Lower Limb Muscle Function During Cycling

by

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Dedication

To my parents, Leo and Anne, who have given me support and encouragement above and beyond the normal limits of parenthood.

Acknowledgements

I once believed that completing masters program was a solitary endeavor. Nothing could be further from the truth. I would like to take this opportunity to express my thanks to the many people who assisted me towards the completion of this thesis.

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Abstract

The purpose of this study was to describe the functional role of the lower limb musculature during stationary cycling using electromyography, muscle-tendon unit length changes, and segmental kinematics. Five subjects were filmed (100 Hz) in synchrony with the collection of LE EMG activity of the gluteus maximus, semitendinosus, semimembranosus, rectus femoris, vastus lateralis, soleus, gastrocnemius, and tibialis anterior muscles during stationary cycling at 160 W (90 r/min). The results showed that extension during the propulsive phase of the pedal cycle was the result of high concentric activity of both the monoarticular and biarticular muscles. Furthermore, these muscles functioned according to their expected anatomical roles (Rasch and Burke, 1978). This investigation, therefore, finds little evidence for the existence of paradoxical muscle function as hypothesized by Lombard (1903), Molbech (1965), or Rasch & Burke (1978).

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Introduction

The necessity of understanding the functional roles of the lower limb musculature during dynamic movements has prompted numerous investigations from a multidisciplinary group of researchers. The definition of the functional role(s) of the muscles are, however, complicated by the number of biarticular muscles that cross the joints of the lower limb and the presence of co-activation of these muscle groups (Wells and Evans, 1987). To account for these factors several models of biarticular muscle function have been proposed.

The earliest hypothesis regarding the functional role of biarticular muscles was proposed by Lombard (1903) and this has since been given the name Lombard's Paradox. Lombard stated that opposing groups of biarticular muscles, when co-activated, would cause extension of both the muscles' included joints. More recently, investigators have defined the paradox as the " activity of a two-joint (biarticular) muscle when the required moment at one of the joints is in the opposite direction to that caused by the muscle " (Gregor, Cavanagh, & Lafortune,1985). A common example of this paradoxical function, in humans, is demonstrated when rising from a chair. During this movement it has been shown electromyographically that the biarticular muscles of the hamstring group and the rectus femoris are co-activated. This would require that the hip extensor moment of the hamstrings (and other active hip extensors) dominate over the hip flexor action of the rectus femoris at the hip joint and that the extensor moment of the quadriceps muscle be larger than the knee flexor moment of the hamstrings at the knee joint. The activity of these two muscle groups during such an movement is apparently appropriate and therefore nonparadoxical at one joint, and apparently inappropriate and paradoxical at the other joint (Andrews, 1987). Unfortunately, Lombard did not test this hypothesis on humans.

Rasch and Burke (1978) examined Lombard 's Paradox (1903) utilizing simple models based on anatomical considerations. The results supported the conditions to produce extension hypothesized by Lombard (1903). In addition, they expanded the analysis to include a situation where one muscle maintained a "belt-like or tendinous " action (constant length) and concluded that extension or flexion would occur if either the agonist or antagonist were passive. It must be noted that the results of this simplistic model were obtained with uncontrolled "muscle model forces" and a limited number of modelled muscles.

Molbech (1965) proposed an alternate theory to explain the functional role of biarticular muscles. He classified the functional role of a biarticular muscle in a closed kinematic chain (steered movement) based on its moment about the instantaneous center of rotation and the associated change in the configuration of the particular joint traversed by the muscle. This model allowed a biarticular muscle to act as both an extensor and a flexor at the same joint. While Molbech stated that the model could be adapted to examine the functional role of any biarticular muscle, Andrews and Hay (1983) have presented several shortcomings in this mathematical model.

Andrews (1985) presented a new classification scheme based on the techniques of equilibrium analysis. The classification of muscle function by this method was based on the assumption that the linkage system would always tend to move away from the initial rest configuration, in such a way as to decrease the length of the muscle in question. The muscle would act to bring its attachment points closer together when the system is released from its initial rest configuration, resulting in extension or flexion of the included joint. The author acknowledged that the results of such an analysis were limited to the confines of the model.

The potential for paradoxical function of the biarticular muscles of the leg during the propulsive phase of cycling has prompted several investigators to examine this activity. These investigators included Carlsoo and Molbech (1966), Gregor et al. (1985), and Andrews (1987).

Carlsoo and Molbech (1966) investigated the functional role of the biarticular muscles of the lower limb during stationary cycling as a means of validating Molbech's (1965) model. Using this model, and the results of the EMG activity the authors determined that the hamstring activity could be divided into three phases: the initial phase where the main activity of the biceps femoris was to counteract the hip flexors, while providing knee flexion (although knee extension had begun); the second phase where the muscle changes the knee joint movement from flexion to extension (called the limiting line), and the third phase when the knee bends as in a free hanging limb. The authors concluded that the hamstrings could act paradoxically to assist in flexion and extension at the knee.

Gregor et al. (1985) examined the EMG activity and net joint moments of force during stationary cycling in an attempt to verify the existence of Lombard's Paradox during cycling. They explained that knee extension during the propulsive phase was due to a flexor moment at the knee and to the leg's orientation allowing the resultant pedal reaction force to pass anterior to the knee joint. This resulted in a net extensor moment that overcame the flexor moment. These authors concluded that the hamstrings and the quadriceps contributed to the moments and avoided "Lombard's Paradox".

Andrews (1987) attempted to clarify the function and the classification of biarticular muscles. He classified the various techniques into two groups. One group, the "Standard Kinetic (SK) technique", included the methods that established a muscle's functional role based on kinetic or moment dependent considerations (i.e., Lombard, 1903; Rasch and Burke, 1978). The second classification method, developed by Andrews (1985), was based on kinematic or configuration and motion considerations and was referred to as the "Andrews Kinematic (AK) Method ".

The application of the SK and the AK techniques to cycling indicated that the hamstrings and quadriceps functioned both paradoxically and nonparadoxically at the hip and knee joints. Furthermore, the SK method indicated that the hamstrings functioned nonparadoxically at the hip, with both muscle groups exhibiting approximately equal regions of paradoxical and nonparadoxical behavior at the knee. In contrast, the AK method determined that the quadriceps functioned nonparadoxically at both the hip and knee, whereas the hamstrings acted paradoxically at the knee and exhibited essentially equal regions of paradoxical and nonparadoxical behavior at the hip. The differences in the conclusions reached by the two methods was expected as the nature of the paradox in each technique was different.

Andrews concluded the manner in which one determines paradoxical/ nonparadoxical biarticular muscle function is affected by the method used to determine when muscle is active, the criterion used to establish the existence of paradoxical behavior, and the classification method used to define the functional role of the muscle. Furthermore, he stated that the least significant of the three was the manner in which muscle activity was determined.

Despite the application of several biarticular muscle models to cycling, the presence or absence of paradoxical activity has not been resolved. Ideally, the presence of paradoxical biarticular muscle function as defined by Lombard (1903) could be determined by a comparison of the lower limb's segmental kinematics and kinetics to the active muscles' force, the type of activation (isometric, concentric or eccentric), and the muscle's moment arm about the included joint(s). Investigations regarding the segmental kinematics and kinetics have revealed consistent patterns (Pons and Vaughan, 1989). Unfortunately, the difficulties associated with the measurement of muscle force and muscle kinematics have limited the understanding of the muscle-tendon units (MTUs).

Early attempts to understand a muscle's force production utilized *in vitro* methods because a controlled environment was required to allow the individual muscle properties to be measured. For instance, Hill (1938) found that the force produced by a muscle was dependent upon the length of the individual muscle fiber (length-tension relationship) and the rate of shortening (force-velocity relationship). More recently, advances in technology have allowed muscle force during movements to be estimated *in vivo* in animals (Gregor, Komi, and Jarvinen, 1987). Experimental difficulties, however, limit the direct measurement of force in humans (Lamontagne, 1986) and have necessitated that researchers approximate the muscle's force production by alternate techniques.

Many investigators have attempted to approximate the force from the neuromuscular activation by processing electromyographic (EMG) signals. While many techniques have been presented to capture and process the raw EMG activity, Winter (1984) has suggested using the full-wave rectified and low pass filtered signal (linear envelope (LE)) because it follows the trend and closely resembles the shape of the force curve by providing an analog pattern that is reliable and reproducible. Yang and Winter (1984) have shown that the time base of the LE EMG can be normalized to 100% of the movement cycle and the amplitude normalized to the peak of within-subject ensemble average to facilitate comparison across subjects.

The EMG activity of the lower limb musculature during cycling has been well described. Houtz and Fisher (1959) examined the electrical activity of fourteen muscles of cyclists riding a stationary ergometer. The experimental protocol included two saddle heights, but no cadence was specified. Despires (1974) examined the effect of alterations in saddle height and load patterns on twelve muscles of elite subjects riding a standard bicycle on a treadmill. Gregor, Green and Garhammer (1982) investigated eight muscles of the lower limb of elite cyclists riding at a constant cadence and workload on a modified laboratory ergometer that was adjusted to the subject's "regular" riding position. Gregor et al. (1985) investigated four muscles of the lower limb on a conventional bicycle attached to a customized ergometer. The above investigations revealed several discrepancies in the activity periods of at least one muscle, yet the observation of large periods of co-activation of the quadriceps and hamstrings during the power stroke were common to all studies. In contrast to the above investigations, Jorge and Hull

(1986) examined several subjects at various loads, cadences, foot-pedal interfaces, and saddle heighιs. These authors reported reciprocal activation of agonist and antagonist muscles at the hip, knee and ankle joints during the pedal cycle.

Several methods have been presented to approximate the MTU kinematics and moment arms during movement. The early approximations were based on mechanical devices (Inman, 1947; Merchant, 1965; Sorbie and Zalter, 1965); whereas, later investigations utilized geometric principles to define the muscles as (a) a series of straight lines joining the origins to insertions (Hardt, 1978), (b) the summation of linear and/or curvilinear sections to represent the muscle (Serig and Arvikar,1973; Frigo and Pedotti,1978; Simonsen, Thomsen and Klausen, 1985), (c) lines that pass through the centroids of thin cross-sectional slices of the muscle (Jensen and Davy,1975), or (d) the summation of linear and/or curvilinear sections in three-dimensions (Brand, Crowninshield, Wittstock, Pedersen, & Clark, 1982; Mansour and Pereira,1987; Pierrynowski,1982). The realization that all models indicate only a relative measure of MTU length (Grieve, Pheasant and Cavanagh 1978) and that investigators must maintain a balance between accuracy and simplicity (Pierrynowski, 1982) has prompted investigators (Hubley, 1981; Curwin, 1984; MacKinnon, 1988; Wilson, 1989) to choose models based on the techniques of Frigo and Pedotti (1978).

Few papers to date have specifically examined the changes in length of the lower limb's MTUs during cycling. Gregor et al. (1987) presented changes of length in the gastrocnemius and the soleus of a single subject at various workloads and cadences on a stationary ergometer. Redfield and Hull (1986) provided an optimization model that required geometric modelling of 13 muscles of the lower limb during cycling, yet they did not present these data.

In summary, the conflicting results of the EMG activity of the lower limb in the previous cycling investigations and the limited understanding of the changes in the MTUs during cycling necessitate the collection of empirical EMG data in synchrony with changes in the MTU length. These data taken together with the segmental kinematics enable a more definitive way of determining the functional role of muscles and the presence or absence of paradoxical biarticular muscle activity.

