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LOWER LIMB MUSCLE FUNCTION DURING DEEP-KNEE BENDING

by

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To Geneviève, Gilbert and Dan, without whom this work would not have been possible. I would like to thank Mr. Scott MacKinnon and Mrs. Cheryl Kozey for their aid in the development of the muscle-tendon-unit software used in this work and Mr. Ralph D. Fournier for the technical support he has provided during this study. Further thanks is extended to the Ontario Graduate Studies Scholarship Fund and " Le Fonds pour La Formation De Chercheurs Et L'Aide à La Recherche " for the financial support provided throughout my graduate studies.

Abstract

To understand the functions of lower limb muscles during simultaneous hip, knee, and ankle extension, electromyographic, kinetic, and muscle length data were collected from experienced subjects performing weighted and unweighted deep-knee bending movements. Functions of the gluteus maximus, biceps femoris, semitendinosus, rectus femoris, vastus lateralis, soleus, gastrocnemius and tibialis anterior muscles about their associated joint(s) was determined by means of a joint kinetic, EMG, and muscle length based muscle function classification system.

Analysis of the results indicated that the prime movers of the deep-knee bending movement were the one-joint soleus, vastus lateralis, and gluteus maximus muscles. Although it has been believed in the past that co-contracting antagonistic two-joint muscles functioned simultaneously as knee and hip extensors during the entire ascent phase of the movement, the data presented in this study suggested that these muscles acted mainly as stabilizers of the hip and knee. Despite the fact that lengthening of the rectus femoris, semitendinosus, biceps femoris, and gastrocnemius muscles during periods of co-contraction revealed an ability of these muscles to act as agonists about one joint, there existed no evidence supporting the paradoxical functioning of these muscles as theorized by Lombard (1903) and Molbech (1965).

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INTRODUCTION

The locomotor muscles play an important role in daily living, enabling us to perform those activities which are necessary for our survival and well-being. A thorough understanding of how these muscles act to maintain and/or change joint configurations is therefore essential to those who are involved in teaching movement, promoting normal joint behavior, and treating joint pathology.

To evaluate the functional behavior of muscles, biomechanists must consider musculoskeletal anatomy, related mechanics, and neuromuscular physiology. Since there are often more muscles present which cross a joint than are required to produce the displacement pattern, determining muscle function throughout the entire movement is not a simple task. Complexity of the task is furthermore complicated when co-contractions of two-joint muscles are considered. Two-joint muscles may contract to produce desired and undesired movements at either or both of their corresponding joints. Moreover, consider a system where the antagonistic muscle is also a two-joint muscle crossing the same two joints, as is the case with rectus femoris and biceps femoris or semitendinosus muscles in man (see figure 1.). During co-contraction of rectus femoris and either of the two biarticular hamstrings, one would expect to observe no motion at the hip and knee since their associated joint actions oppose one another. It has been suggested (Lombard,1903; Molbech, 1965), however, that extension of the hip and knee joints will occur as a result of such coordinated muscle activity.

In an attempt to explain this phenomenon, Lombard (1903) has proposed that rectus femoris and the two biarticular hamstring muscles contract isometrically, acting in a tendon-like fashion, to transfer energy generated at the hip to the knee. During simultaneous hip and knee extension, the hamstrings are said to assume a functional role about the knee joint opposite to that which is normally expected (i.e., act as knee extensors). During this instance, the muscle function about the joint of concern is declared as being paradoxical. Consequently, this has been commonly referred to as "Lombard's Paradox".

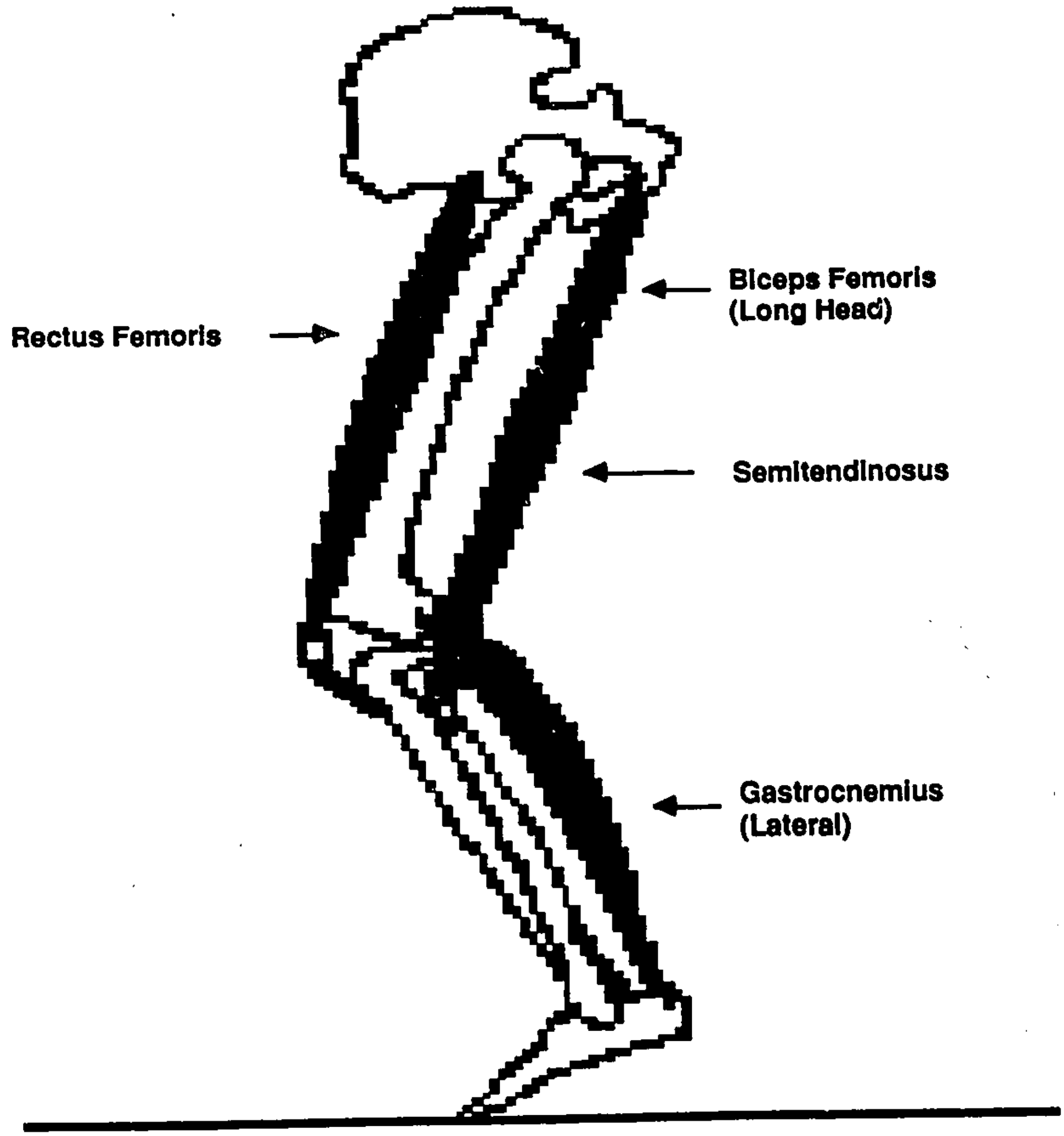

Figure 1. The lower limb biarticular muscles.

Although the reasons by which muscle function can be considered paradoxical are quite straightforward, previous investigators have reported muscle paradoxes in various lower and upper limb movements (Gregor, Cavanagh, & Lafortune, 1985; Simonsen, Thomsen & Clausen, 1985; Olney & Winter, 1985; Wilson, Robertson, & Stothart, 1988). However, the data and methods used to detect paradoxical behavior in these studies are not sufficient to confirm the existence of Lombard's (1903) paradox.

An early study by Carlsoo and Molbech (1966) utilized EMG signals in combination with joint angular velocity to detect paradoxical lower limb biarticular muscle function during deep-knee bends. Muscle paradoxes were reported when predominant EMG activity of the antagonistic biarticular musculature was observed concurrent to hip and knee extension. Although this type of information may provide some evidence of muscle paradoxes, information describing joint kinetics and muscle lengthening rates is required to provide a more precise description of muscle function about the joint. Hence, detection of muscle paradoxes cannot be confirmed by EMG and joint kinematic information alone.

In contrast, Andrews (1985) attempted to define paradoxical activity in deep-knee bending through examination of a first-order differential relationship between muscle length and joint angle. This mathematical function was determined from a simple mechanical representation of the human lower limb. Although the criteria for muscle paradoxes according to this model were not described in detail, it was deemed to be quite inadequate for in vivo purposes since no account for changes in muscle recruitment was considered.

In the same study, Andrews (1985) suggested a systematic means of determining muscle function and used this method in an attempt to detect paradoxical function of the two-joint hamstrings and quadriceps. This system, the Standard Kinetic (SK) method, defined muscle function about a joint in reference to the net joint moment and the moment generated by the individual muscle. Paradoxical muscle function according to SK theory occurs when the direction of an individual muscle moment opposes that of the net joint moment.

Kinesiologists have expressed problems accepting the SK classification system since it provides no means of discriminating between cases in which the muscle is working paradoxically, or as an agonist to decelerate segments, or as a joint stabilizer. The only means by which these conditions could be differentiated would be through simultaneous examination of muscle kinematics and joint kinetics.

With respect to Lombard's (1903) model, the muscle paradox in this case would be detected by observing a muscle moment opposite in direction to the net joint moment while the muscle assumes a constant length. Although these criteria closely agree with those suggested in the initial description of paradoxical phenomena, no studies to date have attempted to detect paradoxical muscle function in deep-knee bending movements using these criteria.

A question is raised about whether or not paradoxical activity truly exists. For a paradox to exist, according to the above definition, the lower-limb two-joint muscles (i.e., biceps femoris, semitendinosus, rectus femoris, and gastrocnemius) must remain the same length or shorten during extension of their associated joints. If no length change or an increase in length is observed then no paradox will occur, instead the muscle will have acted as an eccentric agonist or as a joint stabilizer, respectively.

Purpose

The primary objective of this investigation was to describe the functions of the lower limb muscles during deep-knee bends using a muscle kinematic-joint kinetic-EMG based classification system. The data collected will be used to determine whether or not two-joint hamstrings and quadriceps muscles function, simultaneously, to extend the lower limb joints as proposed by Lombard (1903) and Molbech (1965).

METHODS

Model Description

The human body was modelled as a four segment linked mechanism as illustrated in figure 2. The hip, knee, and ankle joints represented by frictionless pin joints were connected by the appropriate segments modelled as rigid bodies (Winter, 1979). Segmental masses, segmental radii of gyration, and segmental centres of gravity were calculated from proportions described by Dempster (1955) and Plagenhoef (1971). The weighted bar was treated as a particle with its centre of gravity acting at its geometric centre. As stated by McLaughlin, Lardner & Dillman (1978), angular motion of the bar about its centre of gravity was considered negligible during the movement.

The soleus, tibialis anterior, gastrocnemius, vastus lateralis, semitendinosus, biceps femoris, rectus femoris, and gluteus maximus muscle lengths were calculated according to a straight line model presented by Frigo and Pedotti (1978) with modifications made by Hubley (1981). Equations and schematic drawings of each muscle length modelled are included in appendix A.

Subjects

Six male elite-level weight lifters served as the test subjects. Elite athletes were used to control for inter-individual variability in lifting technique. To define segments and their respective joint centres of rotation, markers were placed on the anatomical landmarks listed in table 1.

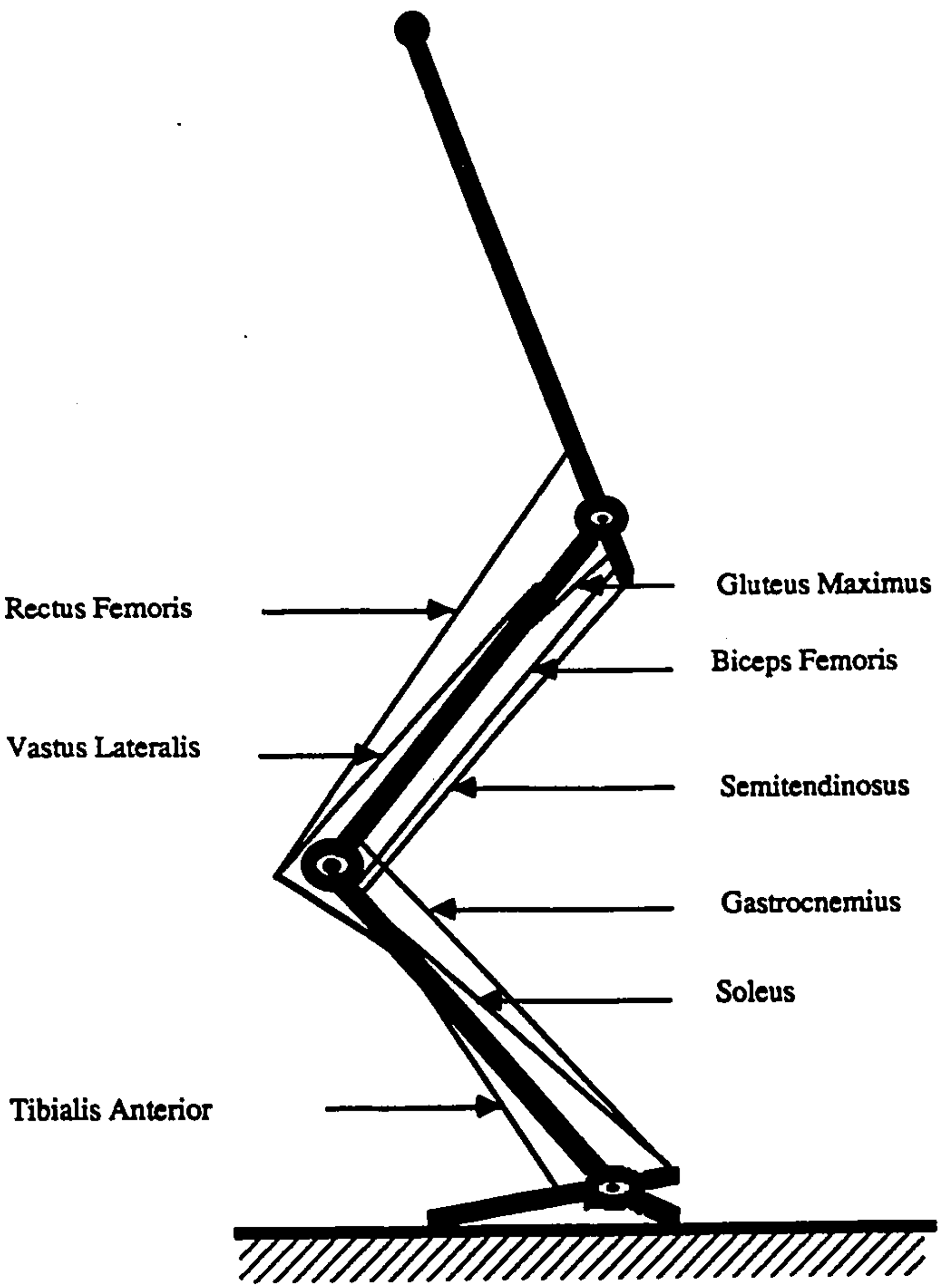

Figure 2. The linked segment model used in this study.

Table 1.

Locations of the anatomical markers placed on all subjects.

Marker #	Landmark
1	Acromion process
2	Greater Trochanter
3	Lateral Femoral Condyle
4	Lateral Malleolus
5	Posterior Calcaneous
6	Fifth Metatarsal-Phalangeal Jt.

