all conditions. These controls allow some general conclusions to be made about the force generated by the peroneous longus during this study. These are: (1) the force of the peroneous longus increases with increasing work load; (2) the force of the peroneous longus increases when modified pedals are substituted for standard pedals; and (3) the force of the peroneous longus increases in response to both increased work load and the modified pedal combined more than it does to either one alone. The force may not have increased in each of these three cases in the same proportion as the electrical activity.

Peak Activity Per Cycle

The mean peak EMG activity per cycle for all pedaling conditions ranged from 0.35 (standard pedal at 1 Kp work load) to 1.12 (modified pedal at 3 Kp work load). Table 4 presents the means and standard deviations for all six test conditions. An ANOVA resulted in F-ratios significant at the 0.01 level for work load, pedal type, and their interaction (table 5).

The differences between all combinations of cell means were significant at the 0.05 level using a protected LSD test except one.

TABLE 4

PEAK ACTIVITY PER CYCLE

WORK LOAD	STANDARD PEDAL MEAN	(SD)	MODIFIED PEDAL MEAN	(SD)
1 Kp	.35	(.17)	.68	(.20)
2 Kp	.53	(.22)	.86	(.26)
3 Kp	.68	(.23)	1.12	(.26)

All values are precentages of the peak activity for gait

TABLE 5

PEAK ACTIVITY PER CYCLE

ANOVA TABLE

SOURCE	D OF F	S OF S	VAR. EST.	F-RATIO
WORK LOAD	2	2.84	1.42	137.92*
PEDAL TYPE	1	3.87	3.87	50.92*
INTERACTION	36	0.19	0.005	7.00*

* Significant at the 0.01 level

There was no difference between the mean peak activity with standard pedals at 3 Kp work load and the mean peak activity with modified pedals at 1 Kp work load (appendix A).

A positive, linear relationship was found between work load and peak activity for the peroneous longus during pedaling with standard pedals. This relationship was slightly curvilinear with modified pedals (figure 5).

Discussion

These results reject the second three null hypotheses described in the introduction (page 4). These were secondary hypotheses concerning peak activity and were tested to determine the cause for the rejection or acceptance of the primary hypotheses concerning total activity.

These findings are in agreement with those reported by Houtz and co-workers and Mohr and co-workers.[24,34] Each of these groups found that increasing the work load during pedaling increased the amplitude of the EMG signal of the muscles they observed. Although they did not study the peroneous longus muscle, they did study other plantarflexor muscles. These investigators did not, however, report the nature of the relationship between work load and amplitude. In this study, the nature of this relationship was very similar to that found between work load and total activity (figures 3 and 5).

Total activity per cycle may be increased in two ways. Either the muscle can be active for a longer period of time during each cycle or the amplitude of the activity may be increased. Some combination