Purpose

The primary objective of this investigation was to classify the functional role of the lower limb musculature during the propulsive phase of the pedal cycle using changes in the LE EMG patterns, the MTUs' lengths, and the segmental kinematics. The secondary objective was to determine whether the biarticular muscles that cross the knee joint functioned in a paradoxical manner.

Methods

System Description

The cyclist and bicycle were represented by a five bar, sagittal plane linkage system interconnected by frictionless pin joints (see Figure 1). Each segment was characterized by its mass, center of gravity, moment of inertia and radius of gyration (Dempster, 1955; Plagenhoef, 1971).

Eight muscles that are considered the prime movers in the sagittal plane (Rasch and Burke, 1978) represented the musculature of the lower extremity. The muscles and the included joints are presented in Table 1. The muscle-tendon unit length and moment arm(s) at joint crossings were approximated *in vivo* based on an anatomical model by Frigo and Pedotti (1978) and later adapted by Hubley (1981). A complete description of the individual muscle-tendon units is included in Wilson (1988) and Hubley (1981).

Subject

Five elite male cyclists between 21-28 years of age volunteered to participate as subjects in this study. The subjects were free of injury, had no known neurological disorders or surgical alterations that affected the biomechanical aspects of cycling, and were capable of maintaining the desired workloads throughout the test protocol.

Table 1

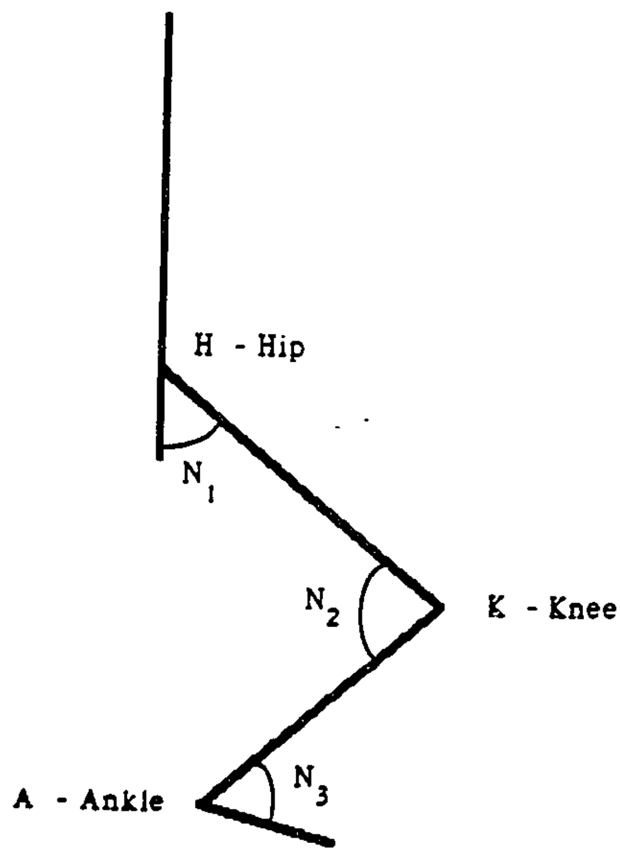
Muscles included in the model

Muscle	Joint Affected		
	Hip	Knee	Knee
gluteus maximus (GM)	X		
rectus femoris (RF)	X	X	
vastus lateralis (VL)		X	
semimembranosus (SM)	X	X	
semitendinosus (ST)	X	X	
gastrocnemius (GA)		X	X
soleus (SO)		X	
tibialis anterior (TA)			X

Table 2

Anatomical markers

Points	Segment
1	Greater Trochanter
2	Lateral Femoral Condyle
3	Lateral Malleolus
4	Posterior Calcaneous
5	Fifth Metatarsal-Phalangeal Joint
6	Pedal Axle Center
7	Crank Axle Center



RHT = Standing height of subjects in centimetres.

Figure 1 System representation

Subject Preparation

Five anatomical landmarks and two additional points were identified by the placement of contrasting white adhesive markers (see Table 2). For EMG collection the skin was prepared according to the methods described by Basmajian (1959). Electrodes (Meditrace silver/silver chloride) were placed 2.5 cm apart parallel to the longitudinal line of the belly of the modelled muscles above the motor points as described in Warfel (1985). Skin resisitance was verified to be below 20 kilohms.

Cycle Ergometer

To simulate road cycling conditions in the laboratory, a standard racing bicycle was stabilized in a stand. The rear wheel of the racing bicycle was removed and the chain was attached to an ergometer flywheel (Monark). The handlebar and saddle were customized to allow easy adjustment in both horizontal and vertical directions. A cadence meter (Avocet) was placed on the handlebar.

Prior to the data collection the saddle height was adjusted to 100% trochanter height (the distance from the proximal aspect of the greater trochanter to the base of the foot with both heels touching (Redfield and Hull, 1986)). The handlebar and stem were adjusted to rider comfort.

Experimental Protocol

After an adequate warm-up on the bicycle each subject was required to pedal at 90 r/min at a constant workload (160 W) as approximated by the resistance scale on the ergometer. When the subject was pedalling at the cadence set on a calibrated electronic metronome (Franz) the data were collected. To accommodate a shortage of bioamplifiers six muscles were captured during an initial trial and a second trial was performed to capture the two remaining muscles (gastrocnemius and soleus). The identical histories of the doubly collected EMG signals justified superimposing the linear envelope EMG (LE EMG) histories from both trials.

Data Aquisition

All trials were filmed (50 frames/s) by a high-speed cinecamera (16 mm Locam) placed orthogonally to the sagittal plane. The filming protocol included filming a grid of known coordinates to calibrate the field of view (Robertson, 1977), filming of static profiles of the subjects, and the filming of the cycling trials. An internal timing light was exposed on the border of the film at a frequency of 100 Hz to verify the true film speed.

The EMG leads were connected directly to a high input impedance (10 megohm) bioamplifier (10–700 Hz band pass filter, gain of 1000). The signals were full-wave rectified and filtered (second order Butterworth filter at 6 Hz cut-off) to emerge as a LE EMG signal. The LE EMG were collected (analogue-to-digital) by a minicomputer (Data General microEclipse).

Synchronization of the cinefilm and EMG signals was accomplished by the camera's shutter pulse correlator (which registers the opening of the shutter). Synchronization pulses activated a sweep of EMG data collection, triggered counting lights on a digital light counter located within the camera's field of view, and were simultaneously recorded on a channel of the microcomputer.

Data Analysis

The marker coordinates of one complete pedal cycle were digitized (Hewlett-Parkard 9874A) and transferred to a mainframe computer (Amdahl 5860). These data were processed by the computer program BIOMECH (Kinesiology Dept., University of Waterloo). This program corrected camera and projector misalignment by a fractional linear transformation, digitally filtered the data (fourth order, Butterworth lowpass filter with 6 Hz as the cut-off frequency), and computed the linear and angular kinematics of the segments using finite differentiation (Pezzack, Winter and Norman, 1977). The 6 Hz cut-off frequency was selected because it retained greater than 95% of the original signal. These kinematic data were then time normalized and averaged across subjects.

MTU lengths for each subject were calculated from the joint angle changes (Frigo and Pedotti, 1978; Hubley, 1981). These data were normalized to percent of cycle time and to the standing anatomical length (i.e., the length of the MTU measured when the subject was standing erect). The rates of change of the MTUs were computed using finite differentiation. The normalized histories were averaged across subjects, resulting in grand ensemble MTU histories.

Each muscle's LE EMG signal was normalized to percentage of cycle time. The time normalized EMG data for each subject were then averaged across five cycles, generating a within-subject ensemble average for each muscle. These within-subject ensemble averages were then normalized to the maximum within-subject (MWSA) amplitude to produce the across-subject grand ensemble averages (Yang & Winter, 1984).

Results and Discussion

Segmental Kinematics

Figure 2 presents the hip, knee, and ankle joints' grand ensemble averaged angular displacements and velocities (n = 5). The abscissae are shown as a function of cycle time (where 100 % represents one complete revolution) and the ordinates have been scaled to allow maximum resolution of the data points. Displacement characteristics are presented in Table 3. Top dead center (TDC), front dead center (FDC), bottom dead center (BDC), and rear dead center (RDC) occurred on average at 0, 24, 52, and 78% of the normalized cycle, respectively.

All three joint displacement histories were characterized by cyclical monophasic patterns, however, the extremes of motion occurred at different percentages of the normalized cycle (see Figure 2 and Table 3). It is interesting to note that extension during the propulsive phase was the result of extension at the three joints, and that the hip and knee joints did not move into full extension during the pedal stroke (see Table 3). Furthermore, the patterns and ranges of motion in this investigation were similar to previous investigations (Nordeen and Cavanagh, 1977; Nordeen-Snyder, 1977; Cavanagh and Sanderson, 1986). Finally, the patterns and ranges of velocity were comparable to values presented by Nordeen-Snyder (1977).

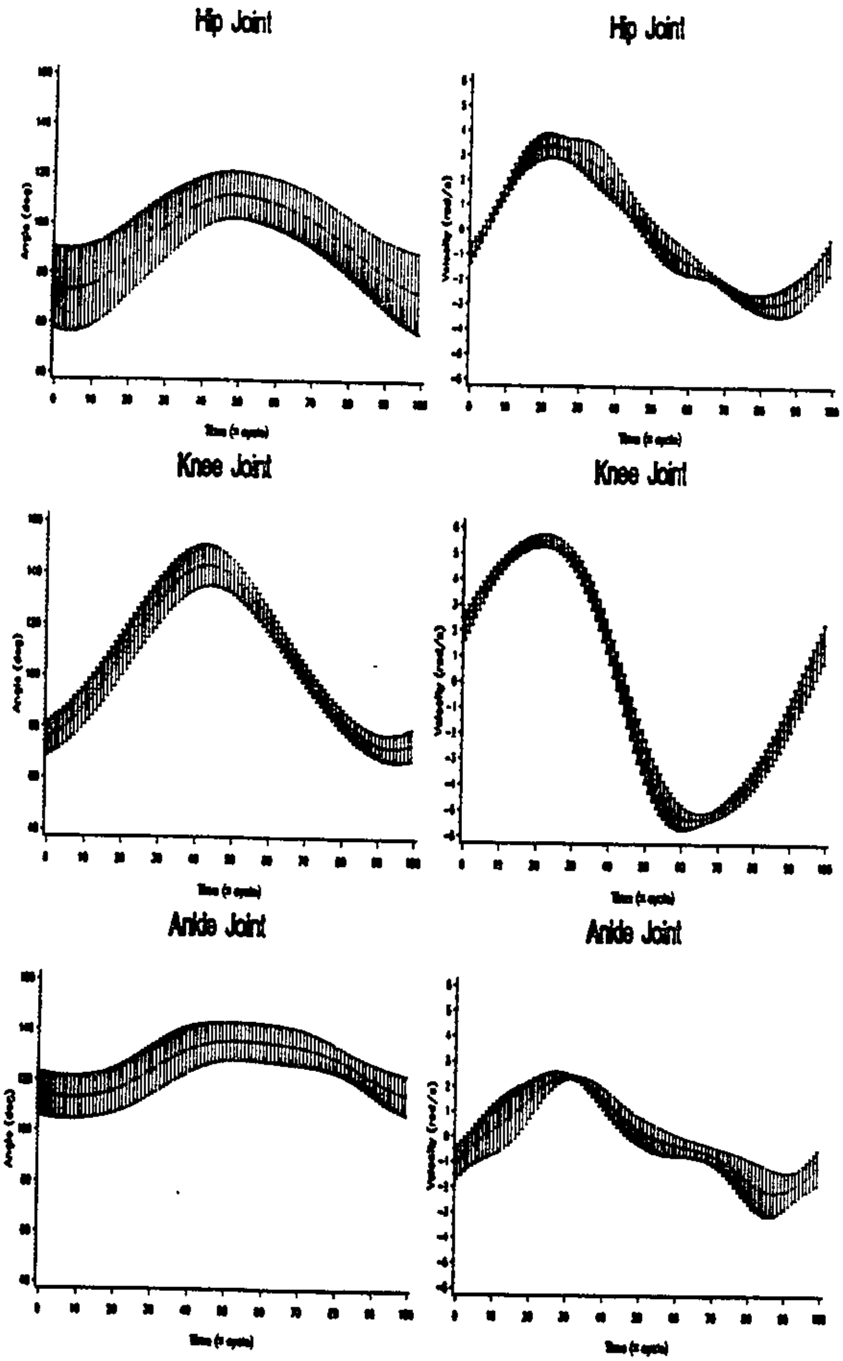


Figure 2 Mean angular displacements and velocities of the hip, knee, and ankle joint joints during the pedal cycle