Prior to EMG data collection, pairs of MEDI-TRACE silver-silver chloride electrodes were placed, 2.5 cm apart, directly over the motor points of the gastrocnemius (GC), soleus (SO), tibialis anterior (TA), semitendinosus (ST), biceps femoris (BF), rectus femoris (RF), vastus lateralis (VL) and gluteus maximus (GM) muscles according to those locations listed by Delagi (1975). Skin impedance at these sites was verified to be below 20 kilohms. High input-impedance (10 megohms) differential amplifiers (10-700 Hz band pass) were used to produce linear envelope EMG (LE EMG) signals (full wave rectified signals filtered with a second order Butterworth filter with 6 Hz cut-off).

Following a warm-up and a resting period, subjects were asked to execute six separate unloaded and loaded deep-knee bends. Ample rest time (i.e., approximately 180 seconds) was allotted between trials to prevent fatigue. The load lifted in the latter movement represented approximately 80% of each subject's previously recorded maximum. These values are listed in table 2 along with subjects' heights and masses.

Subjects executed the movement at a familiar speed (i.e., 2 second descent and 2 second ascent). One cycle of the deep-knee bend movement consisted of the time taken to complete flexion and extension of the ankle, knee, and hip starting from an upright standing position.

Table 2.

Height and mass of subject s, as well as, load lifted during the loaded condition.

Subject	Height (m)	Mass (kg)	Load lifted (N)
DA	1.78	77.0	892.0
MS	1.88	91.0	1115.0
BH	1.83	95.5	1226.4
MB	1.78	81.8	894.0
KM	1.90	91.0	1130.0
KC	1.78	71.8	600.0

Data Acquisition

While the subject executed the movement with his right foot on a force plate (Kistler), cinematographic, electromyographic, and kinetic data were collected during each movement cycle. Synchronization of data acquisition was governed by the shutter pulse correlator of a 16 mm cine camera (Locam). The camera was positioned perpendicular to the plane of motion and all data were sampled at 50 Hz. The kinetic data consisted of vertical (Fz), antero-posterior (Fx), and lateral reaction forces (Fy), and the location of the centre of pressure acting under the right foot of each subject sampled by a minicomputer (DATA GENERAL microEclipse).

Following the lifting trials, LE EMG signals from three maximal isometric contractions of each muscle were recorded and averaged. These maximal values were used to detect the relative state of muscle activation during the movement cycle.

Data Processing

Within each testing condition of lifting, linear envelope signals for each muscle were normalized over time and normalized to peak cycle EMG. The EMG data for each subject were averaged across five cycles, generating an ensemble average EMG pattern for each muscle (i.e. within-subject ensemble). All within-subject ensemble averages were then averaged by amplitude to produce the across-subject grand ensemble average signal (Yang & Winter, 1984) for the muscle in the specific testing condition.

Film images were projected (4.4 % to 6.0 o. life size) and digitized (approx. 0.30 cm accuracy) on a Hewlett Packard 9874A digitization system. Two trials of digitized film data (i.e., one for each condition) for each subject were then transferred to a mainframe computer (Amdahl) for processing with the BIOMECH (Kinesiology Dept., University of Waterloo) analysis package. Joint angle displacements and velocities were derived from a kinematic analysis and combined with the force plate data in a kinetic analysis resulting in the calculation of net joint moments and powers. These kinematic and kinetic data were then time normalized and averaged across subjects

Muscle-tendon-unit (MTU) lengths for each subject were calculated from relative angle changes (Frigo & Pedotti, 1978; Hubley, 1981) and then time normalized and averaged to standing anatomical length (i.e., the length of the MTU measured when the subject was standing erect). MTU length velocities were calculated by differentiation of the raw MTU length data and subsequently normalized over time. These signals and time normalized MTU lengths were then averaged between subjects, resulting in grand ensemble muscle length histories and muscle length velocity histories, respectively.

Classification System

Muscle function about each joint was examined independently. To be able to effectively assign agonist/antagonist muscle roles, the following assumptions were made: (1) all active myofibrils within a muscle contract homogeneously (Basmajian, 1957), and (2) the moment of force of a muscle can have only one direction at each tendon attachment.

Muscle function during the cycle was assigned using grand ensemble LE EMG, net joint moment, and muscle length velocity signals. The method used to classify muscle function is described below.

1. Muscle function was determined only for those instances where grand ensemble EMG activity surpassed 50% of the maximum recorded during the corresponding activity. According to Gregor et al. (1985), this arbitrary value was considered to be a conservative measure for assuring significant muscle activity.

2. The muscle length velocity during periods of significant activity was used to determine the type of muscle contraction. The muscle was described as contracting eccentrically, concentrically, or isometrically when the MTU velocity was greater than zero, less than zero, or equal to zero, respectively.

3. A muscle was described as being an agonist at the joint when the direction of the individual muscle moment was equal to that of the net joint moment. Conversely, antagonist function at the joint was assigned when the direction of the muscle moment was opposite to that of the net joint moment (Andrews and Hay, 1983).

4. Since it was assumed that random uncontrolled muscle contraction was not possible in movement execution performed by experienced subjects, a muscle was redefined to be acting as a joint stabilizer when, according to the above definitions, it was said to be functioning as a concentric or eccentric antagonist.

RESULTS

Kinematics

On average, a greater portion of the cycle was required for the descent movement compared to the ascent. The descent phase was represented from the 0% to 54% of the cycle while the ascent phase constituted the remaining 46% of the cycle (i.e., 54% to 100% of the cycle).

Changes in joint angles during both lifting conditions are illustrated in figure 3. In both conditions, all three leg joints simultaneously flexed during the descent and extended during the ascent. Minimum joints angles occurred simultaneously at the end of descent. Joint ranges of motion experienced during the movement along with peak flexion angles are presented in table 3. An analysis of variance revealed a significant difference ($P = 0.05$) in ranges of motion between joints. However, no significant difference in ranges of motion between the two conditions was found (see table 4.).

In contrast, joint angular velocity patterns, shown in figures 4, 5, and 6 were significantly different between conditions despite having similar ranges of peak flexion and extension velocities. In general, ankle velocity patterns varied to a greater degree than those at the hip and knee among subjects and between conditions. Greater variability of ankle, knee, and hip joint velocities were observed during the unloaded movement.

Furthermore, it is interesting to note similar hip and knee velocity patterns during the unloaded and loaded conditions, as well as, simultaneous peaking of joint relative angle velocities during both the descending and ascending phases of the movement.

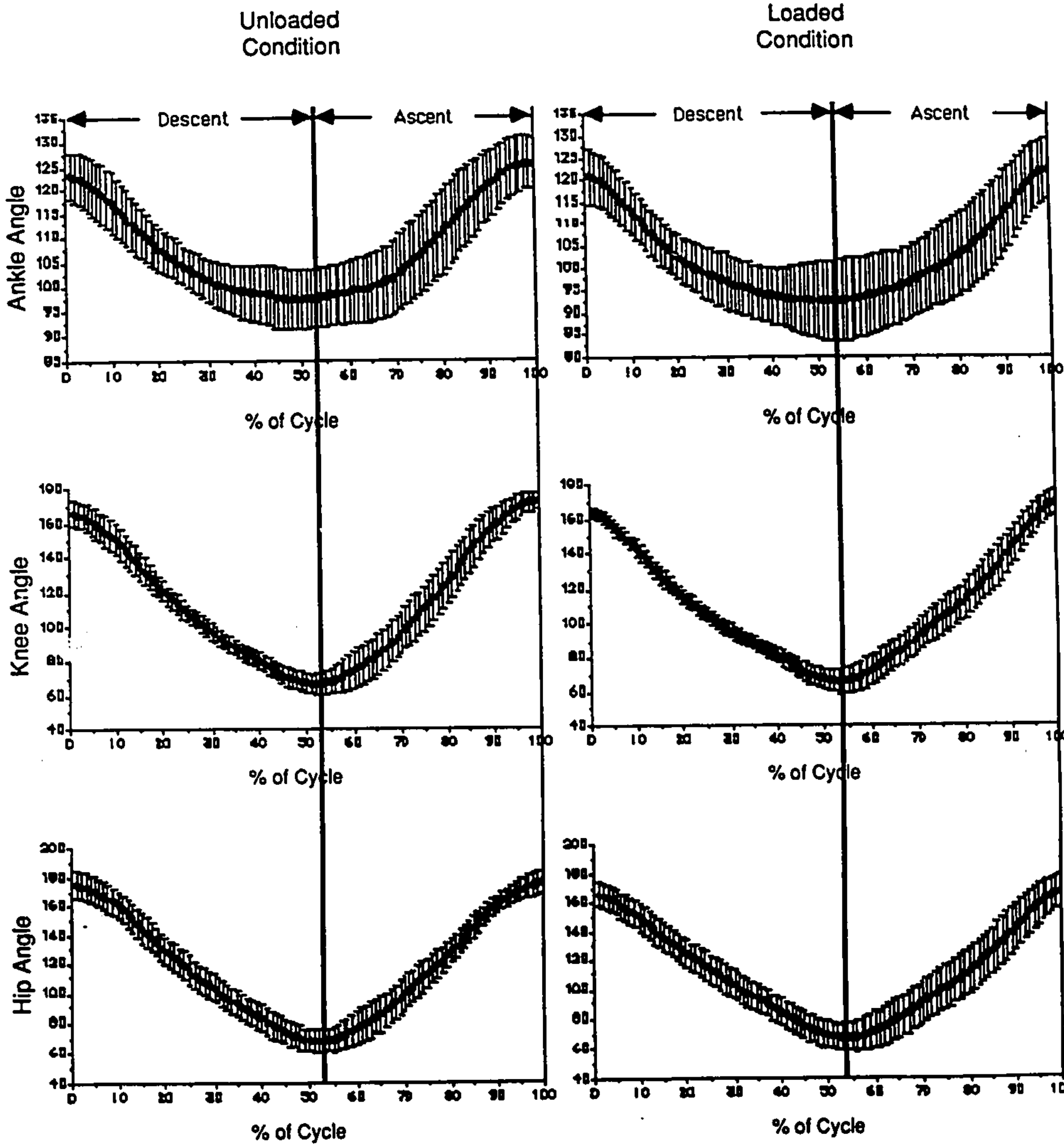

Figure 3. Time-normalized means (n=6) and standard deviations of the joint angle changes (degrees) observed during both the unloaded and loaded deep-knee bend movements.

Table 3.

Ranges of mean joint angles and mean peak joint angular velocities for both movement conditions.

Joint	Joint range of motion (deg)	Peak - ve angular Velocity (rad / s)	Peak +ve angular Velocity (rad / s)
Without Load			
Ankle	91.54 - 130.8	0.370 - 0.683	0.269 - 0.805
Knee	61.08 - 176.8	1.865 - 1.321	1.313 - 2.291
Hip	62.14 - 183.9	1.703 - 1.364	1.358 - 2.152
With Load			
Ankle	83.45 - 129.3	0.755 - 0.555	0.568 - 0.937
Knee	59.81 - 176.0	1.880 - 1.411	1.509 - 2.361
Hip	58.18 - 178.1	1.621 - 1.066	1.359 - 2.302

Table 4.

Summary of ANOVA to detect differences in joint range of motion between the two movement conditions.

Source:	df:	Sum of Squares:	Mean Square:	F-test:	P value:
Between joints	2	6204.043	3102.022	4.071	.0175 *
Between Conditions	1	2812.088	2812.088	3.69	.0552
Interaction	2	261.512	130.756	.172	.8424
Error	594	452650.641	762.038		

* significant at 0.05 level

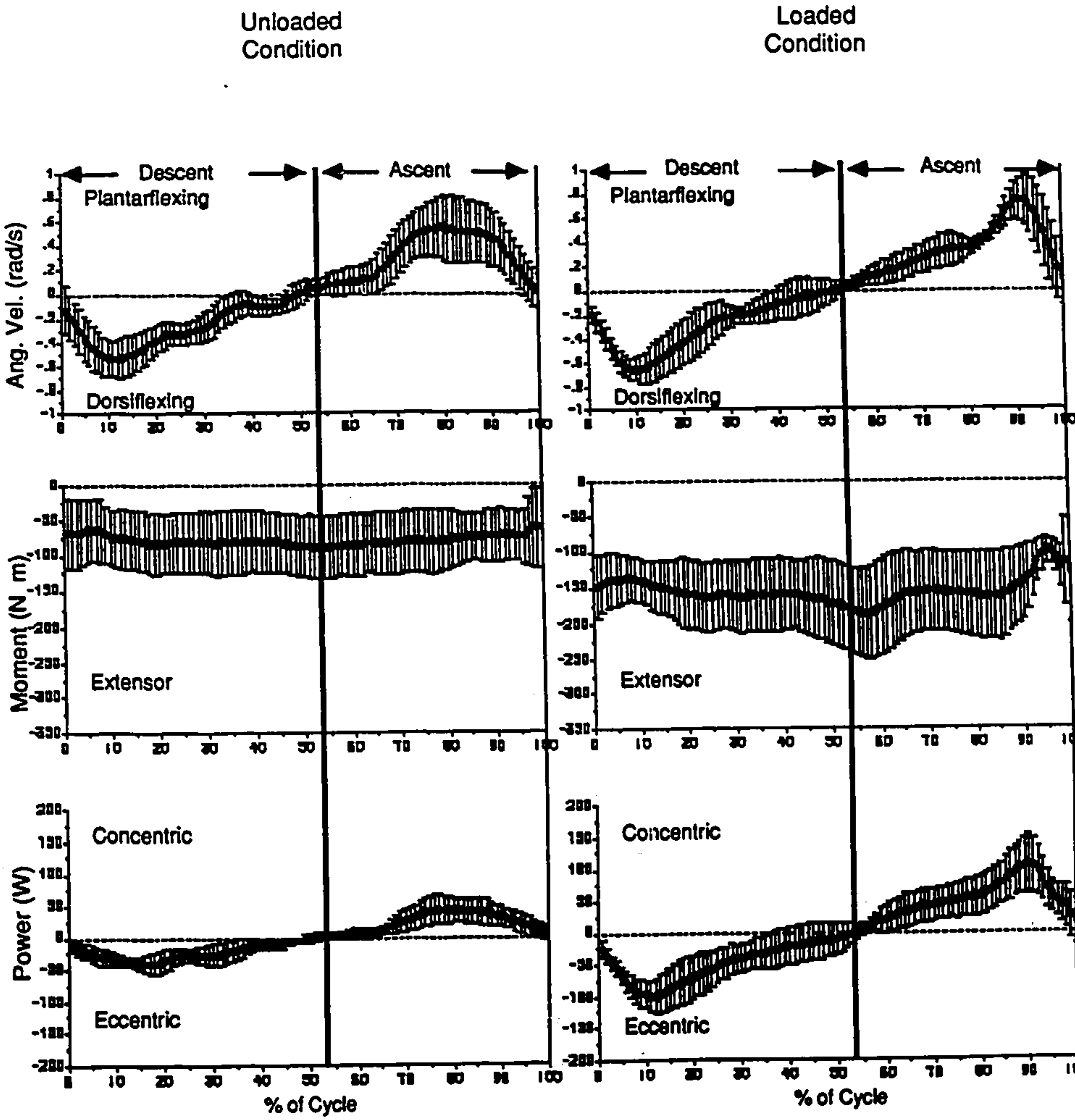

Figure 4. Time-normalized means (n=6) and standard deviations of the ankle joint velocity, net joint moment, and net joint power for the unloaded and loaded movement conditions.