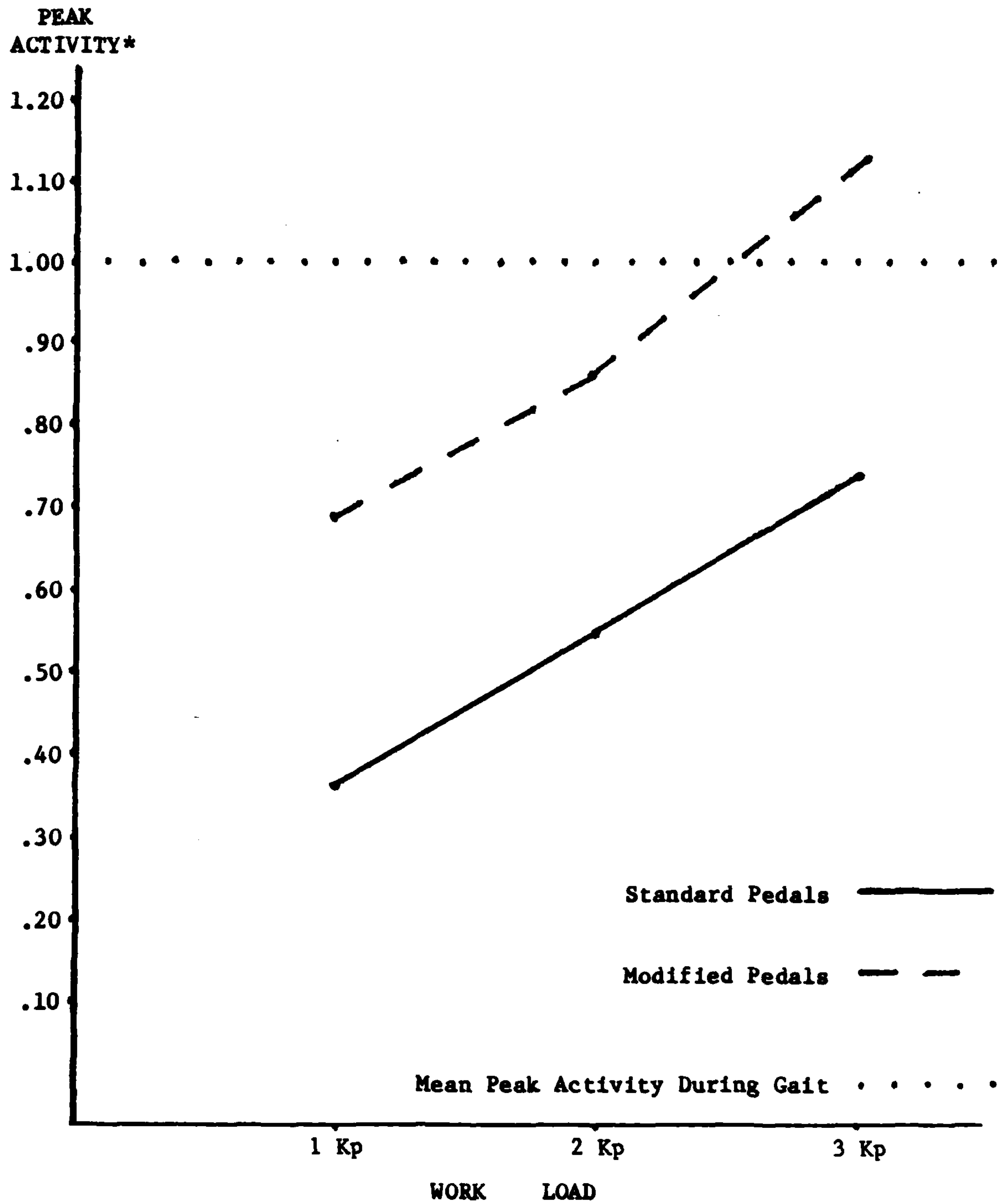

Figure 5: Plot of mean peak activity values for standard and modified pedals across work load.

* All values are percentages of the peak activity for gait

of the two is also possible. Because the nature of the relationship between work load and peak activity is very similar to that found between work load and total activity, the increase in total activity appears to be caused primarily through increased amplitude.

Phasic Activity Per Cycle

The mean percentage of the cycle during which the peroneous longus was active for all pedaling conditions ranged from 35% (standard pedals at 1 Kp work load) to 57% (modified pedals at 3 Kp work load). Table 6 presents the means and standard deviations for all six test conditions. An ANOVA resulted in F-ratios significant at the 0.01 level for work load and pedal type but not their interaction (table 7).

The protected LSD test indicated that the differences between any combination of a standard pedal condition with a modified pedal condition, regardless of work load, was significant at the 0.05 level. No significant differences were found between any combination of cell means when both conditions used the modified pedals. When both conditions used standard pedals, the only difference found to be significant was that between 1 and 3 Kp work loads (appendix A).