Table 3

Displacement characteristics of the hip, knee, and ankle joints during cycling

Joint	Range of Motion (deg)	Mean Standard Deviation (deg)	Time to Maximum (% cycle)	Time to Minimum (% cycle)
Ankle	112.98 - 135.72	8.94	9	54
Knee	72.21 - 142.84	24.64	-6	44
Hip	73.05 - 111.66	13.73	4	50

Muscle Kinematics

The mean time-normalized MTU lengths are presented in Figure 3. The abscissae are shown as a function of cycle time. The ordinates present the MTU lengths normalized to the standing anatomical length and scaled to allow maximum resolution of the data points.

The one-joint muscles performed their expected roles. For example, the vastus lateralis, a knee extensor, shortened as the knee extended and lengthened as the knee flexed. In contrast, the biarticular muscles had more complex patterns. The biarticular hamstrings (semimembranosus and semitendinosus) and gastrocnemius lengthened as the crank approached TDC and increased in a relatively linear fashion to reach a peak length prior to BDC (see Figure 3). These MTUs then quickly shortened (see Figure 4) to reach a minimum length at approximately 90% of the pedal cycle. Despite the similar patterns, the ranges and extreme lengths of the MTUs were different (see Tables 4 and 5). The biarticular rectus femoris MTU, on the other hand, shortened as the crank approached TDC and continued to decrease in length until 37% of the pedal cycle. The MTU then increased to reach peak length at approximately RDC.

A comparison of the ranges of the MTUs' lengths revealed that the semitendinosus experienced, on average, the greatest change of length during the pedal cycle (see Table 4). This occurred because the knee joint had the greatest range of motion (ROM) of the three joints examined in this study (see Figure 3) and because semitendinosus attaches at a greater distance from the knee center of rotation than semimembranososus. The MTUs of the semimembranosus and gastrocnemius also experienced larger ranges of motion in comparison to the monoarticular MTUs (with the exception of the gluteus maximus); whereas, the rectus femoris MTU remained near the same length throughout the pedal cycle.

Examination of MTUs' lengths during the pedal cycle in relation to their resting anatomical length revealed that gluteus maximus, semimembranosus, and soleus remained greater than resting anatomical length throughout the pedal cycle; whereas rectus femoris remained less than resting anatomical length throughout the pedal cycle. The remaining MTUs lengths varied about their resting anatomical lengths during the pedal cycle. These results do not support the common belief that interjoint coordination allows the biarticular muscles to function within a smaller range of motion, and closer to their resting anatomical length.

Figure 4 presents the mean time-normalized MTU velocity histories. Table 5 presents descriptive statistics on these histories. Positive velocity, zero velocity, or negative velocity indicate that the MTU was lengthening, remaining the same length, or shortening, respectively.

In comparing the MTU lengths to the results of Gregor et al. (1987) it should be noted that different techniques were used to derive the instantaneous MTU lengths. Despite using different techniques similar patterns were noted in the gastrocnemius and soleus MTU lengths. The absence of the brief stretch/shortening cycle in the gastrocnemius history observed by Gregor et al. and a phase shift in the soleus history were the only major differences.

Table 4

Displacement characteristics of the muscle-tendon units during cycling

Muscle-Tendon Unit	Normalized MTU *				Time	
	Extremes (cm/cm)		Range (cm/cm)	Standard Deviation (cm/cm)	Peak Minimum (% cycle)	Peak Maximum (% cycle)
Gluteus Maximus	1.08	1.23	0.15	0.02	54	12
Semimembranosus	0.99	1.13	0.14	0.04	86	45
Semitendinosus	0.34	0.61	0.27	0.05	91	46
Rectus Femoris	0.93	0.96	0.03	0.02	37	77
Vastus Lateralis	0.96	1.04	0.08	0.03	49	92
Gastrocnemius	0.95	1.04	0.09	0.03	88	46
Soleus	1.09	1.15	0.06	0.04	60	19
Tibialis Anterior	0.91	0.94	0.03	0.02	17	60

* normalized to standing anatomical length

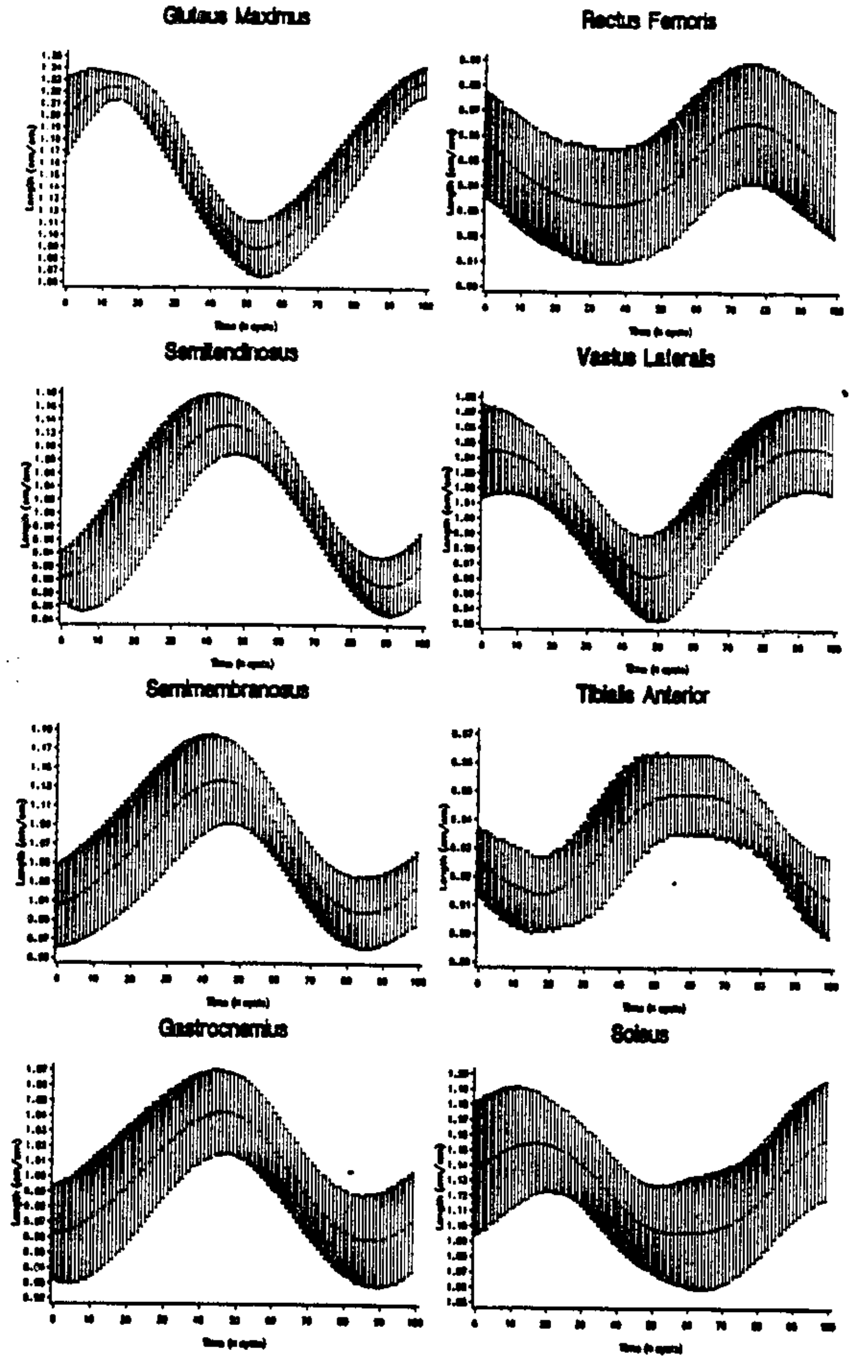


Figure 3 — Mean changes of length of the muscle-tendon units during the pedal cycle

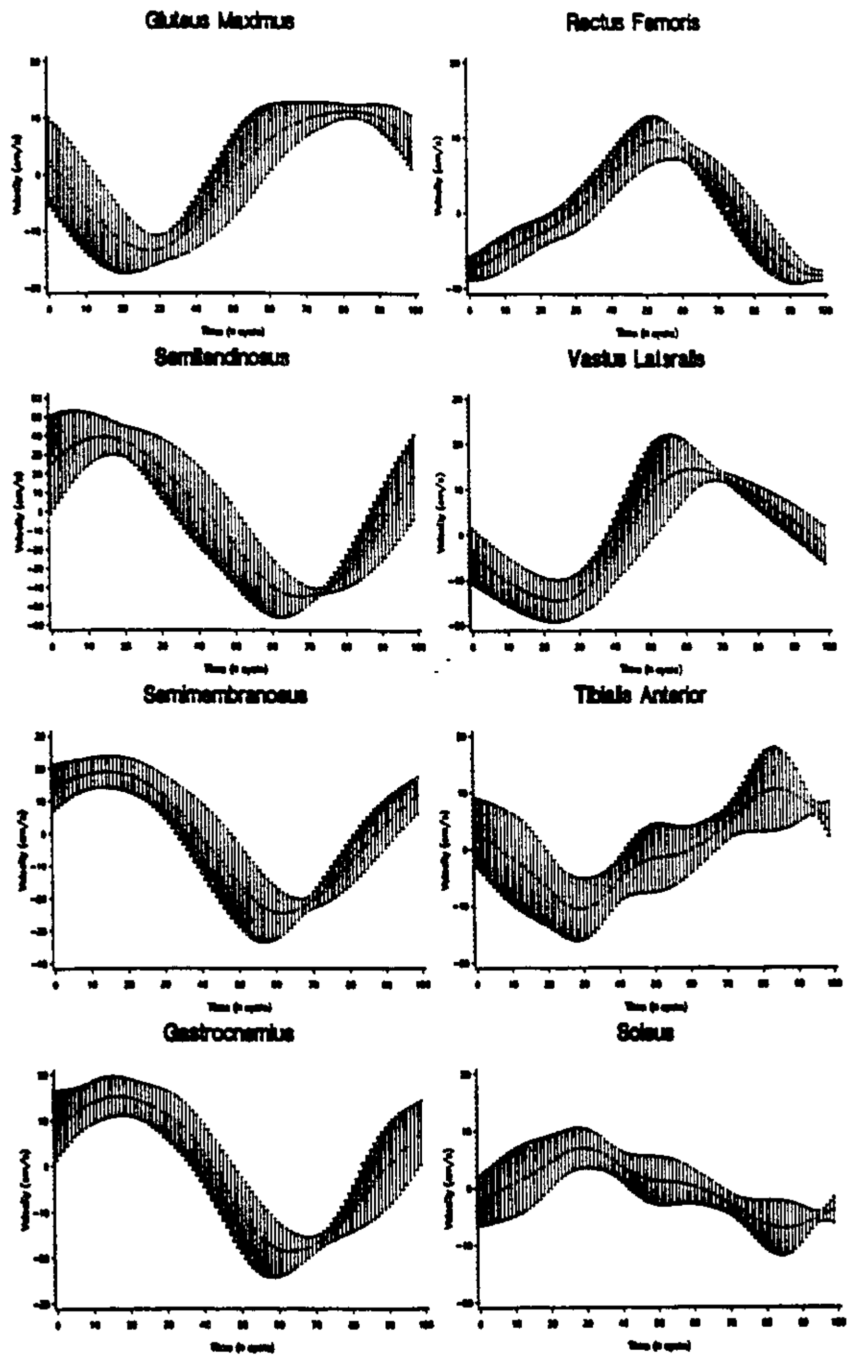


Figure 4 Mean rates of change of length of the muscle-tendon units during the pedal cycle