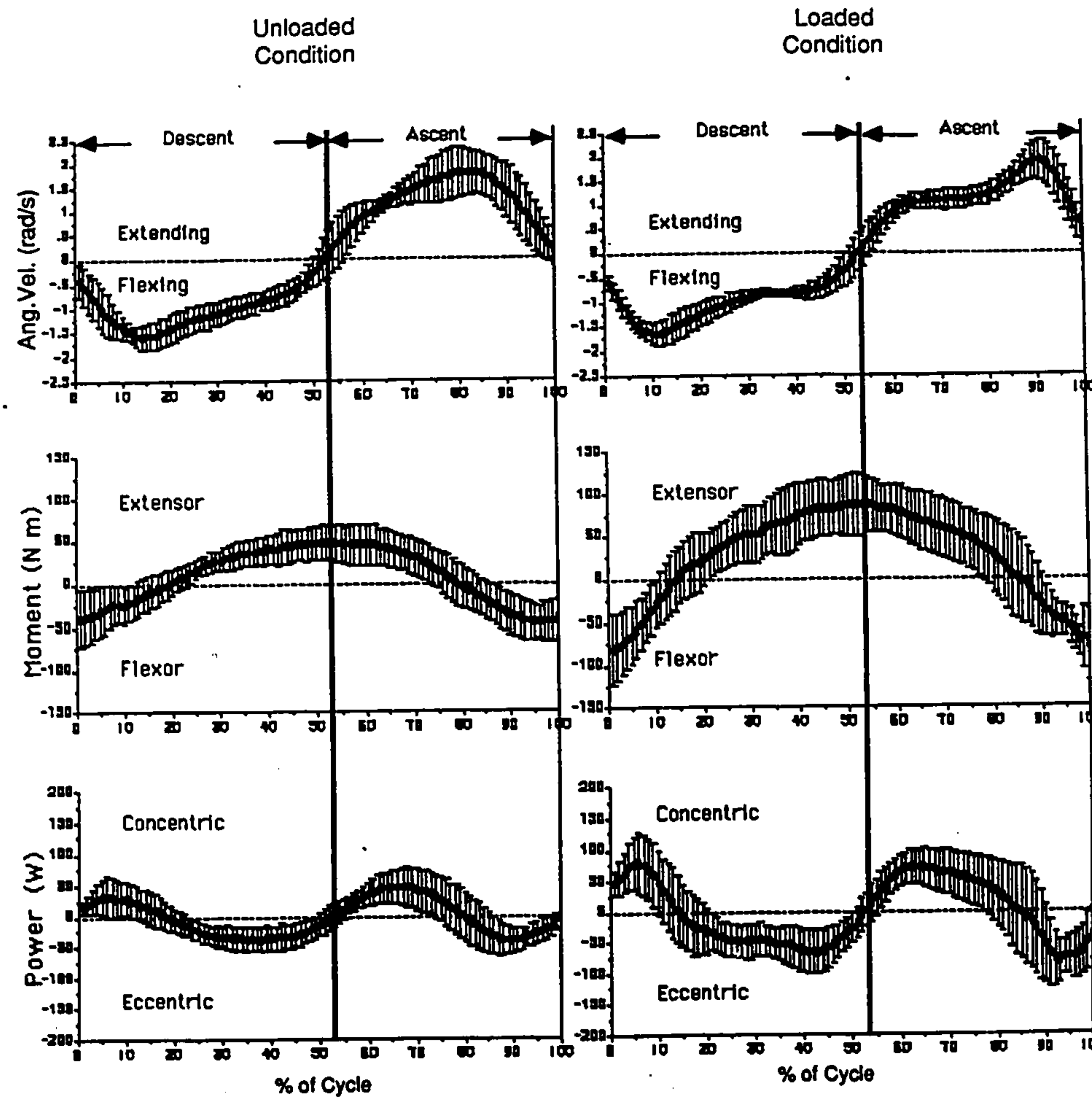

Figure 5. Time-normalized means (n=6) and standard deviations of the knee joint velocity, net joint moment, and net joint power for the unloaded and loaded deep-knee bend movements.

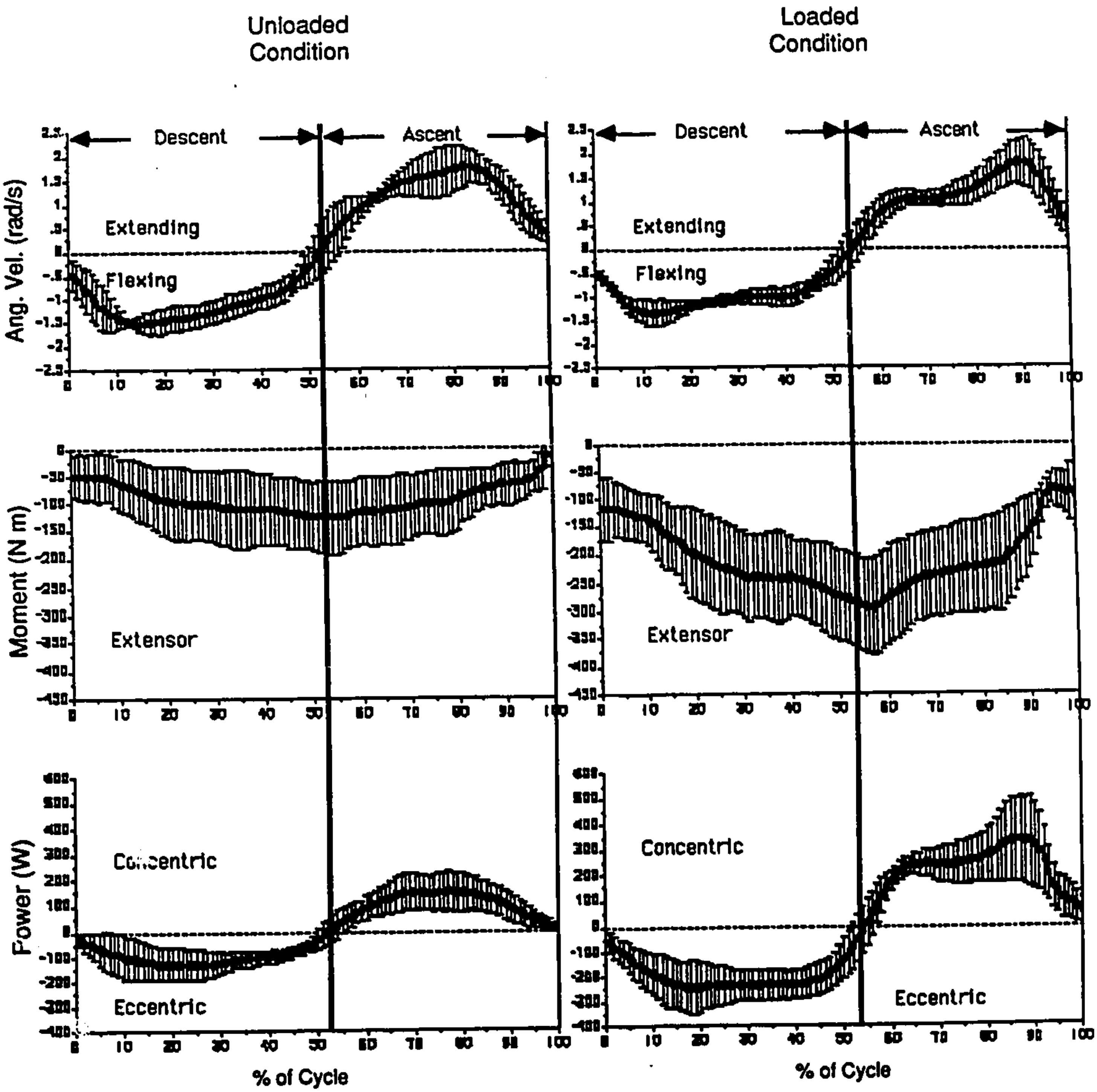

Figure 6. Time-normalized means (n=6) and standard deviations of the hip joint velocity, net joint moment, and net joint power observed during the unloaded and loaded deep-knee bend movements.

Kinetics

The net joint moments represent the net mechanical effect of all connective tissues (i.e., muscles, ligaments, bone, etc..) crossing the joint. The moments provide information concerning which muscle groups were dominant during a phase in the movement. The average net moment histories for the ankle, knee, and hip are presented in figures 4, 5, and 6, respectively. Time of onset and range of magnitude of average peak joint moments are listed in table 5. Although relative magnitudes of these moments varied between subjects, the general patterns exhibited by each individual subject were similar.

Table 5.

Ranges of mean peak joint moments and powers, and occurrence of these within the cycle for both movement conditions.

Joint	Peak moment (N·m)	Occurrence (% cycle)	Peak power (W)	Occurrence (% cycle)
Without Load				
Ankle	42.79 - 132.0	55	24.06 - 62.58	79
Knee	27.77 - 68.64	52	20.30 - 72.65	66
Hip	60.83 - 195.5	54	77.21 - 232.1	77
With Load				
Ankle	122.8 - 250.3	57	61.02 - 151.7	90
Knee	53.37 - 119.4	54	46.53 - 115.5	93
Hip	207.6 - 380.2	57	170.0 - 503.3	86
% Increase between unloaded and loaded conditions				
Ankle	113.5%		145.5%	
Knee	79.22%		74.38%	
Hip	132.1%		117.7%	

Moments at each joint were smaller during the unloaded condition but followed the same general pattern. All three joint moments contributed positively to the net impulse, however, the knee only contributed positively for 60% (.from 20% to 80%) of the total cycle as compared to 70% (from 15% to 85%) during the loaded condition.

The net power at a joint is indicative of the type and instantaneous rate at which work is done by the net moment about a joint and, thus, was used in this study to differentiate muscle group function about a joint into distinct phases. Time of occurrence and magnitude of average peak power values are presented in table 5.

Time-normalized and averaged ankle, knee, and hip joint power signatures are illustrated at the bottom of figures 4, 5, and 6. These curves show that similar power patterns existed among subjects and between conditions. As expected, joint power values proved to be greater for the loaded condition. Maximal power values recorded at the ankle and hip were concentric, while that for the knee was eccentric. The mean percentage increase in peak power at the ankle, knee, and hip brought about by the increased load were 145.5 %, 74.38 %, and 117.7 %, respectively.

According to figures 4 and 6, muscle functions at the ankle and hip can be divided into two phases. In the descending phase, energy was absorbed by joint extensors in the control of downward movement while in the ascending portion of the movement, work was generated by these muscle groups during extension of their respective joints.

In contrast, muscle function at the knee joint (see figure 5) was divided into four distinct phases -- two during descent and two during ascent. Initially during descent, positive work was performed by the knee flexors to unlock the knee. From 15% and 20% of the cycle for the unloaded and loaded conditions, respectively, to the bottom of the descent, energy was absorbed by the knee extensors in an effort to control knee flexion. Positive work was performed by the knee extensors during the initial phase of ascent which started at 54% of the cycle for the unloaded and 85% for the loaded condition. The final portion of the movement was characterized by negative power being performed by the knee flexors. This final bout of energy absorption was most likely required to prevent hyperextension of the joint at the end of the ascent.

Muscle kinematics

The individual muscle kinematic data were used to identify the muscles which had a lengthening/shortening sequence present during the deep-knee bending movement. Muscle lengths were normalized to standing anatomical length as a means of determining the degree of lengthening or shortening during the movement. The average time-normalized length curves for the eight muscles examined in this study are presented in figures 7 and 8, for the unloaded and loaded deep knee bend movements, respectively. These curves illustrate the patterns followed for all six subjects during both the loaded and unloaded conditions. Ranges of shortening and lengthening for each muscle in both lifts are listed in table 6.

In general, all muscles remained within the normal functional range (i.e., 0.8 to 1.2 resting length according to Pierrynowsky, 1982) throughout the motion. In exception to this rule, gluteus maximus and semitendinosus lengthened beyond their normal functional range at the initiation of the ascent phase.

The one-joint muscles, soleus, vastus lateralis, and gluteus maximus, followed similar length change patterns. These muscles were at their maximum recorded lengths at the start of the upward movement and continued to shorten to the end of the cycle. The two-joint muscles, gastrocnemius, biceps femoris, and semitendinosus, as well as, tibialis anterior, shortened in the downward phase and lengthened throughout the entire ascending phase. The semitendinosus muscles experienced, on average, the greatest length changes during the movement. In contrast to the unimodal patterns exhibited by these muscles, rectus femoris muscle length-time pattern was bimodal. It lengthened and shortened during both the descending and ascending phases. Furthermore, its length experienced the smallest length change of all the muscles studied.

The average muscle velocity curves are found in figures 9 and 10 for the unloaded and loaded movement conditions, respectively. A positive velocity in these figures indicated muscle lengthening. Relatively small standard deviations as well as similar patterns between conditions once again indicate minimal variations among individuals and between conditions. All muscles attained velocities which allowed significant force production (i.e., < 0.35 m/s for muscle composed of 100% fast twitch fibres as described by Pierrynowsky, 1982). On the average, peak lengthening/shortening velocities varied between 0.03 and 0.06 m/s, however, the biarticular muscles rectus femoris and semitendinosus differed noticeably. As seen from table 6, peak velocities recorded for these muscles were 0.02 m/s and 0.10 m/s, respectively.

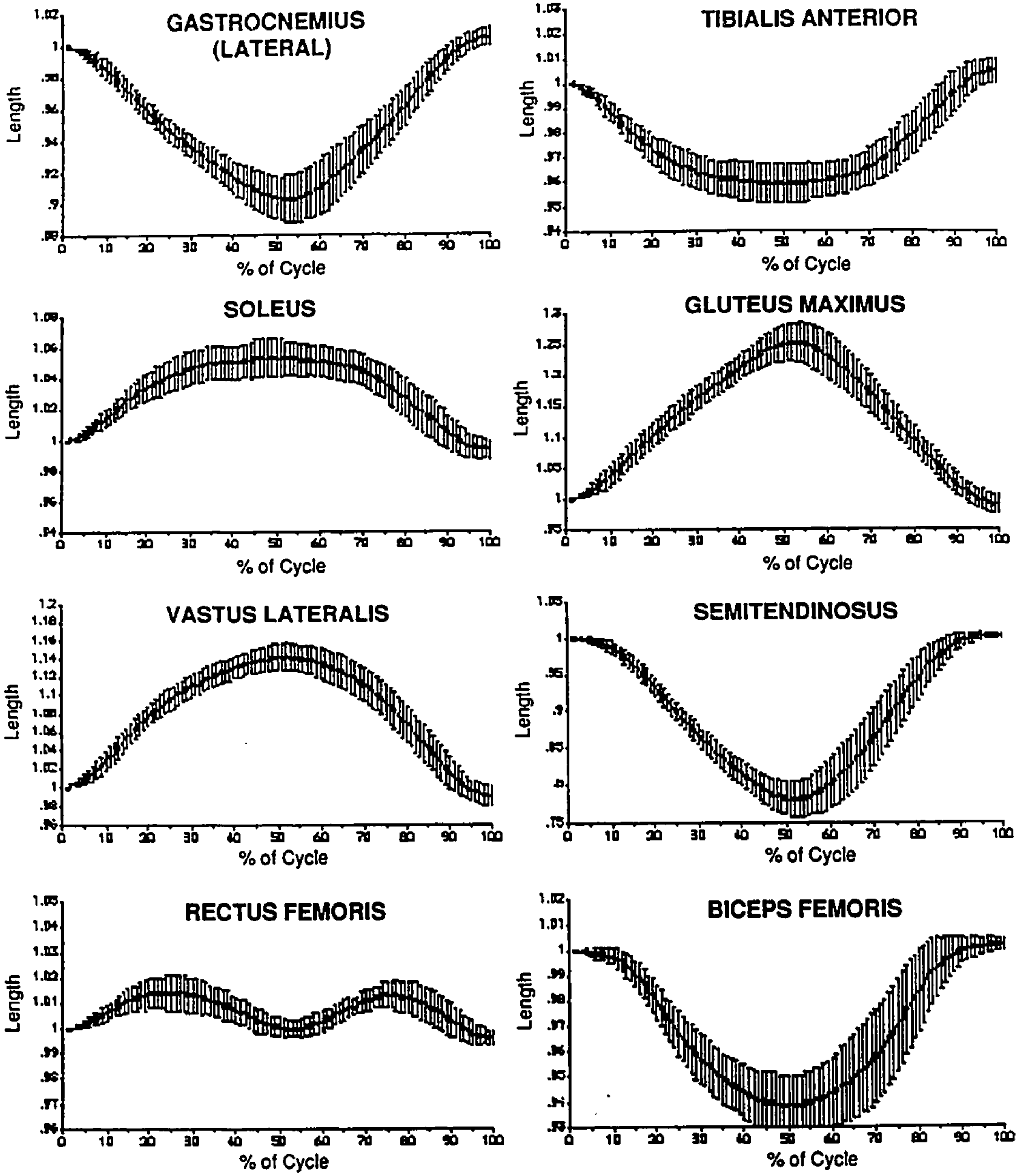

Figure 7. Mean (n=6) and standard deviation of the muscle lengths normalized to anatomical length during the unloaded deep-knee bend movement.