The earliest mean onset of peroneous longus electrical activity occurred with modified pedals at 3 Kp work load. At all work loads, the modified pedal conditions had activity before the standard pedal conditions. Also, at all work loads, the modified pedal conditions had activity beyond 180 degrees of the cycle while the standard pedal conditions did not. Figure 6 presents the mean onset and termination

TABLE 6

PHASIC ACTIVITY PER CYCLE

WORK LOAD	STANDARD PEDAL MEAN	(SD)	MODIFIED PEDAL MEAN	(SD)
1 Kp	.35	(.10)	.53	(.11)
2 Kp	.41	(.10)	.57	(.09)
3 Kp	.44	(.11)	.56	(.12)

All values are precentages of the cycle during which the peroneous longus muscle was active

TABLE 7

PHASIC ACTIVITY PER CYCLE

ANOVA TABLE

SOURCE	D OF F	S OF S	VAR. EST.	F-RATIO
WORK LOAD	2	.075	.0375	12.50*
PEDAL TYPE	1	.649	.649	37.08*
INTERACTION	36	.156	.0043	1.40

* Significant at the 0.01 level

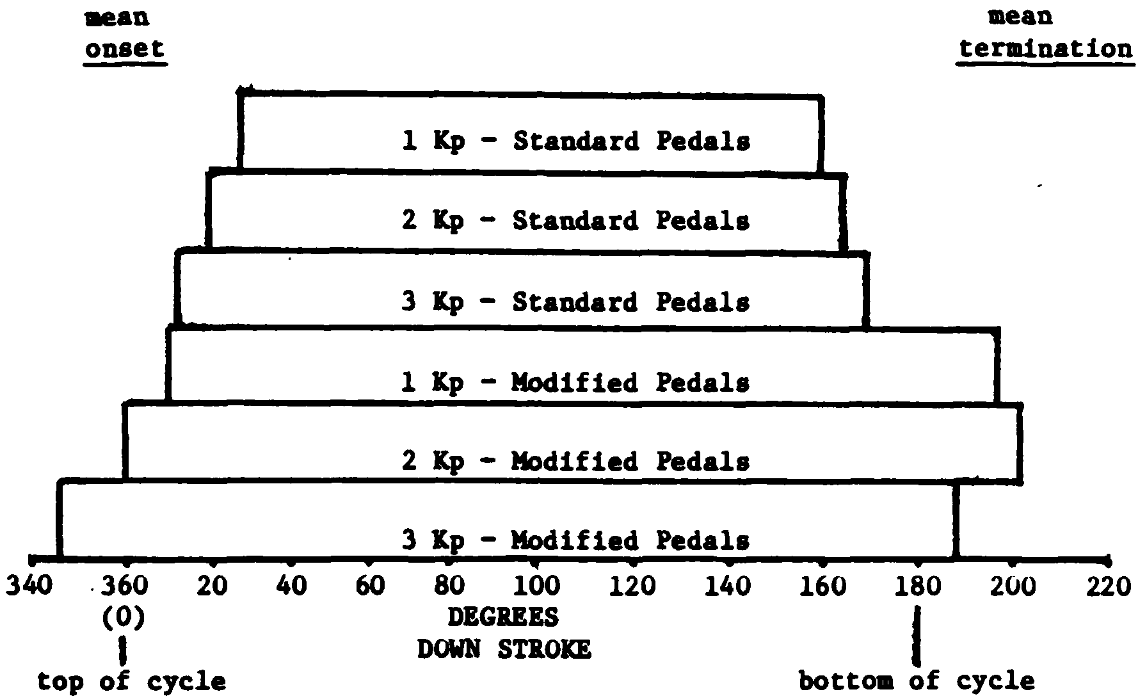

Figure 6: Phasic activity. Onset and termination of EMG activity of the peroneous longus during pedaling in all six conditions.

of electrical activity of the peroneous longus for all six pedaling conditions.