Table 5

Extreme rates of change of length of the muscle-tendon units during cycling

Muscle-Tendon Unit	Range of Velocity	
	Minimum (cm/s)	Maximum (cm/s)
Gluteus Maximus	-0.15	0.12
Semimembranosus	-0.28	0.20
Semitendinosus	-0.49	0.45
Rectus Femoris	-0.09	0.11
Vastus Lateralis	-0.16	0.17
Gastrocnemius	-0.20	0.16
Soleus	-0.12	0.12
Tibialis Anterior	-0.08	-0.16

Electromyography

Table 6 presents the periods of high muscle activity (greater than 50% MWSA) during the pedal cycle. The grand ensemble averaged LE EMGs are presented in Figure 5. The abscissae are shown as a function of cycle time and the ordinates have been scaled to 100% of the MWSA. For a muscle to achieve 100% on this scale all subjects would need to achieve their maximum activity at the same point in the pedal cycle. Low standard deviations would require that all subjects reach approximately the same level of activity at that cycle time. Notice that in some cases the LE EMG appears to be non-cyclical (i.e., gluteus maximus). This occurred because of variations during the onset and termination of muscle recruitment during the within-subject cycles.

The gluteus maximus became increasingly active prior to TDC and reached peak activity near 12% of the pedal cycle. The activity level decreased shortly after FDC and continued at this low level until RDC.

The hamstring group (semimembranosus and semitendinosus) were also active during the propulsive phase of the pedal cycle. The onset of increased activity in these muscles occurred at a similar times (see Figure 8), but semitendinosus revealed an irregular pattern and reached peak activity before semimembranosus. Low activity in both muscles was observed by RDC.

The activity patterns from the two sites of the quadriceps group (rectus femoris and vastus lateralis) were different. Onset of increased activity for the rectus femoris began prior to RDC and reached a maximum value at TDC. In contrast, the onset of high activity of vastus lateralis began near RDC and was maintained at this level through FDC. These muscles became inactive shortly after FDC, and remained inactive until RDC.

The gastrocnemius and soleus showed their greatest activity within the propulsive phase of the pedal cycle. Both muscles increased activity to reach a peak level following FDC, then gradually declined, and became inactive before RDC. During the propulsive phase the TA activity remained within 20% of its lowest activity level, then increased rapidly during recovery to reach peak activity near 85% of the cycle.

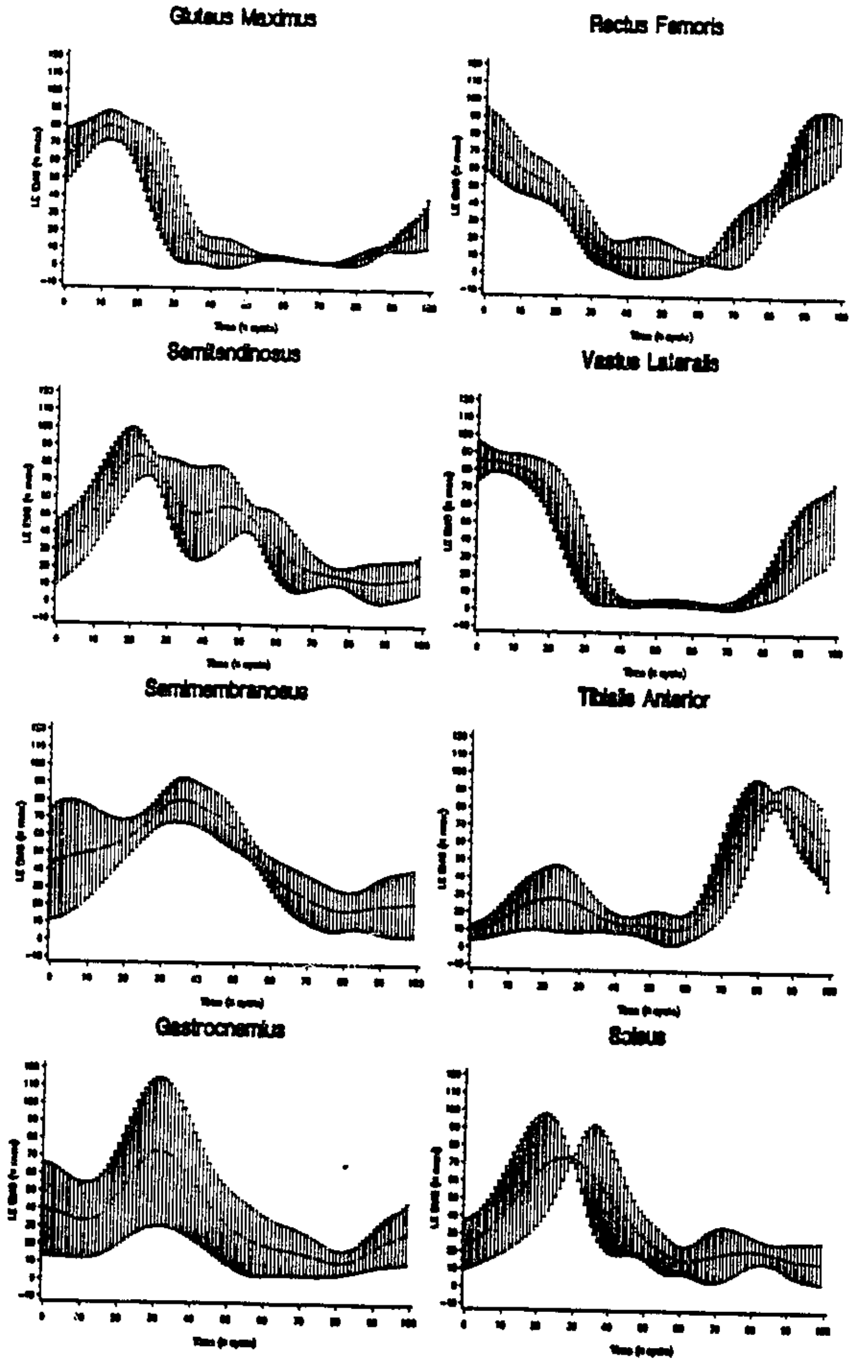


Figure 5 Grand ensemble LE EMG of the lower limb musculature during the pedal cycle

Table 6

LE EMG characteristics during cycling

Muscle	Period of High Activity *			Peak	
	Start (% cycle)	End (% cycle)	Duration (% cycle)	Time (% cycle)	Amplitude (% MWSA**)
Gluteus Maximus	0	24	24	12	79.92
Semitendinosus	10	51	41	22	83.78
Semimembranosus	11	55	44	36	79.82
Rectus Femoris	-16	18	44	0	76.31
Vastus Lateralis	-1	25	26	0	50.26
Gastrocnemius	21	42	21	32	72.54
Soleus	14	40	26	27	73.08
Tibialis Anterior	72	100	28	85	86.37

* LE EMG greater than 50 % maximum within–subject amplitude
** percentage of maximum within–subject amplitude

Figure 6 presents the periods of the LE EMG activity of the muscles during the pedal cycle. The abscissae are shown as a function of cycle time. The black , grey, and white areas represent periods of activity greater than 50%, between 25 and 49%, and less than 25% MWSA, respectively.

Large periods of high co-activation (greater than 50% MWSA) of antagonist muscle groups occurred at the hip and knee joints (see Figure 6). These periods involved monoarticular and biarticular muscles. Unlike the conditions of

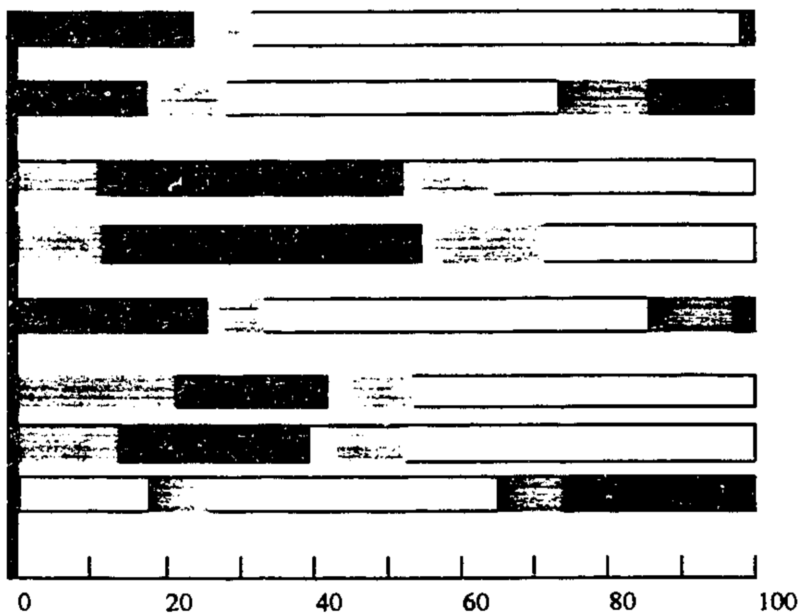


Figure 6 Periods of co-activation during the pedal cycle

Lombard's Paradox (1903), however, there was little co-activation of biarticular antagonists (i.e., rectus femoris and hamstrings (semitendinosus and semimembranosus)). At the ankle joint only small periods of co-activation, at a low level, were observed between antagonist muscles groups. On the other hand, there were large periods of co-activation of agonist groups occurred at the hip, knee, and ankle joints. Additionally, the largest periods of these co-activations occurred in the posterior muscle groups (i.e., semitendinosus and semimembranosus).

A comparison of this investigation's LE EMG activity and several of the previous investigations (Gregor et al., 1982; Gregor et al., 1985; Jorge and Hull, 1986; Gregor et al., 1987) shows discrepancies in the onset and termination of agonist and antagonist muscle activity. The differences are, however, quite reasonable when the variability of the different protocols, the equipment, and the subject population in these investigations are considered.

Classifying Muscle Function During the Pedal Cycle

The original purpose of this study was to classify the functional role of the lower limb musculature during the propulsive phase of the pedal cycle by changes in LE EMG patterns, MTUs' lengths, and segmental kinematics. This classification system defines muscle activity by its LE EMG and defines the contraction as concentric, isometric, or eccentric by changes in its MTU length.