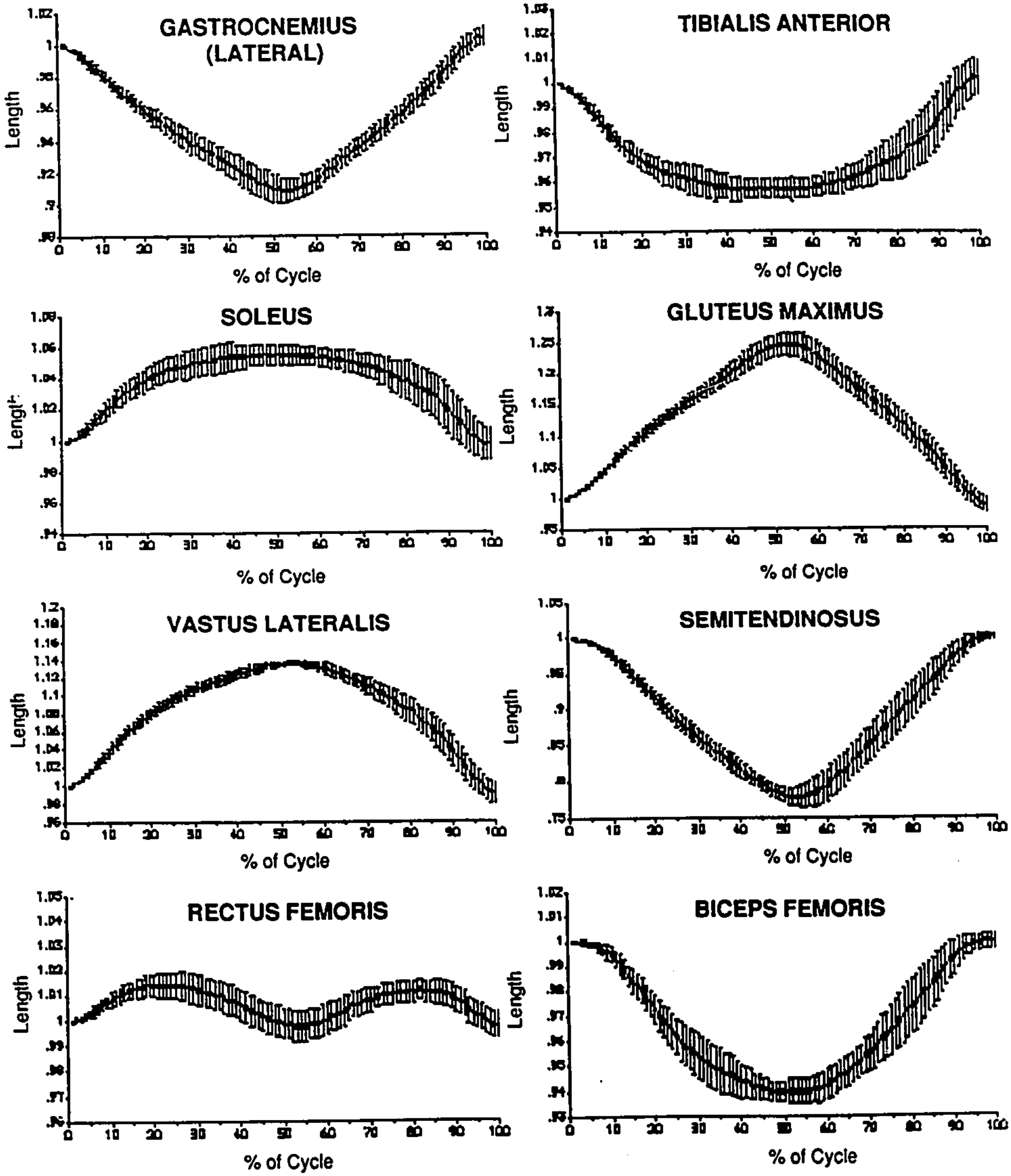

Figure 8. Mean (n=6) and standard deviation of the muscle lengths normalized to anatomical length measured during the loaded deep-knee bend movement.

Table 6

Normalized length and velocity ranges of all muscles during both movement conditions.

Muscle	Range of muscle length	Range of muscle length velocity (m/s)
Without Load		
Soleus	0.993 - 1.064	-0.037 - 0.034
Semitendinosus	0.758 - 1.007	-0.121 - 0.166
Tibialis anterior	0.950 - 1.009	-0.024 - 0.027
Vastus lateralis	0.980 - 1.156	-0.079 - 0.067
Gastrocnemius	0.888 - 1.010	-0.056 - 0.064
Biceps femoris	0.927 - 1.004	-0.042 - 0.051
Rectus femoris	0.993 - 1.021	-0.025 - 0.024
Gluteus	0.974 - 1.282	-0.067 - 0.051
With Load		
Soleus	0.987 - 1.061	-0.044 - 0.038
Semitendinosus	0.758 - 1.007	-0.115 - 0.127
Tibialis anterior	0.950 - 1.009	-0.027 - 0.032
Vastus lateralis	0.980 - 1.156	-0.086 - 0.069
Gastrocnemius	0.901 - 1.010	-0.046 - 0.057
Biceps femoris	0.927 - 1.004	-0.039 - 0.039
Rectus femoris	0.993 - 1.021	-0.028 - 0.023
Gluteus	0.974 - 1.282	-0.069 - 0.053

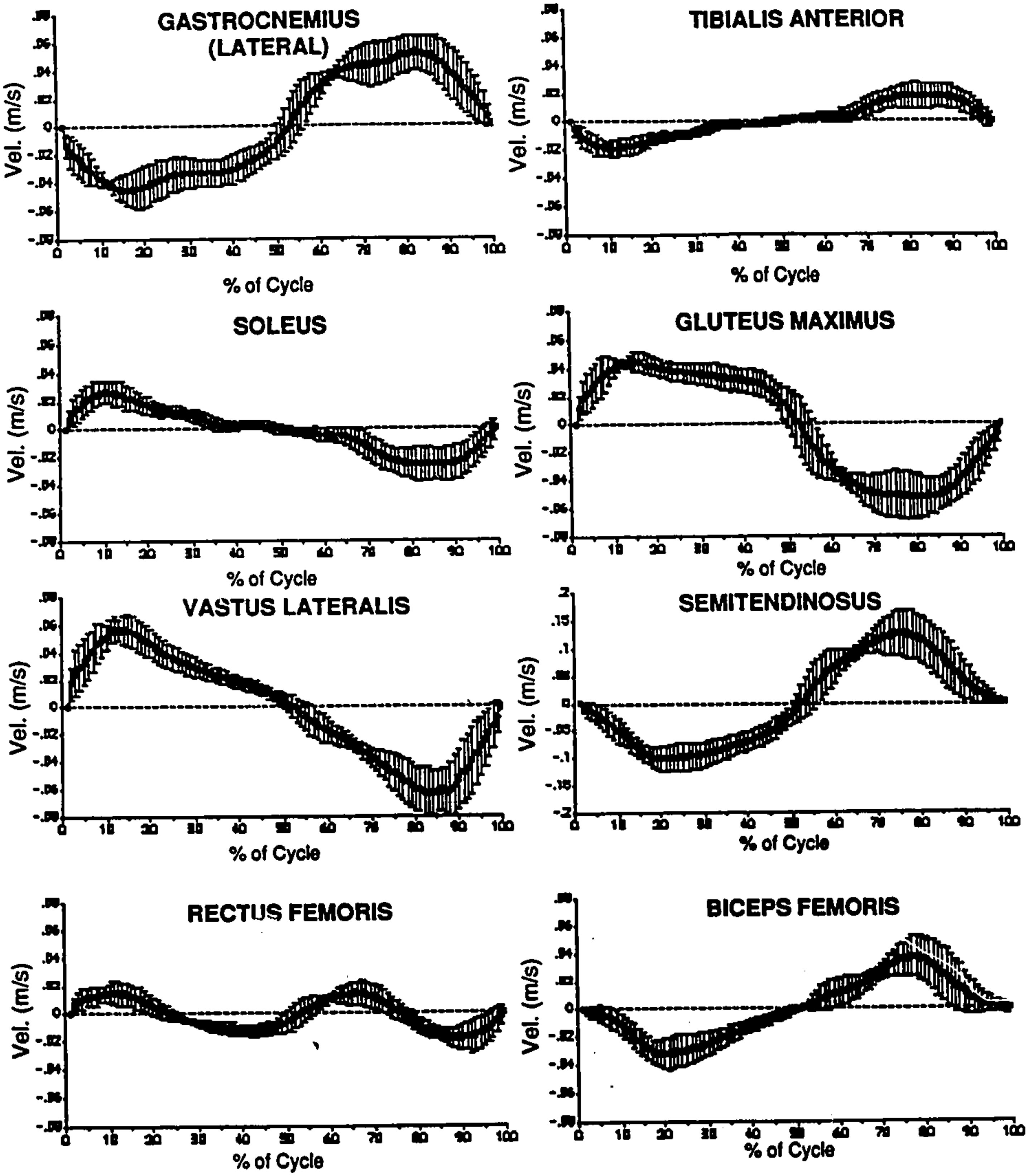

Figure 9. Average (n=6) and standard deviation of the muscle length velocity curves for all muscles studie during the unloaded deep-knee bend movement.

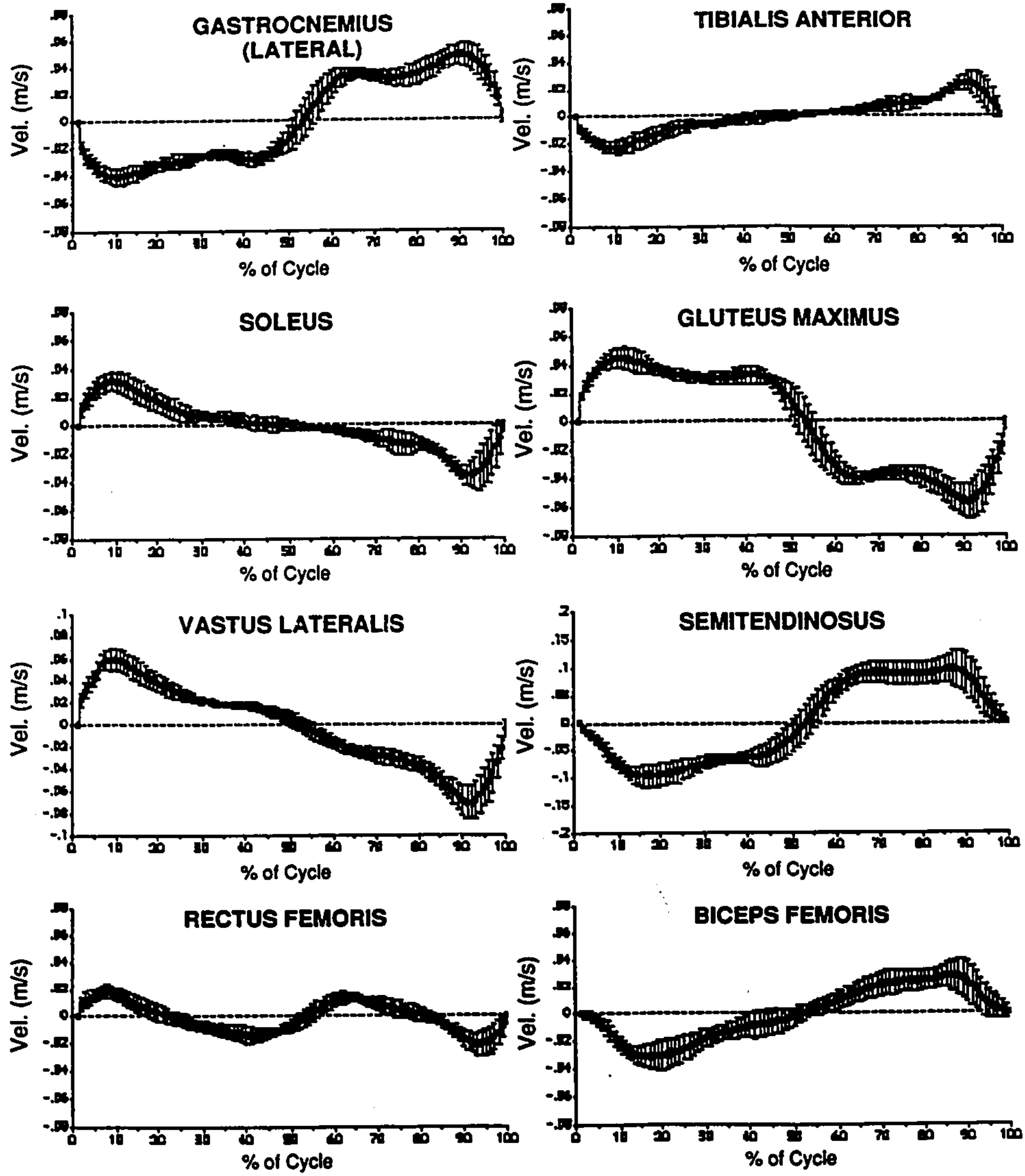

Figure 10. Average (n=6) and standard deviations of the muscle length velocity curves for all muscles studied during the loaded deep-knee bend movement.

Electromyography

Electromyograms were used to determine the activity level of the individual muscles during the phases of the deep-knee bending movement. These values, together with the muscle kinematic data were used to determine which muscles showed an eccentric/concentric contraction sequence during the motion. Although absolute signal values were significantly greater for the loaded condition, general patterns were similar across subjects. Grand ensemble curves are presented in figures 11 and 12 as a representative set of the individual curves. Times of onset and duration of recruitment, as well, as peak EMG values for all eight muscles are listed in table 7. Onset of significant muscle activity during the cycle was considered to have occurred when signal level exceeded 50% of maximum activity during the cycle.

From observation of the EMG signal magnitudes for each muscle during the cycle, it is apparent that the 50% across-subject ensemble criterion level selected in this study may have been too high to classify muscle role in the descent phase of the deep-knee bend motion. Since a muscle undergoing lengthening can generate a greater force at the same EMG level as a muscle undergoing shortening, an elevated criterion activation level would be less sensitive for classifying muscle function in a movement that imposes mechanical demands on a muscle group during lengthening when compared to one that elicits muscle activity during shortening. In fact, this is the case in the deep-knee bend because the net joint moments measured during the descent phase were equivalent to those measured in the ascent (see figures 4, 5 and 6), yet, EMG activity in this study was not considered significant during the former. This indicated that the studied muscle groups produced the same amount of work in descent and ascent but performed it in an eccentric fashion during the former. Hence, the independent relationship between EMG level and eccentric muscle tension suggested that other factors such as muscle tendon length, muscle cross sectional area and moment arm length should be considered to determine the significance of muscle force output during eccentric work.

However, since the original intention of this investigation was to study the extension phase of the deep-knee bend, and this phase was in majority concentric at each joint, the 50% across-subject ensemble criterion level was felt to be justified. It must be understood that as a result of this elevated EMG criterion level, the duration of muscle co-activation periods observed during the entire cycle would be underestimated in this study. In the final run, results would have been essentially the same for ankle and hip muscle function determination while that for the knee would have had additional periods of knee stabilization activity. For this reason, a lower EMG criterion level would be recommended for future use of this classification system in the study of other related movements.