Discussion

The seventh and eighth null hypotheses as presented in the introduction (page 4) were rejected by these results but the ninth was not. As with the secondary hypotheses concerning peak activity, these secondary hypotheses concerning phasic activity were tested to help determine the cause for the rejection or acceptance of the primary hypotheses concerning total activity.

The significant effect of increased work load on the phasic activity of muscles active during pedaling has not been reported by any other investigator. Houtz and co-workers and Mohr and co-workers reported no change in the phasic activity of the muscles they studied with increasing work load.[24,34] The disagreement between their results and the results of this study may be explained in several ways. They may have used different criteria for determining the onset and termination of activity. Also, they did not report using statistical tests for significance. Although this study did find a significant effect from increased work load on the phasic activity of the peroneous longus muscle, this effect was only significant between 1 and 3 Kp work loads. Increases of only 1 Kp did not produce significant differences in the phasic activity. Pedal type had the greatest effect on the phasic activity. The modified pedal required peroneous longus activity not only for plantarflexion, but also for lateral stability. However, since the ankle has to dorsiflex during the upstroke of the cycle, peroneous longus activity at that time would be counter productive. For this reason, the peroneous longus has to cease activity soon after 180 degrees. The foot is probably kept on the modified pedal during this time by internal rotation of the hip. The internal rotation of the hip was documented by this investigator using cinematography to observe the kinematics of pedaling with modified pedals in a one subject pilot study. The lack of a significant effect from the interaction of work load and pedal type was expected. The modified pedal produces such a large increase in the phasic activity that the effect of increased work load is minimal in comparison. As a result, adding additional resistance while using modified pedals does not affect the phasic activity. These results indicate that the increase in total activity with increasing work load as found by this study is not influenced a great deal by increases in phasic activity. The increase in total activity with modified pedals is influenced by increases in phasic activity, but the major influence comes from the increase in peak activity.

Relative Intensity of EMG Activity

The normalized results presented in the previous sections on total and peak activity support the two additional hypotheses concerning the relative intensity of the peroneous longus muscle activity presented in the introduction (page 4). The highests normalized mean value for total activity using standard pedals was 64% at 3 Kp work load. The value for peak activity was 68%. This indicates that pedaling with standard pedals, even at heavy work loads, requires less than 70% of the electrical activity of the peroneous longus while walking. However, with modified pedals, pedaling at 3 Kp work load required 119% (total activity) and 112% (peak activity) of the electrical activity of the peroneous longus while walking. The gait data were probably inflated because of the lack of experience of seven of the subjects with walking on a treadmill. Because of this inflation, the percentages for pedaling may be slightly lower than they would be if over ground walking had been used to normalize the data.

Making conclusions about the forces produced by the peroneous longus during pedaling as compared with gait must be done with care. Factors that affect the electrical activity of a muscle were controlled as much as possible, but walking and pedaling are not identical activities. For this reason, the most that may be concluded is that pedaling with modified pedals at the 3 Kp work load requires a force from the peroneous longus at least similar in magnitude to that required by gait, and clearly greater than that required by pedaling with standard pedals at any work load.

Comparing the force of the peroneous longus during gait with the force during a maximum eversion contraction based on the electrical activity is even more tenuous than the pedal-gait comparison. Gait required 46% of the total electrical activity of the peroneous longus during the maximum eversion contraction and 51% of the peak activity. However, the same problem of comparing dissimilar activities is present here as in the pedal-gait comparison. The only conclusion about the force of the peroneous longus during gait that may be made is that it is probably somewhat less than half the maximum force possible from the peroneous longus.

Phasic Activity During Gait

The phasic activity of the peroneous longus during gait as found in this study agreed with the work of Gray and Basmajian, Walmsley, and Sutherland.[22,49,44] Each of these investigators reported the onset of peroneous longus activity shortly after heel strike. The present study found activity beginning at a mean of 7% (SD=4%) of the gait cycle. The previous investigations also reported the termination of activity around 54% of the gait cycle. The present study found the activity ceasing at a mean of 53% (SD=3%) of the gait cycle. This agreement between previous gait studies and this study suggests that the electrical activity of the peroneous longus was accurately monitored during this study.