Based on this classification system, the main function of the gluteus maximus was to contribute to hip extension by active shortening during the propulsive phase. To facilitate the function this muscle exhibited a pre-stretch prior to its concentric activity. This muscle functioned according to its expected anatomical role (Rasch and Burke, 1978).

The hamstrings (semitendinosus and semimembranosus) had significant periods of both eccentric and concentric activity. Initially they functioned eccentrically during hip and knee extension. The high rate of lengthening and high LE EMG value suggests the production of a large resistive force. Immediately afterwards they began their concentric activity to assist hip extension and knee flexion, as expected.

In contrast, the quadriceps (rectus femoris and vastus lateralis) were characterized by mainly concentric functions. Vastus lateralis assisted knee extension prior to TDC and maintained this concentric function through FDC. Rectus femoris assisted these functions at the knee, and also assisted in hip flexion and extension. Both muscles showed a short period of eccentric activity (after 30 and 40% of the cycle, respectively) as the hamstring group (semitendinosus and semimembranosus) became increasingly active. These muscles functioned according to their expected anatomical roles.

At the ankle joint, the soleus assisted in the prevention of dorsiflexion by an initial eccentric action and then performed concentrically to assist plantar flexion. The gastrocnemius's initial eccentric activity resisted knee extension and prevented dorsiflexion. This function was followed by a concentric activity that assisted the hamstrings in producing knee flexion and assisted the soleus in producing plantar flexion. These muscles functioned according to their expected anatomical roles (Rasch and Burke, 1978). The tibialis anterior's main function was to assist in dorsiflexing the foot during recovery stroke in preparation for the subsequent cycle by concentric activity, as expected.

It is widely believed that the major difference between walking and cycling is the lack of eccentric contractions during cycling (Gregor et al., 1987). The belief has been disproved by Gregor et al.'s (1987) investigation that showed eccentric contractions in the gastrocnemius MTU. This investigation showed eccentric contractions in all muscles examined.

Comparison of Muscle Function and Segment Kinetics

The consistency of the joint moments reported in several investigations (Gregor, 1976, Gregor et al, 1985; Jorge and Hull, 1986; Redfield and Hull, 1986) during experimental conditions similar to this investigation justifies a comparison to these studies. The ensuing comparison uses the average hip, knee, and ankle net joint moment histories reported by Gregor (1976). These data (see Figure 7) were collected from five subjects during stationary cycling (94 r/min at 140W).

Gregor's (1976) results showed that a hip extensor moment was maintained during the initial three quadrants of the pedal cycle with the peak value following TDC. Moments near zero or small flexor moments were observed in the final quadrant.

This investigation revealed that the hip extensor moment at TDC was the result of the concentric activity of the gluteus maximus overcoming the flexor activity of the rectus femoris (and other hip flexors). When the gluteus maximus became inactive the hip extensor moment was maintained by the concentric action of the semitendinosus and semimembranosus.

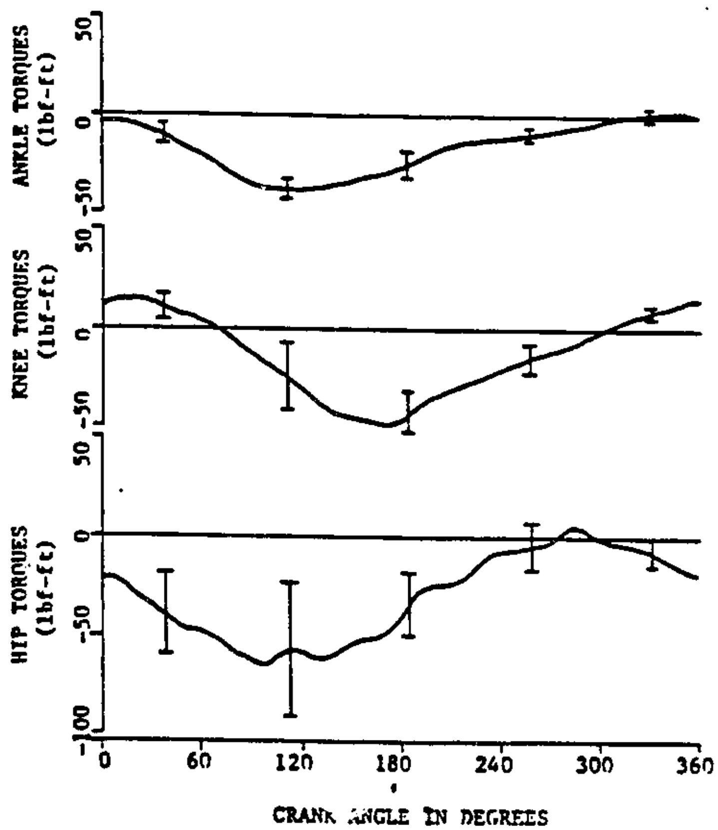


Figure 7 Net joint moments about the hip, knee, and ankle joints during the pedal cycle (from Gregor, 1976)

The knee joint was characterized by an extensor moment with the peak moment occurring prior to TDC. The net joint moment then reversed direction and this flexor moment was maintained until the final quadrant. Gregor et al. (1985) explained that the continued knee extension, in the presence of a flexor moment, during the propulsive phase was the result of the orientation of the lower limb and pedal which caused the resultant pedal reaction force to pass anterior to the knee joint. This force produced an net extensor moment that overcame the flexor moment at the joint to produce knee extension.

This investigation indicated the extensor moment was initially the result of concentric activity of rectus femoris and vastus lateralis (and possibly other knee extensors) overcoming the hamstrings (semitendinosus and semimembranosus) and gastrocnemius (which were lengthening) at the knee joint. When the rectus femoris relaxed, the vastus lateralis maintained the dominant extensor moment until shortly after FDC. Following FDC the hamstring group contributed to the flexor moment at the knee by a concentric contraction.

The ankle joint exhibited an extensor (plantar) moment for the two quadrants of the pedal cycle and remained close to zero for the duration of the pedal cycle with minor fluctuations (dorsiflexor moments) occurring during the final quadrant of the pedal cycle. The plantar flexor moment initially resisted the dorsiflexion caused by the powerful hip moment driving the near vertical shank down on the foot (Gregor,1976) and then caused plantar flexion. This investigation revealed that the plantar moment is the result of the soleus and gastrocnemius activity.

Evaluation of the Models of Biarticular Muscle Function During the Propulsive Phase of the Pedal Cycle

Lombard's Paradox (1903) requires a biarticular muscle to produce a moment that is opposite to the direction of the required moment at one of the joints. Examination of the muscles' functions during the propulsive phase of the pedal cycle shows that both the monoarticular and biarticular muscles, during periods of high activity, functioned concentrically to contribute to the net joint moments. For example, the knee was initally extended by the concentric activity of the vastus lateralis and rectus femoris, and was then flexed by the concentric activity of the semitendinosus, semimembranosus, and gastrocnemius. This investigation, therefore, finds little evidence for the existence of Lombard's Paradox.

Rasch & Burke (1978) hypothesized that co-activated monoarticular and biarticular muscles may act in a " tendinous or belt-like action" to extend a distal joint. This investigation showed, during knee extension, that the quadriceps group overcame the hamstrings (semitendinosus and semimembranosus) and gastrocnemius which were resisting lengthening. These MTUs may have pulled across these remote joints and contributed to extension. It should be noted, however, that Rasch's & Burke's (1978) situation would require an isometric contraction which did not occur.

According to Carlsoo's and Molbech's (1966) adaptation of Molbech's mathematical model (1965) the biarticular hamstring and gastrocnemius function during the propulsive phase of the pedal cycle to extend simultaneously the hip and knee joints. Despite having observed activity in these muscles during this period, changes in MTUs indicated a lengthening (eccentrically activated) contraction. This type of contraction is incapable of performing positive mechanical work. These muscles, therefore, were absorbing energy about the knee, and could not be considered as knee extensors during the propulsive phase of the pedal cycle. The data presented in this investigation, therefore, does not support the application of Molbech's (1965) model to cycling.

Conclusion

Based on the results obtained in this investigation the following conclusions can be drawn. Extension during the propulsive phase of the pedal cycle was the result of periods of high concentric activity of both the monoarticular and biarticular muscles. Furthermore, these muscles functioned according to their expected anatomical roles (Rasch and Burke, 1978). This investigation, therefore, finds little evidence for the existence of paradoxical muscle function as hypothesized by Lombard (1903), Molbech (1965), or Rasch & Burke (1978).

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Appendix B

Review of Literature

Review of Literature

Muscle-Tendon Unit Models

The experimental difficulties associated with the *in vivo* measurement of muscle-tendon unit lengths (MTUs) have resulted in different techniques to approximate the MTUs during dynamic activities. Early techniques to approximate the MTU were based upon mechanical representation of muscle function (Inman 1947; Merchant,1965; Sorbie and Zalter, 1965); wheres, later investigations utilized geometric principles.

Morrison (1970) represented the MTUs of the quadriceps femoris, gastrocnemius, and hamstring group by a straight-line model from the the assumed origin and insertion. To account for the deflection of gastrocnemius muscle over the condyles of the femur and to compensate for the effects induced by patellar movement on the quadriceps MTU several adjustments were made to the basic model. During activity, the model approximated changes in the MTUs by the product of the angular displacement and the perpendicular distance from the joint center to the line of action. Morrison's decision to define the position of the hamstring group by a single point origin and insertion has limited the interpretation of the functional role of the individual MTU in this group. Morrison did, however, use this model to examine the muscle function during level and graded walking, as well as stair climbing.

Seireg and Arvikar (1973) used a straight-line model to represent 29 muscles of the human lower limb muscles in the sagittal plane. Some muscles in this model were represented by more than one line-of-action to account for their complex geometry (i.e. muscles crossing the ankle joint were modelled as two adjoining lines). The authors claimed that this model provided a better representation of the MTU than previous techniques, yet the calculations on origins and insertions were obtained from two-dimensional skeleton illustrations. These data (origins and insertions) must be considered a limitation of the technique.

Realizing the shortcomings in the previous models, Frigo and Pedotti (1978) presented equations to approximate eleven MTUs as the summation of linear and/or curvilinear sections, and derived the origins and insertions from dry skeleton data. The lower limb was represented by a three-link system in which muscle lengths were trigonometrically expressed as a function of the associated joint's relative angle (hip, knee, and ankle). The instantaneous MTUs could were calculated by entering the relative angular orientations of the joints into the model. In an effort to improve of Frigo's and Pedotti's (1978) model, Hubley (1981) used the data of five male cadavers to develop a technique to customize the equations to standing anatomical height. More recently, MacKinnon (1988) has adjusted the equations of the soleus and gastrocnemius to account for individual characteristics of the leg.

An alternate method of estimating MTU was presented by Grieve, Pheasant and Cavanagh (1978). This technique was based upon sectioned cadaver limbs throughout a range of knee and ankle excursions and used a least squares polynomial regression technique to predict the instantaneous length of the gastrocnemius during activity. The authors used this technique to predict the MTU during walking.

While many investigators had used the straight-line models, Jensen and Davey (1975) suggested that a centroidal-line model provided a more valid measurement of MTU because it accounted for the curved paths that muscles generally follow. The centroidal-line model divides the muscle into successive cross-sections and the muscle's line of force through the centroid of each cross-section. A comparative investigation of the gluteus medius, rectus femoris and sartoris muscles revealed that the centroidal-line representations were superior to the straight-line representations. Andrews and Hay (1983) have stated that this technique could produce unreasonable results because of difficulties in clearly defining a truly transverse cross section, and because it was based on cadaver data where lack of muscle tone may have been significantly different than *in vivo* measurements. This coupled with the difficulty in implementing the centroidal approach has caused in several researchers to question its usage (Brand, Crowninshield, Wittstock, Pedersen, & Clark, 1982; Pierrynowski, 1982).