All muscles, with the exception of tibialis anterior, rectus femoris and gluteus maximus, elicited different EMG patterns between the two movement conditions. In general, muscles were recruited earlier and activity lasted longer during the loaded condition. When peak EMG levels measured in both conditions were compared to peak signal levels recorded from a maximal voluntary isometric contraction (i.e., MVC), intensity of muscle recruitment was found to be significantly ($p<0.10$) greater in the loaded movement. EMG values normalized to peak MVC varied considerably amongst subjects in both conditions. Similar results have been reported by Yang and Winter (1983). According to these authors, factors such as the subject's degree of motivation during the test, the number of surrounding synergistic muscles, a lack of visual feedback, muscle substitution brought upon by fatigue, and antagonist co-contraction affect this variability to a great extent. Despite the large variability, these values indicate that all muscles were recruited at a significant intensity during both movement conditions.

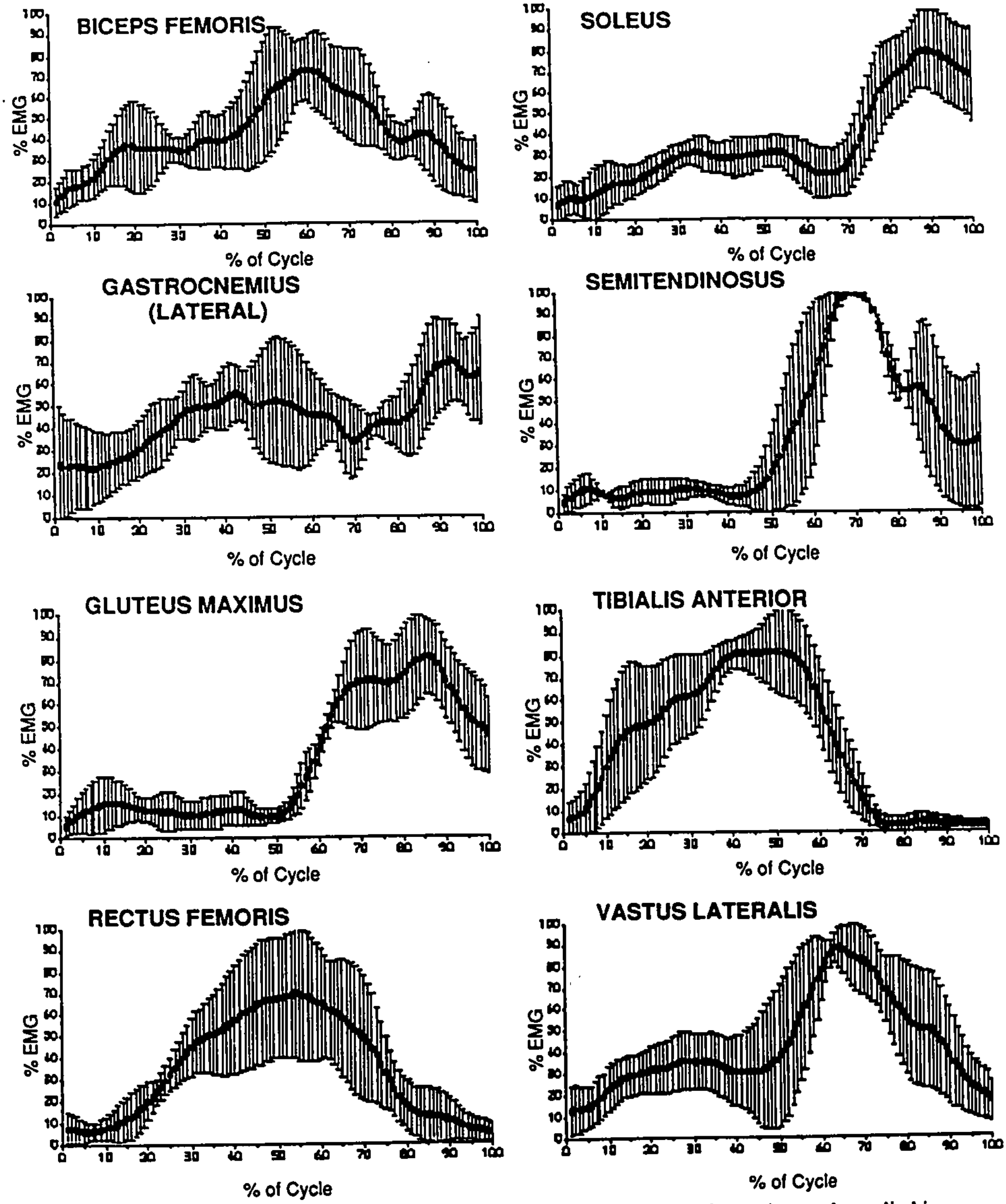

Figure 11. Grand Ensemble EMG (n=6) and standard deviation curves for each muscle studied in the unloaded movement condition.

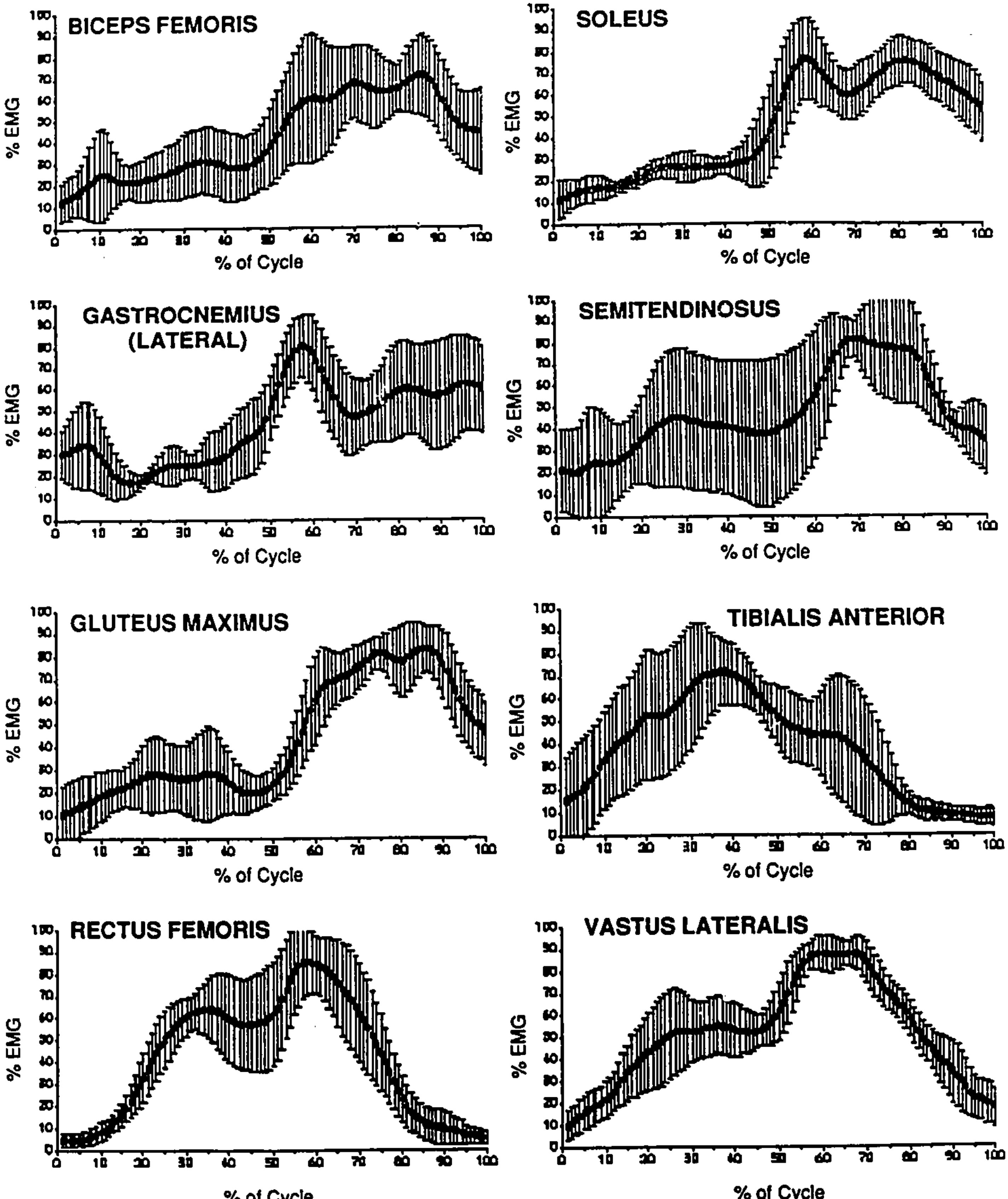

Figure 12. Grand Ensemble EMG (n=6) and standard deviation curves for each muscle studied in the loaded movement condition.

Table 7

Peak normalized EMG values of all muscles during both movement conditions. % MVC denotes the peak EMG value (in volts) expressed as a percentage of the maximum EMG value obtained during a maximal voluntary isometric contraction. The range of these values obtained from the six subjects is included in the last column of this table.

Muscle	Duration (% cycle)	Time of peak (% cycle)	Range of Peak EMG's (% max) [a]	Range of Peak EMG's (% MVC) [b]
Without Load				
Soleus	76 - 100 (24%)	89	61.83 - 99.21	22.26 - 88.26
Semitendinosus	59 - 87 (28%)	70	99.44 - 99.55	11.83 - 80.10
Tibialis anterior	21 - 61 (40%)	50	63.46 - 99.10	59.27 - 92.17
Vastus lateralis	54 - 85 (31%)	64	79.29 - 97.37	39.60 - 192.19
Gastrocnemius	37 - 54 (18%)	43	43.51 - 68.06	4.52 - 39.61
	85 - 100 (15%)	93	51.05 - 88.87	13.95 - 75.91
Biceps femoris	48 - 77 (29%)	61	57.05 - 89.61	9.71 - 88.31
Rectus femoris	34 - 69 (35%)	54	39.56 - 99.83	23.99 - 117.50
Gluteus	63 - 98 (35%)	86	65.19 - 97.82	9.07 - 129.68
With Load				
Soleus	52 - 100 (48%)	59	57.39 - 94.60	65.54 - 145.96
Semitendinosus	59 - 89 (30%)	69	72.88 - 90.38	22.93 - 79.70
Tibialis anterior	19 - 51 (32%)	37	56.68 - 86.38	20.06 - 77.54
Vastus lateralis	25 - 82 (57%)	67	81.75 - 94.67	68.82 - 228.33
Gastrocnemius	50 - 67 (17%)	58	65.74 - 95.12	20.03 - 33.92
	74 - 100 (16%)	96	40.26 - 84.54	35.61 - 65.26
Biceps femoris	55 - 93 (38%)	86	52.36 - 89.76	41.33 - 248.02
Rectus femoris	25 - 73 (48%)	58	70.66 - 100.00	86.05 - 208.08
Gluteus	58 - 98 (40%)	86	72.95 - 93.27	47.35 - 208.42

[a] Percentage of maximal EMG during movement cycle.

[b] Peak EMG during the cycle expressed as a percentage of peak EMG measured during a maximal voluntary isometric contraction (MVC).

In both conditions, the descent phase was characterized by dominant tibialis anterior, vastus lateralis, and rectus femoris activity (see table 7 for peak values). Tibialis anterior activity decreased below 50% maximum just prior to the initiation of the ascent.

The ascent phase was characterized by large bursts of soleus, gluteus maximus, semitendinosus, gastrocnemius, and biceps femoris activity. Vastus lateralis and rectus femoris also exhibited a peak of activity during the early parts of this phase.

In reference to the antagonistic muscle activity of the ankle, there existed no co-activation of the soleus and tibialis anterior or of the gastrocnemius and tibialis anterior muscle groups in either movement condition. Soleus and gastrocnemius exhibited quite similar patterns, acting simultaneously during the middle portion of the cycle.

Similarly, rectus femoris and vastus lateralis were recruited simultaneously and showed similar activity patterns. The EMGs also revealed co-activation of semitendinosus and biceps femoris during the ascending phase, however, biceps femoris was recruited before and longer than semitendinosus.

Similar to Carlsoo's and Molbech's (1966) findings, significant bouts of co-activation of antagonistic muscles, acting about the knee and hip (i.e., rectus femoris, biceps femoris, and semitendinosus) were observed during both the descending and ascending phases. Furthermore, significant co-activation periods of gastrocnemius, semitendinosus, and biceps femoris activity appeared in the ascending phase.

Durations of biarticular muscle co-activation periods and times of occurrence for both conditions are shown in table 8. In general, the duration of muscle co-activation increased from the unloaded to the loaded condition. In exception to this, biceps femoris and rectus femoris showed a 3% decrease in duration in the loaded condition. The greatest increase in duration between conditions was obtained from gastrocnemius and semitendinosus followed by gastrocnemius and biceps femoris.

Time of onset of semitendinosus and rectus femoris co-activation was identical between conditions while that for biceps femoris and rectus femoris, and for gastrocnemius and biceps femoris occurred later on in the cycle during the loaded condition.

Gastrocnemius and semitendinosus co-activation occurred earlier and lasted longer during the loaded condition.

Table 8

Occurrence and duration of biarticular muscle co-activation during both movement conditions.

Muscle Pair	Time of occurrence (% cycle)	Duration (% cycle)
Without Load		
Semitendinosus - Rectus femoris	59 - 69	10%
Biceps femoris - Rectus femoris	48 - 69	21%
Gastrocnemius - Biceps femoris	48 - 55	7%
Gastrocnemius - Semitendinosus	85 - 87	2%
With Load		
Semitendinosus - Rectus femoris	59 - 73	14%
Biceps femoris - Rectus femoris	55 - 73	18%
Gastrocnemius - Biceps femoris	55 - 67 & 74 - 93	12% & 19%
Gastrocnemius - Semitendinosus	50 - 67 & 74 - 89	17% & 15%
Increase in duration between the unloaded and loaded conditions		
Semitendinosus - Rectus femoris		4%
Biceps femoris - Rectus femoris		-3%
Gastrocnemius - Biceps femoris		24%
Gastrocnemius - Semitendinosus		30%

DISCUSSION

To simplify the ensuing discussion, muscle functions about the ankle, knee, and hip during the lifting cycle were summarized in tables 9, 10, and 11, respectively. As a means of describing function in greater detail, the following variables were included in these tables; the period of significant muscle activity within the cycle (i.e., when the EMG exceeded 50% of the across-subject ensemble average), the direction of the net joint moment, the direction of the moment created by the specific muscle about the joint, the direction of the joint angular velocity, and the direction of the muscle velocity. Muscle function during the descent and ascent phases was assigned according to the classification system described previously.

To simplify the explanation of the results, muscle function at the associated joint(s) were described for each individual muscle studied, starting with those acting about the ankle and progressing towards the hip. The discussion concludes with consideration of the functions of the biarticular muscles during periods of co-activation with reference to the theories proposed by Lombard (1903) and Molbech (1965).

Tibialis anterior

As seen in table 9, tibialis anterior functioned as a stabilizer about the ankle during descent by contracting concentrically to resist plantarflexion. When movement of the total body centre of gravity in the antero-posterior direction was considered, it appeared to be working in effort to maintain balance by resisting posterior rotation about the ankle. During this phase, tibialis anterior prevented the subject from falling backwards as the total system centre of gravity was simultaneously lowered and moved posteriorly. For a brief period during the ascending phase, it stabilized the ankle by means of an eccentric contraction.

Table 9

Muscle function about the ankle during both movement conditions.

Muscle	Active period (%)	Joint moment direction	Muscle moment direction	Direction of joint movement	Type of contraction	Function
Without Load						
Descent phase						
Tibialis anterior	21 - 55	extensor	flexor	flexion	concentric	antagonist
Gastrocnemius	37 - 55	extensor	extensor	flexion	concentric	agonist
Ascent phase						
Soleus	76 - 100	extensor	extensor	extension	concentric	agonist
Tibialis anterior	55 - 61	extensor	flexor	extension	eccentric	antagonist
With Load						
Descent phase						
Tibialis anterior	19 - 51	extensor	flexor	flexion	concentric	antagonist
Soleus	52 - 55	extensor	extensor	flexion	concentric	agonist
Ascent phase						
Soleus	55 - 100	extensor	extensor	extension	concentric	agonist
Gastrocnemius	50 - 55	extensor	extensor	flexion	concentric	agonist
Gastrocnemius	55 - 67	extensor	extensor	extension	eccentric	agonist
Gastrocnemius	74 - 100	extensor	extensor	extension	eccentric	agonist

Soleus

In contrast, the soleus muscle functioned agonistically during both the descent and ascent phases. Throughout the entire ascent movement it functioned as a concentric agonist causing plantarflexion of the ankle. Despite the difference in the duration, soleus muscle function was similar in both movement conditions.