Conclusions

The following conclusions are presented based on the results of this study:

1. Increasing the work load during pedaling with either standard or modified pedals increases the total EMG activity of the peroneous longus.
2. The increase in total EMG activity of the peroneous longus with increasing work load is primarily the result of increased peak activity.
3. Substituting modified pedals for standard pedals increases the total EMG activity of the peroneous longus during pedaling.
4. The increase in total EMG activity of the peroneous longus during

pedaling in response to modified pedals is primarily the result of increased peak activity but is also influenced by increased phasic activity.

5. Pedaling with standard pedals, even at heavy work loads, requires less total EMG activity of the peroneous longus than does gait.
6. Pedaling with modified pedals at moderate and heavy work loads requires total EMG activity similar to that required during gait.
7. The activity of the peroneous longus muscle during gait is probably less than half of its maximum isokinetic contraction.

Clinical Implications

The results of this study suggest that bicycle ergometry can be used to increase the endurance capability of the peroneous longus. The progressive increase in peroneous longus activity with increasing work load and changing pedal type offers effective means of controlling the stress placed on the peroneous longus during rehabilitation. However, the low intensity levels found with standard pedals, even at heavy work loads, indicate that real increases in endurance may depend on progressing to the modified pedals with at least moderate levels of resistance. The standard pedals would be of value primarily early in the rehabilitation process when only slight activity from the peroneous longus is desired for the reduction of edema. The modified pedals would be effective at the start of the strengthening portion of the rehabilitation program before weight bearing can be performed safely. However, even the modified pedals cannot take the place of functional weight bearing activities.

Recommendations

Although this study suggests that bicycle ergometry can increase the endurance capability of the peroneous longus, further work needs to be done. Clinical studies are needed to determine if endurance is, in fact, improved and recurring injuries really are reduced when bicycle ergometry is included in the rehabilitation process. In addition, the effect of varying the rate of pedaling should be investigated. The literature suggests that increasing the rate of pedaling increases the activity of the muscles involved. The peroneous longus should be specifically studied in this regard. Repeating this study with female subjects and subjects from the general population with a wide age range should also be done. This would improve the ability to generalize the findings of the present study.

Additional aspects of the rehabilitation of the peroneous longus muscle need to be investigated. When the ankle suddenly turns as a result of stepping into a hole or on someone's foot, a powerful contraction of the peroneous longus is required to prevent abnormal motion and resultant soft tissue or bony damage. This power requirement may dictate that high speed isokinetic training should be included in the rehabilitation of the peroneous longus muscle. Studies that focus on this power requirement should be performed.

The modified pedal used in this study was designed after several pilot studies, but with no professional engineering consultation. The modified pedal could probably be improved with the addition of engineering and design skills. With the proper financial and technical support, a modified pedal could be produced that could be used alone without the need of a standard pedal as a base. An improved modified pedal that could be used alone would have better balance characteristics and would make clinical use more practical.

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APPENDIX A

LSD TABLE FOR TOTAL ACTIVITY PER CYCLE PRESENTING DIFFERENCES (U1 - U2) BETWEEN CELL MEANS FOR EACH CONDITION

CONDITION	SP/1Kp	SP/2Kp	SP/3Kp	MP/1Kp	MP/2Kp	MP/3Kp
SP/1Kp	0					
SP/2Kp	.19	0				
SP/3Kp	.35	.16	0			
MP/1Kp	.41	.22	.06	0		
MP/2Kp	.62	.43	.27	.21	0	
MP/3Kp	.90	.71	.55	.49	.28	0