To improve the representation of MTUs, Seireg and Arvikar (1975) modified their previous straight-line model of the lower limb (Seirig and Arvikar, 1973) to include three dimensional origins and insertion of 31 muscles. Deviations in the muscle path created by other musculature and bony prominences were accounted for in this model. Again the accuracy of this method was limited by the origin and insertion coordinates obtained from illustrations.

Brand et al. (1982) used radiographic negatives to develop a systematic means of determining three-dimensional origins and insertions of 47 lower limb and pelvic muscles. This technique used selected anthropometric points measured from the individual, and determined the locations of these points and the MTUs through mathematical transformations. Reference systems, defined from radiographically identifiable landmarks, were located on each segment and these systems permitted the calculation of muscle length given any orientation of the segment. A sensitivity analysis determined the magnitude of difference in the computation of the length of the moment arm of these muscles at the joints when the markers were moved in (± 10 mm) the x, y and z direction. The results indicated that differences associated with these three variations were generally less than 10%, yet one case at the ankle exceeded 20%, and the maximum difference reported was 34%. Herzog (1985) has shown that the moment arm variation of up to 50% could occur in certain situations. Despite these errors, the authors have stated that this technique was a suitable method for approximating the muscle origin, insertions, and moment arms in living subjects.

Mansour and Pereira (1987) later presented a technique to allow the transformation of the above data points to a subject during normal locomotion. This technique, while it could be applied to most activities, was limited by the unusual axes and the number of transformations required.

Pierrynowski (1982) presented an alternate technique for measuring lower limb MTUs. His method, characterized each segment as a rigid member with a unique skeletal reference system and based the muscle origins and insertions on several palpable boney landmarks in these reference systems. Coordinate data were obtained from a single disarticulated cadaver. The muscles were represented as an elastic thread connected from the centriod of its area of origin to the centriod of its area of insertion. Between the two end-points, up to four additional points were defined from anatomical considerations, through which the muscle was constrained to pass. The line-of-action was obtained by joining the defined points with a combination of straight and curvilinear sections and the summation of these sections determined the muscle's length . All ligaments were defined as straight-line sections between their areas of origin and insertion. To correct for the underestimation of moment arms caused by one muscle overlying another Pierrynowski used an atlas of anatomy and defined the line-of-action in the transverse plane centered at the hip, knee and ankle joints. The patella was not included in the model and a method to adjust muscles that attached to this boney structure was presented. While this technique allowed muscles to be defined relative to palpable points, specialized equipment, and a long anthropometic measuring session were required to implement the technique. These difficulties have limited the application of this technique.

In summary, the complexity of the musculo-skeletal system has resulted in several representations of the MTU. These geometrical models include both straight-line models, linear/curvilinear-line models, and centroidal-line models, and models in two- and three-dimensions. While the centroidal-line models and three-dimensional models allow for greater accuracy, they require more complex data acquisition and reduction. These representations, therefore, are limited as investigators must maintain a balance between accuracy and simplicity. Furthermore, Grieve et al. (1978) has stated the investigator must realize these models do not directly measure the change of length of the muscle fibers and indicate only a relative measure of these values. The above rational has prompted Frigo and Pedotti (1978) to consider the assumptions made in their simplified model to be acceptable, despite the criticisms lodged against the straight-line models. This statement is supported as numerous investigators have utilized their equations to analyze various movements with apparent success (Hubley, 1981; Curwin, 1984; Simonsen, Thomsen and Klausen, 1985; MacKinnon, 1988; Wilson, 1989).

Muscle-Tendon Unit Models and Their Application to Cycling

Few papers to date have approximated the changes in length of the MTU during cycling. Gregor, Komi, and Jarvinen (1987) described the changes in the lengths of the MTUs of the gastrocnemius and the soleus of a single subject at various workloads and cadences during cycling on a stationary ergometer. The results indicated that the soleus MTU decreased in length during the power stroke of cycling, whereas the gastrocnemius MTU experienced an increase in length during the first quadrant of the pedal stroke. Furthermore, the gastrocnemius reached peak activity during this lengthening phase. The authors suggested this indicated the presence of a stretch-shortening cycle during the power stroke, and stated that this stretch-shortening cycle resulted in increased force production during the second quadrant of the cycle. From these data the authors concluded that previous assumptions that cycling lacked any lengthening contractions were incorrect.

Redfield and Hull (1986) performed an optimization model that required geometric modelling of 13 muscles of the lower limb during cycling. Despite rigorous modelling of the moment arm lengths, the cross-sectional areas of the muscle, and the MTUs of the lower limb, the authors chose not to present this data.

In view of the limited attempts to understand the changes in the MTU during cycling it is appropriate to use empirical data from an accepted technique. For this reason many investigators (Hubley, 1981; Curwin, 1984; Simonsen, Thomsen and Klausen, 1985; MacKinnon, 1988; Wilson, 1989) have implemented Frigo's and Pedotti's model (1978).

EMG Techniques and Application

The electrical signal associated with the contraction of a muscle is called an electromyogram (EMG). The electromyogram can be recorded by surface electrodes or indwelling electrodes (Winter, 1984). Early electromyographic studies generally used intrinsic fine wire electrodes; whereas, most investigations in the past decade have used bipolar surface electrodes. The increased usage of the bipolar surface electrodes has prompted several investigators to examine the reliability of electromyograms taken from bipolar surface electrodes (Komi and Buskirk, 1970; Yang and Winter,1983), and this has resulted in the development of many techniques to capture and process the raw EMG activity. One such technique, the full-wave rectified and low pass filtered signal (linear envelope (LE)), has been recommended by Winter (1979) because it follows the trend and closely resembles the shape of the force curve providing an analog pattern that is reliable and reproducible (Winter, 1984). Additionally, Yang and Winter (1984) have suggested EMG reliability can be improved if all treatments should be completed within one day with no electrode manipulation and several trials per condition (a minimum of three) are used in calculation of an average signal.

Yang and Winter (1984) evaluated four EMG normalization methods on the inter-subject variability of electromyographic profiles in normal gait. Their results showed that the smallest variability was produced by averaging to the maximum of the subject ensemble average and to the mean of the subject ensemble average. Yang and Winter (1984) also recommended that normalization of the linear envelope signal to either the peak ensemble or the mean ensemble average for detecting phasic abnormality between different types of contraction actions.

Although the relationship between significant muscle recruitment as approximated form the EMG and force production is not completely understood, the EMG continues to be utilized in the estimation of muscle force production. Additionally, Crowninshield and Brand (1981) have stated that the EMG can be used in the direct assessment of muscle force, as a measure of temporal constraint, and for the temporal validation of muscle force. Finally, MacKinnon (1988) has compared the LE EMG to changes in the MTU to determine if a muscle is acting concentrically, eccentrically, or isometrically.

EMG Activity During Cycling

A classic investigation by Houtz and Fisher (1959) examined the electrical activity of the fourteen muscles (lower limb and upper body) of three subjects pedalling at very low cadences on a stationary ergometer at two saddle heights (21 and 25 inches). The obtained consistent patterns across subjects that were resistant to alterations in the workload and saddle height. While the patterns of activity were not affected, increased workload and increased saddle height resulted in increased and decreased magnitude of the lower limb muscle activity, respectively. In general, random EMG activity of the upper body muscles and co-activation of agonist and antagonist muscles of the thigh during the power stroke of cycling were present. The lack of detail on the processing techniques utilized, however, has limited the interpretation of the muscle function.

Carlsoo and Molbech (1966) investigated the muscles of the lower limb of five non-cyclists during stationary cycling at a single workload and low cadence. The activity of six muscles was examined by surface and indwelling electrodes. Additionally, the maximal activities were calculated, and these values were related to the pedal position. The results of this investigation were similar to Houtz and Fisher (1959). Again the EMG activity was described only as active or inactive (on/off). The problems associated with this form of analysis can lead to inaccurate conclusions.

Despirés (1974) studied the effect of saddle height and load pattern during treadmill cycling (27 km/h) rather than using a restrictive stationary ergometer. Three elite cyclists rode a racing bicycle on three inclines (0, 2 & 4 degrees) at two saddle heights (95 and 105% of pubis symphysis height) at a racing cadence and workload (250 W at a 90 r/min). The electrical activity of twelve muscles was sampled for five seconds on a pen recorder during all trials and integrated.

In addition to the expected random activity of the rectus abdominis and the erector spinae muscles (Houtz & Fisher, 1959), the activity of pectoralis major and the flexor digitrorum superficialis were found to be variable and minimal. The EMG activity of lower limb muscles followed a consistent pattern, and was characterized by co-activation of agonist and antagonist muscle groups of the thigh. An increase in saddle height produced an increase in the total duration of activity (altered the on/off pattern), yet did not increase the magnitude of the activity. This was in direct contrast to the results of Houtz and Fisher (1959). An increased incline generally produced greater magnitudes of the EMG activity, yet did not alter the pattern of activity (altered the on-off pattern). The muscle activity in this study was presented as an absolute value (quantified as a single number) over the entire cycle of activity and not as the relative value of activity at discrete parts of the cycle. This provided only a qualitative analysis throughout the pedal cycle and therefore has limited interpretations. For instance, a large IEMG value could represent a high level of activity through ten degrees with little or no activity throughout the remainder of the cycle, and visual inspection of the on-off activity in the polar plots and a high IEMG value would incorrectly suggest the muscle is highly active throughout the pedal cycle. A value of the EMG activity at discrete intervals would enable a better comparison of muscular activity to the kinematics and kinetics of cycling.

Jorge and Hull (1986) utilized such a semi-quantitative processing technique to examine the EMG activity of six cyclists (tourist, recreational and racing) during simulated cycling at two saddle heights, several workloads, and shoe-pedal interface conditions on stationary rollers. The EMG activity of eight muscles of one lower limb was recorded for two trials of one revolution each. Only four muscles were sampled at a time due to a limited number of amplifiers. The raw EMG signals were divided into ten equal sections (10% of the cycle (.75 ms intervals)), full wave rectified and integrated. The interval with the greatest activity was considered 100% and values were expressed relative to this maximum. The normalized intervals were then ensemble averaged to produce a mean response for all riders.

Upon examination of polar plots of on/off activity the authors indicated that several differences existed between this investigation and previous studies, and that periods of co-activation of agonist and antagonist muscle groups of the thigh existed during the power stroke. In comparison to previous investigations, the authors arbitrarily set the criteria to determine an "active muscle" by a level of activity of greater than or equal to 50% the normalized LE EMG. From these data the authors concluded that insignificant periods of co-activation existed between the agonist and antagonist muscles of the lower limb throughout the pedal cycle. Furthermore, these periods of activity remained consistent within subjects, but varied among subjects.

The EMG activity remained phasically similar with increasing workloads, whereas the amplitude increased. Reduced saddle height resulted in increased muscle activity of the quadriceps and hamstring groups. This finding disagreed with a previous EMG investigation (Despirés, 1974), yet supported Houtz's and Fisher's (1959) results .