Gastrocnemius

In comparison to the function of monoarticular muscles, the biarticular gastrocnemius muscle functioned differently between conditions. Even though recruitment of this muscle occurred during the descent phase of both conditions, the onset of recruitment occurred earlier in the unloaded movement. During the descent, gastrocnemius worked as a concentric agonist at the ankle, aiding in the production of the net ankle extensor moment. In the ascent phase of the loaded condition, the gastrocnemius continued to function agonistically about the ankle as it lengthened. Contrary to conventional lines of thought, gastrocnemius appeared to contribute to the ankle plantarflexor moment while performing eccentric work. However, since this muscle spans both the ankle and knee joints, its function at this time is better understood when activities about the knee are considered.

Gastrocnemius functioned as an antagonist at the knee by applying a flexor moment in opposition to the net knee extensor moment. It continued to oppose the dominant knee extensor moment for approximately two thirds of the ascent phase, while undergoing lengthening. This pattern of activity suggested that gastrocnemius functioned mainly as a knee joint stabilizer and ankle plantarflexor during the descent and ascent phases of deep-knee bending.

In the loaded movement condition, gastrocnemius functioned as an eccentric agonist at the knee during the last 15% of the ascent. Since this activity appeared simultaneous to the negative work performed by the net knee moment this muscle absorbed energy to prevent knee hyperextension.

Vastus lateralis

As expected, muscle activity about the knee was characterized throughout the cycle by dominant agonistic action of the monoarticular vastus lateralis. During descent this muscle acted eccentrically to absorb energy in the control of knee flexion. Throughout the ascent period, it contracted concentrically to extend the knee and hence contributed to the production of the net knee extensor moment. Vastus lateralis seemed to initiate extension of the knee and had less contribution during the last 20% of the ascent.

Rectus femoris

Although rectus femoris was also considered to act as a prime knee extensor, it was recruited differently to the vastus lateralis during both movement conditions. This muscle functioned as an agonist about the knee during late descent by its concentric contraction. During the same time period, it acted antagonistically to flex the hip. While active in the ascent, rectus femoris maintained these functions about the knee and hip, however, it experienced lengthening in this phase. In both cases this muscle contributed positively to the production of the net extensor moment measured at the knee. Since the dominant extensor moment at the hip would have the additional effect of rotating the pelvis posteriorly, a resistive hip flexor moment would be required throughout the entire movement to negate this effect and stabilize the pelvis. According to these data, rectus femoris would appear to be functioning in such a manner throughout the deep-knee bending movement.

Biceps femoris and semitendinosus

Despite having slightly different recruitment patterns, biceps femoris, and semitendinosus muscles functioned similarly between conditions. At the knee, these muscles functioned in the same manner as gastrocnemius. During late descent (i.e., 48% to 55% of the cycle), biceps femoris acted to resist the net knee extensor moment during flexion, functioning as a knee joint stabilizer, while at the hip, it contributed positively to hip extension. Although semitendinosus was not recruited in the descent phase, the two biarticular muscles were co-activated in the ascent. During this time, these muscles functioned as eccentric antagonists at the knee and eccentric agonists at the hip. Hence, the two-joint hamstring muscles assisted gastrocnemius in maintaining knee joint integrity during knee extension while contributing positively to the simultaneous hip extension.

Similar to the gastrocnemius, both of these muscles functioned as eccentric agonists about the knee during late ascent (i.e., 85% to 100%). Biceps femoris and semitendinosus muscles performed negative work to prevent knee ligament damage caused by excessive knee hyperextension. This type of function was observed in both movement conditions.

Gluteus maximus

Finally, as expected, the prime hip extensor muscle, gluteus maximus, was active as a concentric agonist to hip extension throughout the ascending phase. Although EMG activity was not significant during the descent, this muscle functioned eccentrically to control hip flexion during this phase.

Table 10

Muscle function about the knee during both movement conditions.

Muscle	Active period (%)	Joint moment direction	Muscle moment direction	Direction of joint movement	Type of contraction	Function
Without Load						
Descent phase						
Gastrocnemius	37 - 55	extensor	flexor	flexion	concentric	antagonist
Rectus Femoris	34 - 55	extensor	extensor	flexion	concentric	agonist
Biceps Femoris	48 - 55	extensor	flexor	flexion	eccentric	antagonist
Ascent phase						
Gastrocnemius	85 - 100	flexor	flexor	extension	eccentric	agonist
Rectus Femoris	55 - 69	extensor	extensor	extension	eccentric	agonist
Biceps Femoris	55 - 77	extensor	flexor	extension	eccentric	antagonist
Vastus lateralis	55 - 80	extensor	extensor	extension	concentric	agonist
Vastus lateralis	80 - 85	flexor	extensor	extension	concentric	antagonist
Semitendinosus	59 - 80	extensor	flexor	extension	eccentric	antagonist
Semitendinosus	80 - 87	flexor	flexor	extension	eccentric	agonist
With Load						
Descent phase						
Gastrocnemius	50 - 55	extensor	flexor	flexion	concentric	antagonist
Rectus Femoris	25 - 55	extensor	extensor	flexion	concentric	agonist
Vastus lateralis	25 - 55	extensor	extensor	flexion	concentric	agonist
Ascent phase						
Gastrocnemius	55 - 67	extensor	flexor	extension	eccentric	antagonist
Gastrocnemius	74 - 85	extensor	flexor	extension	eccentric	antagonist
Gastrocnemius	85 - 100	flexor	flexor	extension	eccentric	agonist
Rectus Femoris	55 - 73	extensor	extensor	extension	eccentric	agonist
Biceps Femoris	55 - 85	extensor	flexor	extension	eccentric	antagonist
Biceps Femoris	85 - 93	flexor	flexor	extension	eccentric	agonist
Vastus lateralis	55 - 82	extensor	extensor	extension	concentric	agonist
Semitendinosus	59 - 85	extensor	flexor	extension	eccentric	antagonist
Semitendinosus	85 - 89	flexor	flexor	extension	eccentric	agonist

Table 11

Muscle function about the hip during both movement conditions.

Muscle	Active period (%)	Joint moment direction	Muscle moment direction	Direction of joint movement	Type of contraction	Function
Without Load						
Descent phase						
Rectus femoris	34 - 55	extensor	flexor	flexion	concentric	antagonist
Biceps femoris	48 - 55	extensor	extensor	flexion	isometric	agonist
Ascent phase						
Gluteus max.	63 - 98	extensor	extensor	extension	concentric	agonist
Semitendinosus	59 - 87	extensor	extensor	extension	eccentric	agonist
Biceps femoris	55 - 77	extensor	extensor	extension	eccentric	agonist
Rectus femoris	55 - 69	extensor	flexor	extension	eccentric	antagonist
With Load						
Descent phase						
Rectus femoris	25 - 55	extensor	flexor	flexion	concentric	antagonist
Ascent phase						
Gluteus max.	58 - 98	extensor	extensor	extension	concentric	agonist
Semitendinosus	59 - 89	extensor	extensor	extension	eccentric	agonist
Biceps femoris	55 - 93	extensor	extensor	extension	eccentric	agonist
Rectus femoris	55 - 73	extensor	flexor	extension	eccentric	antagonist

<u>Biarticular muscle function during periods of co-activation</u>

As stated earlier in the methodology, it was assumed that undesirable muscle contractions were non-existent throughout the movement, therefore all muscle recruitment occurred to fulfill a mechanical function. Agonistic muscle function can be exhibited in two general manners, either in a concentric fashion (cf. Rasch & Burke, 46:1978) where the muscle is actively contributing in the production of positive work to accelerate a joint or in an eccentric fashion to decelerate a joint. Under the proposed system, when muscles are found to be functioning as eccentric or concentric antagonists, they were considered to be functioning as joint stabilizers. Stabilizing muscles are recruited to help maintain joint integrity by applying a moment opposite in direction of the net joint moment. In this way disruptive action produced by the dominant moment is countered by the stabilizer moment. This functions also to protect the joint against unexpected perturbations such as pain caused by soft tissue failure, slipping, external blows, etc. Although this type of function is mechanically inefficient, it is deemed necessary for protection of the joint from injury.

When co-active biarticular muscle activity about the three joints are examined during the ascent phase, each muscle is observed to have different functions at the two joints they traverse. In general, these muscles act as eccentric stabilizers at one joint while being eccentric agonists at the other.

As mentioned previously, rectus femoris function at the hip assisted in countering pelvic rotation by resisting the dominant hip extensor moment created by the gluteal and biarticular hamstring muscles. According to Rasch and Burke (47: 1978), rectus femoris should function to brace the pelvis eccentrically to guide it during the ascent movement. Under their classification scheme, this muscle would be considered as a "moving stabilizer" of the hip throughout the movement. It is most likely for this reason that rectus femoris elicits a significantly different recruitment pattern to that of the vastus lateralis during the deep-knee bending motion.

In contrast, the biarticular hamstring muscles, biceps femoris and semitendinosus function reciprocally to the rectus femoris. At the hip, these muscles contributed eccentrically to the generation of the net extensor moment and aid stabilization of the knee. Simultaneously, gastrocnemius plantarflexs the ankle and contributes to knee stabilization. The large amount of knee stabilizing activity observed near the end of descent and initiation of ascent, is not surprising since the knee joint is subjected to the greatest level of structural instability during this period of the movement cycle (McLaughlin et. al. ,1978). Further evidence of knee stabilizing activity can be seen when a low net knee extensor moment is present from mid-descent to mid-ascent despite elevated activity of the vastus lateralis and rectus femoris.

When a monoarticular muscle is observed to function as an eccentric agonist, it acts to decelerate the joint it traverses. However, when a biarticular muscle is recruited in such a manner its role is to accelerate at least one of the joints which it spans. This seems to be the way rectus femoris functions at the knee as it guides the hip during the ascent phase.

More specifically, it would appear that rectus femoris acts to transfer energy from the iliacus to the tibia as well as to absorb energy in the stabilization of the hip during simultaneous hip and knee extension. Similarly, biceps femoris and semitendinosus muscles contribute to production of hip extension, as well as, knee stabilization by means of an eccentric contraction. As in the case of the rectus femoris, these muscles act both to absorb and transmit energy from the tibia to the iliacus. Gastrocnemius also functions in this manner to absorb and transmit energy from the calcaneus to the femur.

These findings are similar to explanations proposed by Rasch and Burke (1978). According to these authors, biarticular hamstrings and quadriceps femoris muscles possess the ability to transmit energy from the hip to the knee during the entire hip and knee extension movement when one or both muscle groups maintain a constant length. This was defined as the "tendinous or belt-like action" of two-joint muscles. Although the antagonistic two-joint muscles studied in this investigation did not show isometric contraction characteristics during ascent, the same effect is made possible when resisted lengthening of these muscles occurs simultaneously.

Examination of the lengthening velocity of rectus femoris and biceps femoris when co-activated (i.e., between 55% and 65% of the cycle) revealed that both of these muscles lengthened at the same rate, however, their velocities differed quite noticeably in the remaining period of ascent. While the isometric contraction condition simulated in Rasch's and Burke's (1978) mechanical model represents an ideal mechanism for simultaneous hip and knee extension, the functions of biarticular quadriceps and hamstrings revealed in this in vivo study seem to be a compromise between this ideal condition and that of optimum joint stability.

According to a mathematical model presented by Molbech (1965), biarticular hamstrings and gastrocnemius act, during simultaneous hip and knee extension, to extend the knee at knee angles greater than 135 degrees. Despite having observed some EMG activity of these muscles during the ascent phase of the movement, very little supporting evidence of this theory was revealed in this investigation. According to the processed data, the knee angle exceeded 135 degrees at 83% (range = 75% - 88%) and 88% (range = 80% - 93%) of the cycle for the unloaded and loaded movements, respectively. Upon examination of the EMG activities, semitendinosus was considered active for only 4% of the total cycle after the knee angle surpassed 135 degrees during the unloaded movement. This value diminished to 1% in the loaded condition. Moreover, biceps femoris and gastrocnemius were not active in this part of the cycle for the unloaded condition. When the extra load was introduced, however, activity of these muscles increased when the knee angle exceeded 135 degrees, for 5% and 12% of the total cycle, respectively. Similar findings were reported for the squatting movement by Stanhope (1984).

Furthermore, the kinematic and kinetic data indicate that the biarticular hamstring muscles were contracting eccentrically during the ascent phase of the cycle to prevent knee hyperextension. Since these muscles were absorbing energy about the knee they can not be considered as knee extensors during the ascent phase of the loaded deep-knee bending movement. Hence, the data presented in this investigation questions the validity of Molbech's (1965) model when applied to in vivo conditions.

CONCLUSIONS

Upon review of lower limb muscle function about the three lower limb joints the proposed classification system has indicated that the agonists of the deep-knee bending movement were the monoarticular soleus, vastus lateralis, and gluteus maximus muscles. Although co-activated antagonistic biarticular muscles have been believed to act as agonists during the entire period of simultaneous ankle, knee, and hip extension, the data presented in this study suggest that these muscles acted mainly as stabilizers of the hip and knee for both the loaded and unloaded movement conditions. Relatively short periods of biarticular muscle co-activation observed during ascent did not support the explanation that these muscles functioned together as agonists during the entire hip and knee extension movement. Despite the fact that lengthening characteristics of rectus femoris, semitendinosus, biceps femoris, and gastrocnemius during co-activation revealed an ability of these muscles to act as agonists about one joint, there existed no evidence supporting the paradoxical functioning of these muscles during the weighted deep-knee bend movement as theorized by Lombard (1903) and Molbech (1965).