U1 - U2 GREATER THAN .0917 SIGNIFICANT AT THE 0.05 LEVEL

LSD TABLE FOR PEAK ACTIVITY PER CYCLE PRESENTING DIFFERENCES (U1 - U2) BETWEEN CELL MEANS FOR EACH CONDITION

CONDITION	SP/1Kp	SP/2Kp	SP/3Kp	MP/1Kp	MP/2Kp	MP/3Kp
SP/1Kp	0					
SP/2Kp	.18	0				
SP/3Kp	.33	.15	0			
MP/1Kp	.33	.15	0	0		
MP/2Kp	.51	.33	.18	.18	0	
MP/3Kp	.77	.59	.44	.44	.26	0

U1 - U2 GREATER THAN .1455 SIGNIFICANT AT THE 0.05 LEVEL

APPENDIX A

LSD TABLE FOR PHASIC ACTIVITY PER CYCLE PRESENTING DIFFERENCES (U1 - U2) BETWEEN CELL MEANS FOR EACH CONDITION

CONDITION	SP/1Kp	SP/2Kp	SP/3Kp	MP/1Kp	MP/2Kp	MP/3Kp
SP/1Kp	0					
SP/2Kp	.06	0				
SP/3Kp	.09	.03	0			
MP/1Kp	.18	.12	.09	0		
MP/2Kp	.22	.16	.13	.04	0	
MP/3Kp	.21	.15	.12	.03	.01	0

U1 - U2 GREATER THAN .0683 SIGNIFICANT AT THE 0.05 LEVEL

APPENDIX B

HUMAN SUBJECT CONSENT FORM

This consent form is required by the University of North Carolina for any research involving human subjects. If you sign this form you are consenting to be a subject in this research study. Please read this entire form carefully before you sign. If you do not understand something, please feel free to ask. Thank you for considering to be a subject in this study.

1. "I understand that this study involves research to observe the action of a lower leg muscle (peroneous longus) during pedaling on a stationary bicycle and while walking on a treadmill."

2. "I further understand that the muscle being observed will be monitored using surface electrodes attached to one or both legs. This requires a small area of one or both legs to be shaved and the skin slightly abraided using a dental burr to reduce skin resistance. This procedure will not be uncomfortable and will not result in any permanent change in my skin but may cause some temporary reddening."

3. "I know that the electrode sites will be located using a small direct current generator which will feel like a slight tingle or small shock."

4. "I will be required to pedal a stationary bicycle at sixty revolutions per minute. I will pedal one minute at each of three different work loads ranging from light to heavy. I will be given a one minute rest between work loads. This procedure will be performed with standard pedals and with modified pedals. I will also be required to walk on a treadmill for approximately one minute at sixty cycles per minute."

5. "Although there are no reasonably expected risks, I understand that there may be some unforeseeable ones. I know that no physical benefit will result and I will receive no monetary compensation."

6. "I further understand that all data will be confidential and my name will not be used in any presentation of this study, written or oral."

7. "I know that I can withdraw from this study at any time without penalty and for any or no reason."

8. "This study will not cost me anything except two hours of my time."

9. "The principal investigator is Danny Lee Holt and I can contact him at anytime at the Physical Therapy Division of the UNC Medical School (933-4828), or at his home (933-9375)."

10. "I may also contact the Chairman of the Committee on the Protection of the Rights of Human Subjects, Edward H. Bishop, MD, at 966-1601, if I feel there has been any infringement of my rights."

11. "I understand that in the event of physical injury directly resulting from the research procedures, financial compensation cannot be provided. However, every effort will be made to make available to me the facilities and professional skills of the University of North Carolina at Chapel Hill."

12. "I understand that all subjects involved in this study, approximately ten to fifteen, are volunteers and are participating under these same conditions."

13. "All my questions have been satisfactorily answered by the principal investigator and my signature below indicates my consent to be a subject in this study."

Subject Signature:______________________________ Date:_______________

Witness Signature:______________________________ Date:_______________

END

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