Gregor, Green and Garhammer (1982) investigated ten competitive male cyclists riding at a constant cadence (85 r/min) and moderate workload (170 w) on a modified laboratory ergometer (the rear wheel was unconstrained) that was adjusted to the subject's 'regular' riding position. The EMG activity of eight muscles on both legs was sampled for one minute during two test periods. The EMG signal processing technique was similar to Jorge's and Hull's (1986) investigation. The results of this investigation revealed similar patterns of muscle activity within subjects, and similar patterns of muscle activity for both the right and left legs; whereas, the absolute magnitudes varied from day to day, and from subject to subject. The patterns were similar to the results of Jorge's and Hull's (1986) investigation with minor differences in amplitudes and temporal phasing. In contrast to the results of Jorge and Hull's investigation (1986), significant periods of co-activation of the agonist and antagonist muscles of the thigh were shown to occur throughout the pedal cycle (when using activity greater than 50% as the criterion).

Gregor, Cavanagh and Lafortune (1985) examined the EMG response of semimembranosus, biceps femoris, rectus femoris and vastus lateralis for four recreational cyclists during a four minute interval of cycling simulated on a roller-resistance system at a moderate workload (160 W at 60 r/min). The raw EMG activity was full wave rectified, integrated over fifteen degree sections, and expressed as a percentage of the maximal interval. In contrast the findings of Jorge and Hull, the results indicated that co-activation always occurred between agonist and antagonist groups of the thigh.

The above investigations have revealed several discrepancies during cycling in the activity periods of the lower limb muscles. However, the observation of large periods of co-activation of the quadriceps and hamstrings during the power phase were common to all studies with the exception of Jorge and Hull's (1986) investigation.

Joint Moments During Cycling

Gregor (1976) investigated the net joint moment of force patterns about the joints of the lower limb during stationary cycling at four workloads. The subjects performed three workloads that were based on their physiological capabilities, and a forth trial at 1.5 kp at 94 r/min. The final trial was time normalized and averaged to attain a mean response of the subjects.

Gregor's results (1976) revealed that a hip extensor moment was maintained during the initial three quadrants of the pedal cycle with the peak value following TDC. Moments near zero or small flexor moments were observed in the final quadrant. The knee joint was characterized by an extensor moment with the peak moment occurring prior to TDC. The net joint moment then reversed direction and reached a peak flexor moment shortly before BDC. This flexor moment was maintained until the final quadrant. Gregor et al. (1985) explained that the continued knee extension, in the presence of a flexor moment, during the propulsive phase was the result of the orientation of the lower limb and pedal which caused the resultant pedal reaction force to pass anterior to the knee joint. This force produced an net extensor moment that overcame the flexor moment at the joint to produce knee extension. The ankle joint exhibited an extensor (plantar) moment for the first 200 degrees of the pedal cycle and remained close to zero for the duration of the pedal cycle with minor fluctuations (dorsiflexor moments) occurring during the final quadrant of the pedal cycle.

Examination of the individual subjects at this workload indicated that all subjects followed similar patterns with minor differences in magnitude and temporal phasing of peak values. The author attributed these differences to variations in cycling ability among subjects. Additionally, the trials at different workloads indicated similar patterns while increases in the magnitude of the moments occurred with increasing workloads.

Jorge and Hull (1985) examined three subjects of different abilities at several workloads and various combinations of pedal reaction forces and cadences. The pedal angle was modelled by a sine function while the joint kinematics were derived from a link-segment model (displacement). Using the tangential force, normal force, subject measurements, and anthropometric data tables the net joint moments of force were calculated. The net joint moments of force were subsequently divided into kinematic moments and static moments. The kinematic moments were due to motion, whereas the static moments were due to pedal reaction forces. This division was made to determine if the muscle functioned to drive the leg and/or to generate pedal forces, and to allow a comparison of the kinematic and static moments.

The net joint moments were found to be the result of a complex relationship between the pedal reaction forces and cadence. The kinematic action of the limbs contributed significantly to the net joint moments at higher pedalling rates; whereas, the pedal reaction forces controlled the net joint moment at lower cadences. Furthermore, the hip moment showed minor changes in the temporal pattern with increased power at constant cadence and with increased cadence at a constant workload. In general, the net resultant moments had different magnitudes about the hip, knee and ankle, yet had similar patterns to the earlier investigations (Gregor, 1976; Gregor et al., 1985). Finally, the authors hypothesized that an optimal cadence would exist where the individual components (described above) would produce average minimum joint moments at the three joints. The authors did not investigate this possibility.

Redfield and Hull (1986) examined the possibility of an optimal cadence as determined by the minimum absolute hip and knee resultant joint moments during cycling. One subject was examined at a single workload (196 W) and three cadences (63, 80, and 100 r/min). The net resultant joint moments were similar to a previous investigation (Hull and Jorge, 1985). Furthermore, the results indicated that an optimal cadence (based on the above criteria) could be determined for any given power level and bicycle-rider geometry, and that the optimal cadences for the conditions in this test were between 95 -105 r/min.

In summary, the investigations of the net joint moments of force during stationary cycling have revealed consistent patterns about all joints of the lower limbs. In general, the hip and ankle joint are characterized by extensor moments throughout the cycle; whereas, the knee revealed large periods of extensor and flexor activity during the pedal cycle.

Explanations of Biarticular Muscle Function

The earliest hypothesis to address the functional aspects of biarticular muscles was proposed by Lombard (1903), hence the name Lombard's Paradox. From the examination of frog limbs, Lombard stated that the functional role of a two-joint muscle was dependant on the magnitude of the moment arm at each joint and the activity of other muscles that cross the joint. Furthermore, he hypothesized that a biarticular muscle could behave like a single-joint muscle when one end was constrained by an external force; whereas, a different situation occurred when both ends were free to move. In the latter case, he hypothesized that a closed path of energy exchange existed between two-joint muscles during extension of both joints and that the flow of energy within this path (a figure eight pattern) was transmitted in an endless chain, with the energy progressing in the direction of better leverage. Three conditions were outlined by Lombard (1903) for a muscle to extend a joint which it can flex. These included the following: (a) it must have a better leverage at the end by which it acts as an extensor, (b) there must be a two joint muscle that flexes the joint which the muscle in question extends and extends the joint which it flexes, and (c) the muscle must have sufficient leverage and strength to make use of the passive tendon action of the other muscle.

Lombard stated that these principles were applicable to the biarticular muscles of the lower limb in man. He hypothesized that the simultaneous hip and knee extension observed during walking was the result of simultaneous co-activation of the hamstring and quadriceps groups.

While Lombard validated his theory by examination of the biarticular muscles of a frog (Lombard & Abbot, 1907), no study to date has validated this theory *in vivo* in humans. This is unfortunate as Andrews (1987) has suggested a systematic means of determining the *in vivo* existence of Lombard's Paradox (1903) in humans. Additionally, Andrews classified all techniques based on moment dependent explanations of muscle function (i.e., Lombard's Paradox) as the"Standard Kinetic (SK) technique".

Rasch and Burke (1978) examined Lombard 's Paradox (1903) utilizing simple models based on anatomical considerations. Their results supported the conditions that Lombard (1904) hypothesized to produce extension. In addition, Rasch and Burke

(1977) expanded the analysis to include a situation where one muscle maintained a "belt-like or tendinous "action. The results showed that extension or flexion would occur if either the agonist or antagonist were passive. It must be noted that the results of this simplistic model were obtained with uncontrolled 'muscle model forces', an assumption of constant moment arm lengths (which is different than the human joints) and a limited number of modelled muscles. It is, therefore, limited in providing empirical proof of Lombard's Paradox (1903) *in vivo* in humans. As suggest by Lombard (1903), and Andrews (1987), an examination of moment arms and muscle forces *in vivo* is needed to verify this model.

Molbech (1965) proposed an alternate theory to explain the functional role of two-joint muscles. Molbech (1965) stated that certain two joint muscles in a closed kinematic chain (steered movements) deviate from their widely accepted role as flexors and act as extensors. In this model, Molbech (1965) classified the functional role of a muscle based on its moment about the instantaneous center of rotation and the associated change in the configuration of the particular joint traversed by the muscle.

Molbech presented a description of the two-dimensional movement of a two-bar linkage (see Figure 8). The lower rod (representing the shank) was fixed by a single hinge joint at point A and point K; whereas, the upper bar (representing the thigh) was restricted to movements along the y axis (slider joint). Both rods were of equal length. The line of the muscle force affecting the upper segment was between the muscle origin (C') and insertion (C). The system's instantaneous center of rotation (point O) was located at the intersection of a horizontal line through H and the extension of AK through K. Therefore, any force acting in this configuration at O and perpendicular to the arc would have no effect in displacing point K because movements were restricted to a tangent to the radius from point O. Molbech referred to this line as "limiting line" of the system. If the direction of force acting on C was directed towards a point C' beyond L on the x-axis, the rod would move closer to C', the angle HKA would increase, and raise point H. In comparison, a force on the other side of the limiting line would reduce the angle AKH and move the point H down the rod. This muscle force could therefore act as an extensor and a flexor, and the change in function occurred at the limiting line (computed mathematically near 135 degrees). Molbech also stated that the system could be adapted to examine the functional role of any biarticular muscle.

Andrews and Hay (1983) noted several shortcomings in the mathematical model presented by Molbech (1965). These included: the lack of instruction indicating whether the segment on which the muscle force acted was the one where the muscle originated, or the one where it inserted; the classification procedure allowed no way to determine, *a priori*, what reference frame was to be used when establishing the location of the instantaneous center of rotation for any body segment; and the classification procedure did not indicate which of the instantaneous centers of rotation to choose during horizontal curvilinear motion.

Andrews (1985) presented a new classification scheme based on the techniques of equilibrium analysis. The classification of muscle function by this method was based on the assumption that the linkage system would always tend to move away from the initial rest configuration, in such a way as to decrease the length of the muscle in question. The muscle would act to bring its attachment points closer together when the system is released from its initial rest configuration. This method, therefore, established the functional role of a muscle based on the algebraic signs of the first partial derivatives of the muscle length taken with respect to the systems independent generalized coordinates, and evaluated in the configuration of interest.

For a one degree of freedom system such as a vertical lift (see Figure 9), these partial derivatives collapsed to a single total first derivative which was simply the slope of the curve representing the variations in the muscle's length plotted as a function of the system's generalized coordinate Ø . The angle Ø was then used to determine the configurations of the ankle, knee, and hip joints. If a decrease in Ø corresponded to joint flexion and if the slope of this curve was positive, then a shortening of L could only coincide with a decrease in Ø indicating that the muscle was a flexing agent of Ø. When the slope was negative, a decrease in Ø could only create an increase in Ø, therefore it acted as an extending agent of Ø. When the slope of this curve was zero, muscle functional role was indeterminant.

The author acknowledged that the results of such an analysis were limited to the confines of the model. Other limitations, the author did not address, included the model's inability to account for physiological factors such as increased resistance to the movement or the speed of the movement, and the difficulty to utilize the classification of a muscle's functional role, as defined in this model, in a clinical setting.