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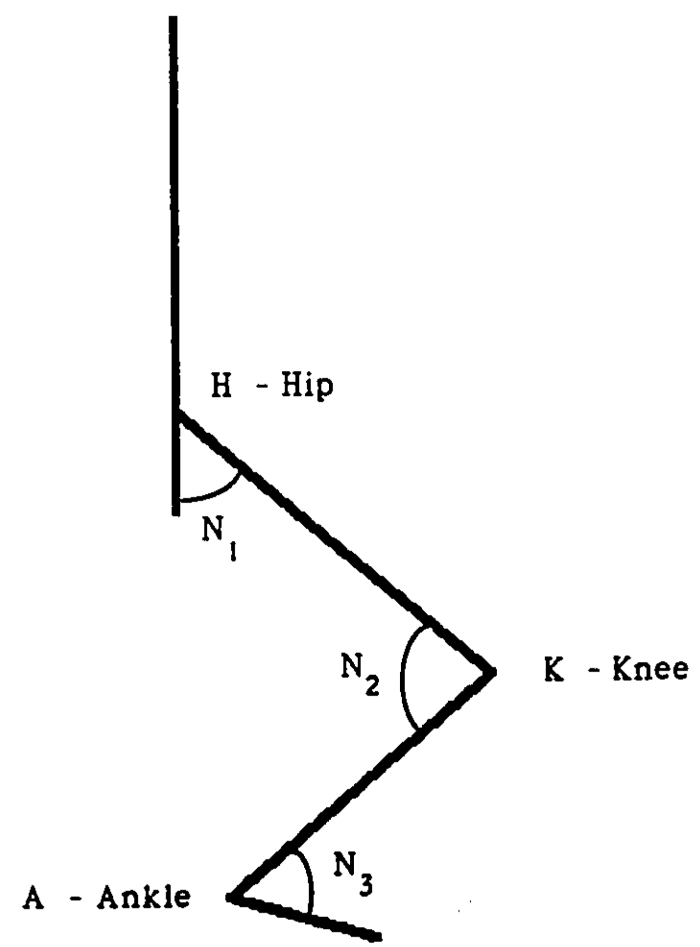
Appendix A

Model used for muscle length calculations

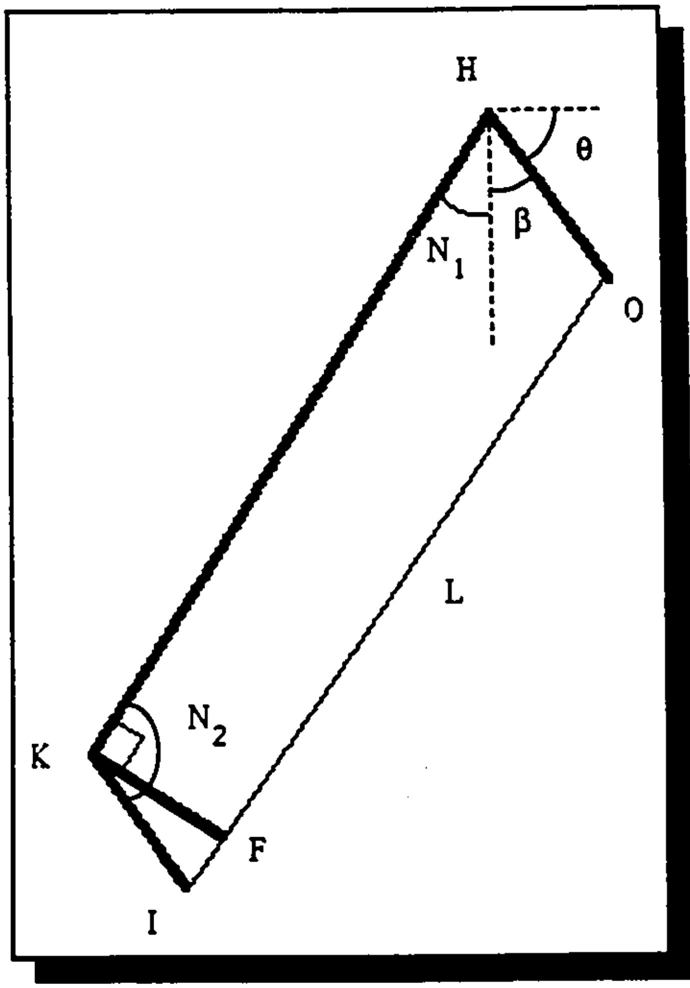
This appendix contains the models and equations used to calculate the lengths of the biceps femoris, gastrocnemius, gluteus maximus, rectus femoris, semitendinosus, soleus, tibialis anterior, and vatus lateralis. These were derived from a model presented by Frigo and Pedotti (1978) and modified by Hubley (1981).



RHT = Standing height of subjects in centimetres.

Reference model for muscle length determination.

Biceps Femoris



O = origin: ischial tuberosity.

I = insertion: head of the fibula.

θ = 62°

β = 28°

OH = 6.3

KF = 2.5

KI = 6.7

HR = 4.6

HK = 0.2320·RHT

$\angle OHK = N_1 + \beta$

$OK = \sqrt{((HK)^2 + (OH)^2 - 2\cdot(HK)\cdot(OH)\cdot(\cos\angle OHK))}$

$\angle OKH = \text{Arcsin}\ (OK / (\sin(\angle OHK)\cdot(OH))$

$\angle OKI = N_2 - \angle OKH$

$OI = \sqrt{((OK)^2 + (KI)^2 - 2\cdot(OK)\cdot(KI)\cdot(\cos\angle OKI))}$

$\angle FKI = N_2 - 90°$

$\angle FIK = \text{Arcsin}\ (OI / (\sin(\angle OKI)\cdot(OK))$

$\angle IFK = 180° - \angle FKI - \angle FIK$

$DB = \sin(\angle FIK)\cdot(KI) / (\sin\angle IFK)$

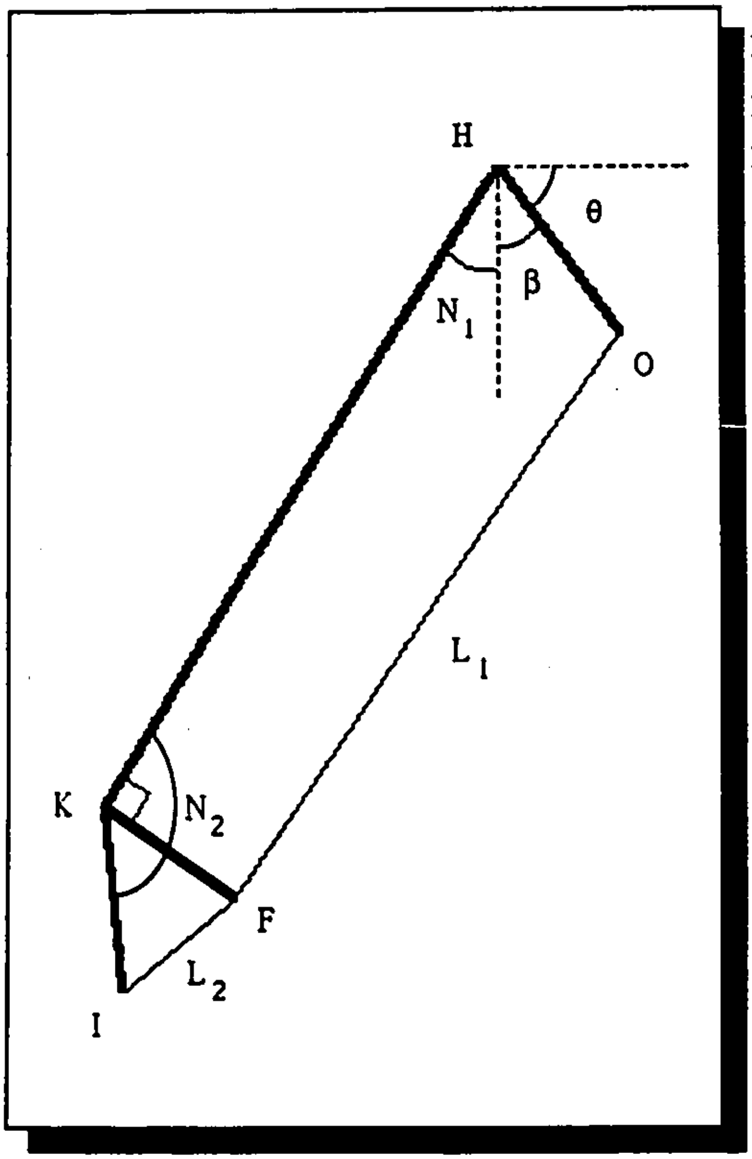
IF DB ≥ KF Then

L = OI

Else

(cont'd)

Biceps Femoris (cont'd)

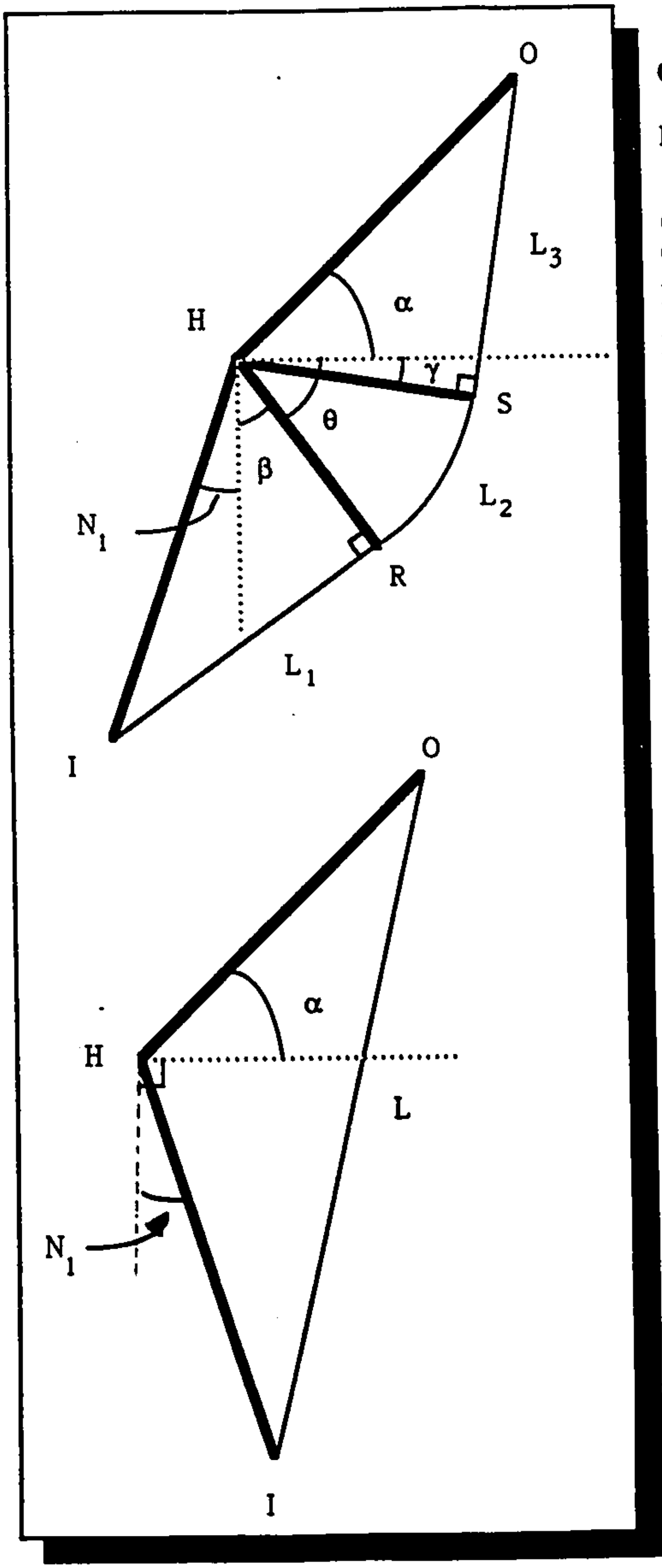


$L_2 = \sqrt{((KF)^2 + (KI)^2 - 2\cdot(KF)\cdot(KI)\cdot(\cos \angle FKI))}$

$L_1 = \sqrt{((OK)^2 - (KF)^2)}$

$L = L_1 + L_2$

Gluteus Maximus



O = origin: posterior of the sacrum.

I = insertion: posterior shaft of femur.

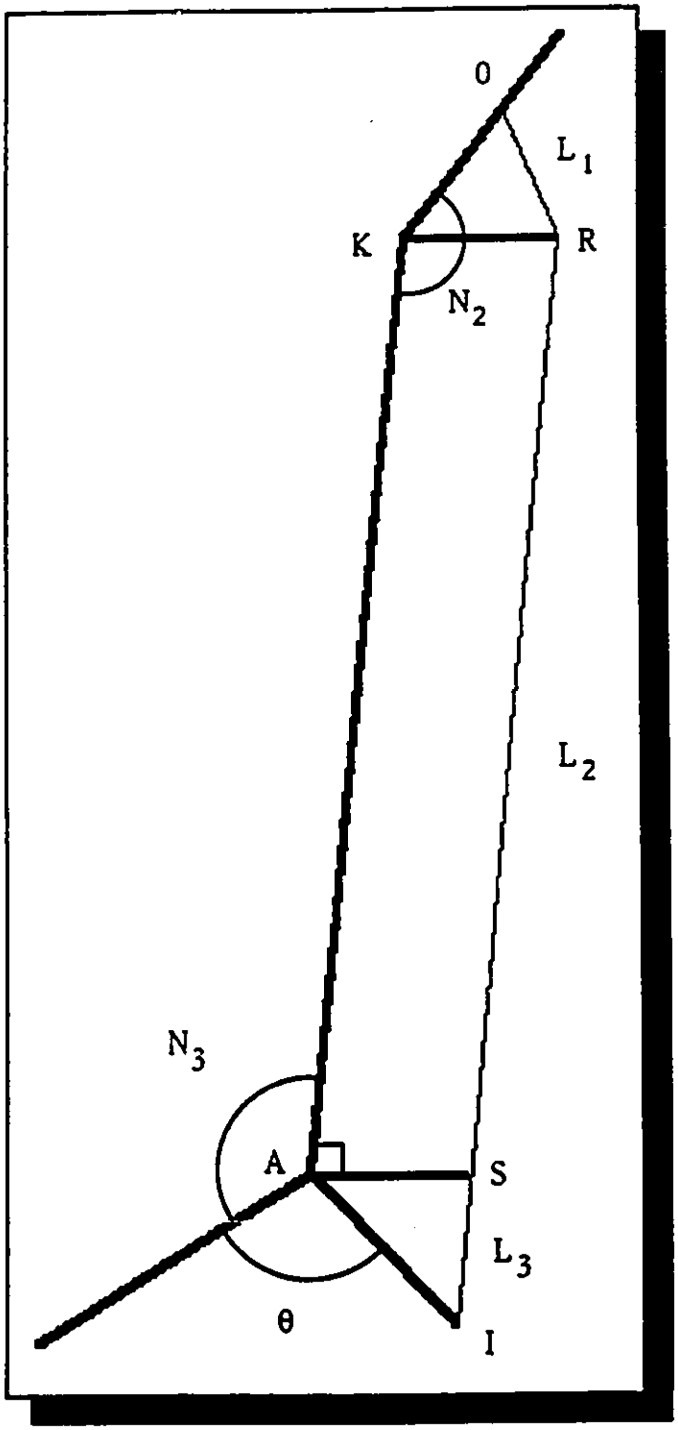
α = 43°
OH = 8.8
HI = 14.8
HS = 4.6
HR = 4.6

$\angle OHS = \text{Arccos}(HS / OH)$
$\gamma = \angle OHS - \alpha$
$\angle IHR = \text{Arccos}(HR / HI)$
$\beta = \angle IHR - N_1$
$\theta = 90° - \beta$
$\angle SHR = \theta - \gamma$

If $\angle SHR > 0°$ Then
$L_1 = \sqrt{((OH)^2 + (HS)^2)}$
$L_2 = (\angle SHR / 360°)\cdot(2\pi)\cdot(HS)$
$L_3 = \sqrt{((HI)^2 + (HR)^2)}$
$L = L_1 + L_2 + L_3$
Else
(cont'd)

$\angle OHI = \alpha + 90° - N_1$
$L = \sqrt{((HI)^2 + (OH)^2 - 2\cdot(HI)\cdot(OH)\cdot(\cos \angle OHI))}$

Gastrocnemius



O = origin: posterior head of the femoral condyles.

I = insertion: tendo Achillis on the calcaneus.