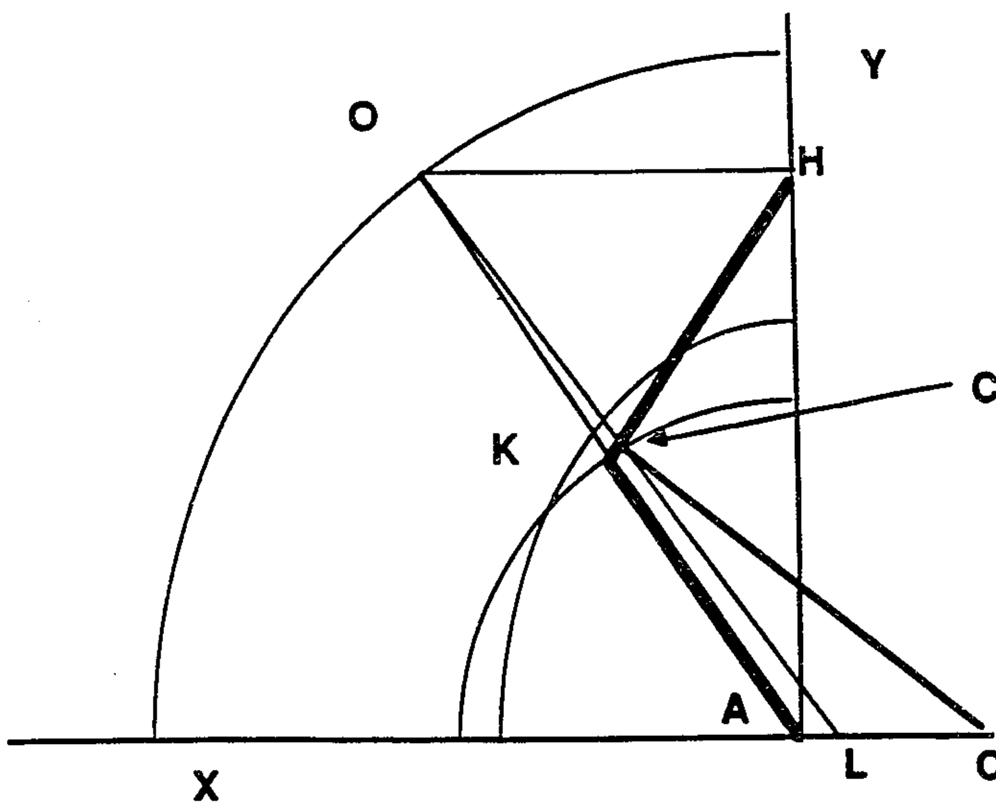
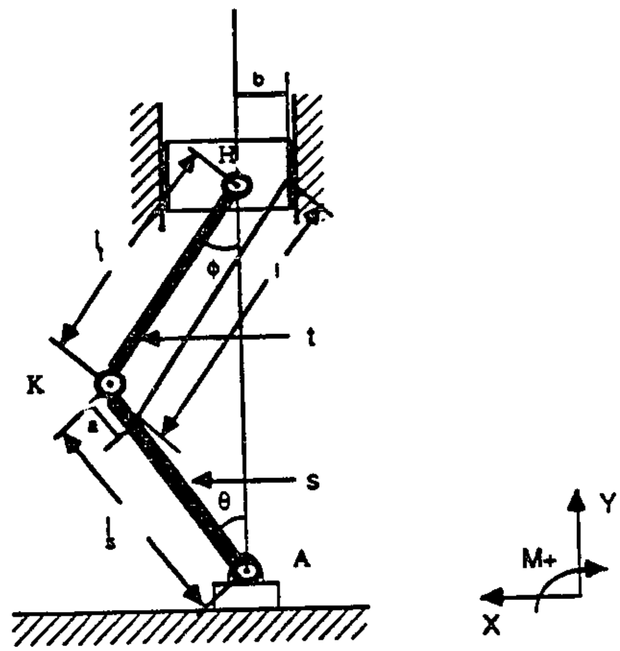


Figure 8 Molbech's mathematical model (from Molbech, 1965)



$$l^2 = a^2 + b^2 + l_t^2 + 2b(\,l_s - a\,)\sin(\theta) + 2al_t\cos(\theta + \phi)$$

$$l_m = l / l_t \;;\; \alpha = a / l_t \;;\; \beta = b / l_t \;;\; \gamma = l_s / l_t$$

$$\partial l_m^{\,2} / \partial\theta^2 = 2\beta\,(\gamma - \alpha)\cos(\theta) - 2\alpha[\,1 + (d\phi/d\theta\,)]\,\sin(\theta + \phi)$$

<u>Figure 9</u> Andrews's anatomical representation (from Andrews, 1985)

Explanations of Biarticular Muscle Function During Cycling

Gregor et al. (1985) examined the EMG activity and net joint moments of force during stationary cycling in an attempt to verify the existence of Lombard's Paradox during cycling. Upon examination of the results the authors concluded that the hamstrings and the quadriceps muscles functioned non-paradoxically (according to Lombard (1903) or Andrew' s (1975) SK technique)) at both the hip and knee, and avoided "Lombard's Paradox".

Redfield and Hull (1985) examined three male subjects during simulated cycling on rollers. Their results showed no muscle co-activation. This was in agreement with a previous EMG investigation (Jorge and Hull, 1986). It was concluded from these data that muscles of the lower limb functioned according to their expected anatomical role.

Carlsoo and Molbech (1966) investigated the functional role of the biarticular muscles of the lower limb of five non-cyclists during stationary cycling as a means of validating the Molbech's (1965) anatomical model . To simplify the analysis the ankle joint was assumed to follow a circular path, and the hip joint and the ischial tuberosity were considered to remain in a stationary position (see Figure 10). Using this model and the results of the EMG activity the authors determined that the hamstring activity could be divided into three phases: the initial phase (starting after passing TDC position), when the main activity of the bicep femoris was to counteract the hip flexors while providing knee flexion (although knee extension had begun); the second phase, when the muscle changes the knee joint movement from flexion to extension (limiting line); and the third phase, when the knee bends as in a free hanging limb. In conclusion, the authors stated that the hamstrings could assist in flexion and extension.

Andrews (1987) later examined the functional role of the biarticular muscles by the 'Andrews Kinematic Method (AK)' (Andrews, 1985) and compared the results to the SK method. Data for the SK method came from Gregor et al. (1986). To complete the AK analysis the bicycle was modelled as a one degree of freedom, four bar planar system and the generalized coordinate was the crank angle. The seat-tube axis was inclined to 20 degrees to the vertical axis. The hamstring muscle was represented as a straight-line segment and the quadriceps line of action was represented by a straight-line segment and an adjoining circular arc (see Figure 11). The limb lengths were derived from Dempster, and the muscle origins and insertions were points scaled to conform with those of a related study by Stanhope (1982). Muscle activity was determined from Gregor et al. (1985).

The results of the SK and AK techniques indicated that the hamstrings and quadriceps functioned both paradoxically and non-paradoxically at the knee and hip joint. Furthermore, the SK method indicated that the hamstrings functioned non-paradoxically at the hip, with both muscle groups exhibiting approximately equal regions of paradoxical and non-paradoxical behavior at the knee joint. In contrast, the AK method indicated that the quadriceps generally functioned non-paradoxically at both the hip and knee; whereas, the hamstrings generally acted paradoxically at the knee and exhibited essentially equal regions of paradoxical and non-paradoxical behavior at the hip. This difference was expected as the nature of the paradox in each technique was different. Finally, it must be noted that Andrews model has not been validated *in vivo* in humans

Andrews concluded that the manner in which one determines paradoxical/ non-paradoxical biarticular muscle function is affected by the method used to determine when muscle is active, the criterion used to establish the existence of paradoxical behavior, and the classification method used to classify the functional role of the muscle. Furthermore, Andrews stated that the least significant of the three was the manner in which muscle activity was determined.

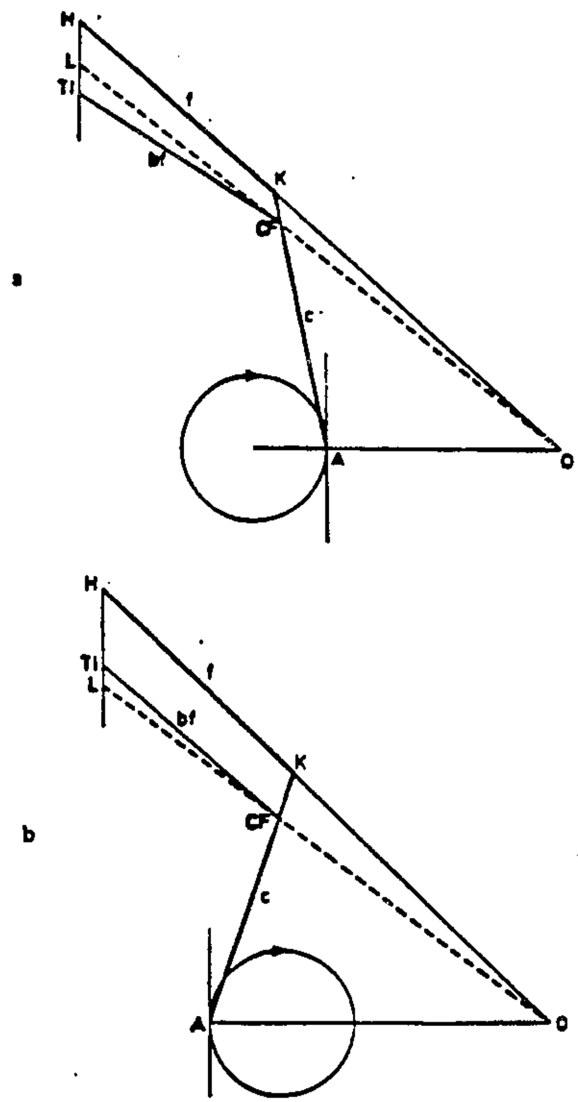


Figure 10 Carlsoo's and Molbech's explanation of cycling mechanics (from Carlsoo and Molbech, 1966)

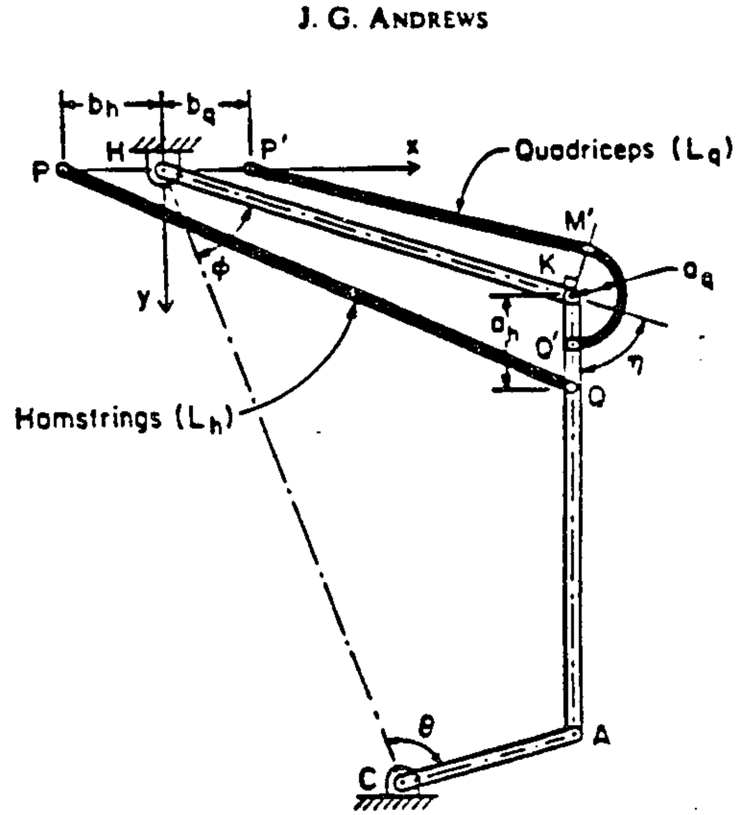


Figure 11 Andrews's explanation of cycling mechanics (from Andrews, 1987)

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