$KR = 0.0218◊RHT$

$AI = 0.0376 \cdot RHT$

$AS = KR$

$\theta = 120°$

$L_2 = KA = 0.2470 \cdot RHT$

$\angle OKR = N_2 - 90°$

$L_1 = (\angle OKR / 360°) \cdot (2\pi) \cdot (KR)$

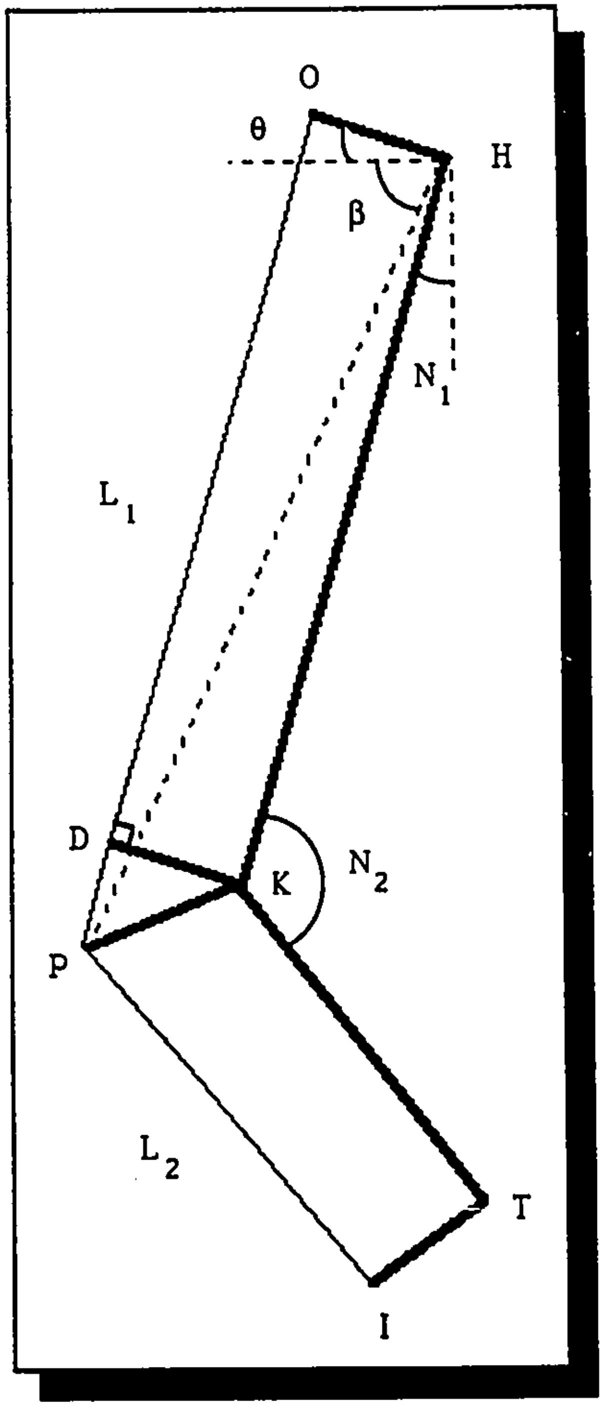
$\angle IAK = 360° - \angle TKP - N_3$

$\angle IAS = \angle IAK - 90°$

$L_3 = \sqrt{((AS)^2 + (AI)^2 - 2 \cdot (AS) \cdot (AI) \cdot (\cos \angle IAS))}$

$L = L_1 + L_2 + L_3$

Rectus femoris



O = origin: anterior inferior spinous process of the ilium.

I = insertion: patellar tendon.

$\theta = 42°$

$\angle TKP = 83°$

$L_2 = 0.0455 \cdot RHT$

$HK = 0.2320 \cdot RHT$

$PK = 0.0242 \cdot RHT$

$OH = 0.0273 \cdot RHT$

$DK = 0.0212 \cdot RHT$

$\angle HKP = 360° - N_2 - \angle TKP$

$PH = \sqrt{((HK)^2 + (PK)^2 - 2 \cdot (HK) \cdot (PK) \cdot (\cos \angle HKP))}$

$\angle PHK = \text{Arcsin}\left(\dfrac{(PK) \cdot \sin(\angle HKP)}{PH}\right)$

$\beta = 90° - \angle PHK - N_1$

$\angle OHP = \beta + \theta$

$OP = \sqrt{((OH)^2 + (HP)^2 - 2 \cdot (OH) \cdot (HP) \cdot (\cos \angle OHP))}$

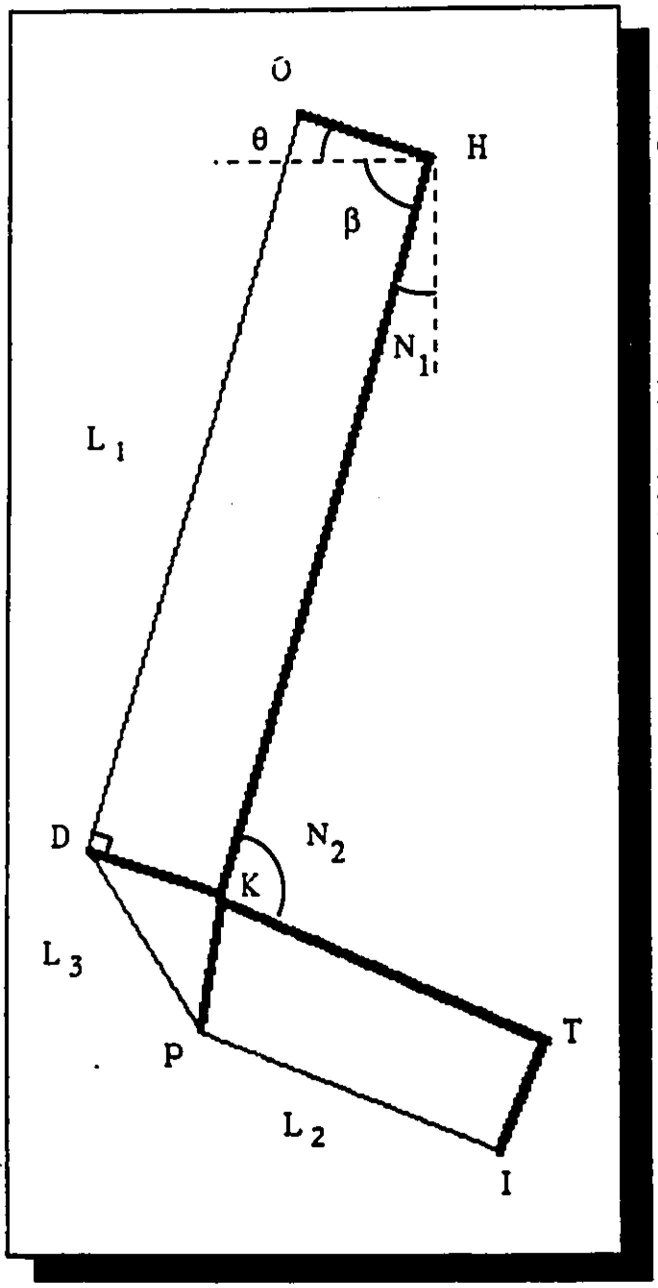
$L_1 = OP$

If $N_2 \geq 151°$ Then
$L = L_1 + L_2$

Else

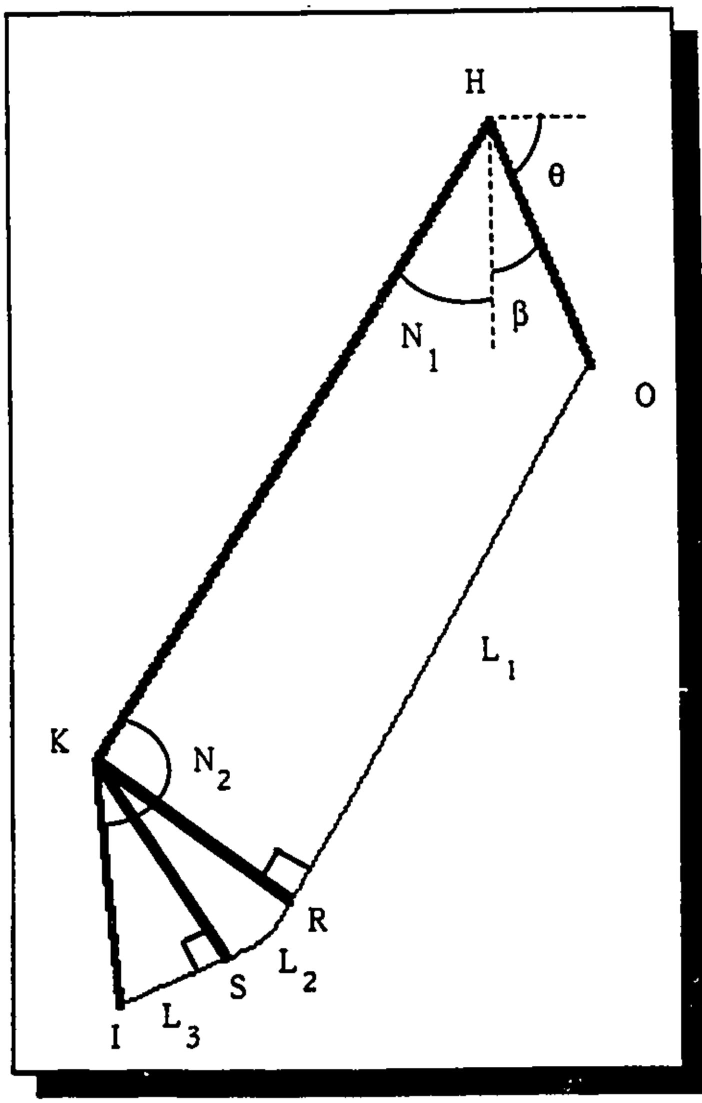
(cont'd)

Rectus Femoris (cont'd)



β	$= 90° - N_1$
$\angle OHK$	$= \beta + \theta$
OK	$= \sqrt{((OH)^2 + (HK)^2 - 2 \cdot (OH) \cdot (HK) \cdot (\cos \angle OHK)}$
$\angle OKH$	$= \text{Arcsin}\left(\dfrac{(OH) \cdot \sin(\angle OHK)}{OK}\right)$
$\angle OKD$	$= \text{Arccos}\,(DK / OK)$
$\angle HKD$	$= \angle OKH + \angle OKD$
L_1	$= \sqrt{((OK)^2 - (DK)^2)}$
$\angle DKP$	$= 360° - \angle TKP - N_2 - \angle HKD$
L_3	$= \sqrt{((DK)^2 + (PK)^2 - 2 \cdot (DK) \cdot (PK) \cdot (\cos \angle DKP))}$
L	$= L_1 + L_2 + L_3$

Semitendinosus



O = origin: ischial tuberosity.

I = insertion: tibial shaft.

θ = 60°

β = 30°

HO = 5.7

KI = 12.8

KR = KS = 2.5

$\angle OHK = N_1 + \beta$

$OK = \sqrt{((HK)^2 + (OH)^2 - 2\cdot(HK)\cdot(OH)\cdot(\cos \angle OHK))}$

$L_1 = \sqrt{((OK)^2 - (KR)^2)}$

$L_3 = \sqrt{((KI)^2 - (KS)^2)}$

$\angle IKS$ = Arccos (KS / KI)

$\angle SKH = N_2 - \angle IKS$

$\angle OKR$ = Arccos (KR / OK)

$\angle HKO$ = Arcsin (OK / (sin ($\angle OHK$)· (OH))

$\angle HKR = \angle HKO + \angle OKR$

$\angle RKS = \angle SKH - \angle HKR$

IF $\angle RKS > 0°$ Then

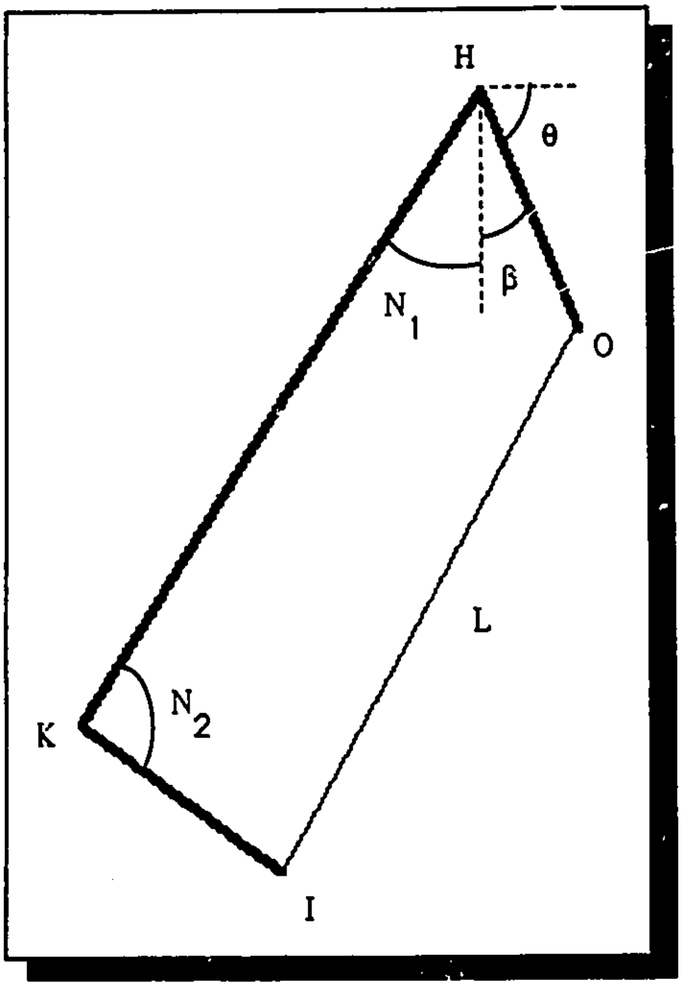
$L_2 = (\angle RKS / 360°)\cdot(2\pi)\cdot(KR)$

$L = L_1 + L_2 + L_3$

Else

(cont'd)

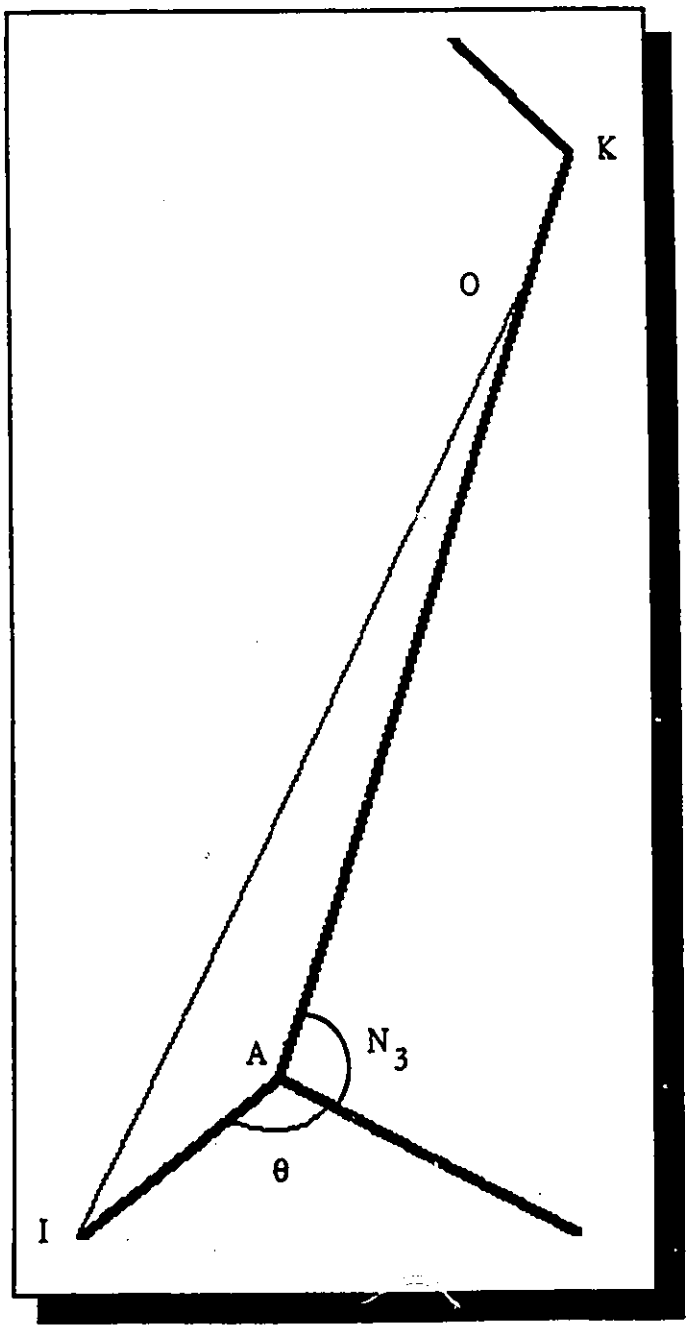
Semitendinosus (cont'd)



$\angle OKI = N_2 - \angle HKO$

$L = \sqrt{((OK)^2 + (KI)^2 - 2 \cdot (OK) \cdot (KI) \cdot (\cos \angle OKI))}$

Soleus



O = origin: posterior head of the fibula and oblique line of the tibia.

I = insertion: tendo Achillis on the calcaneus.

OA = 0.1920 RHT

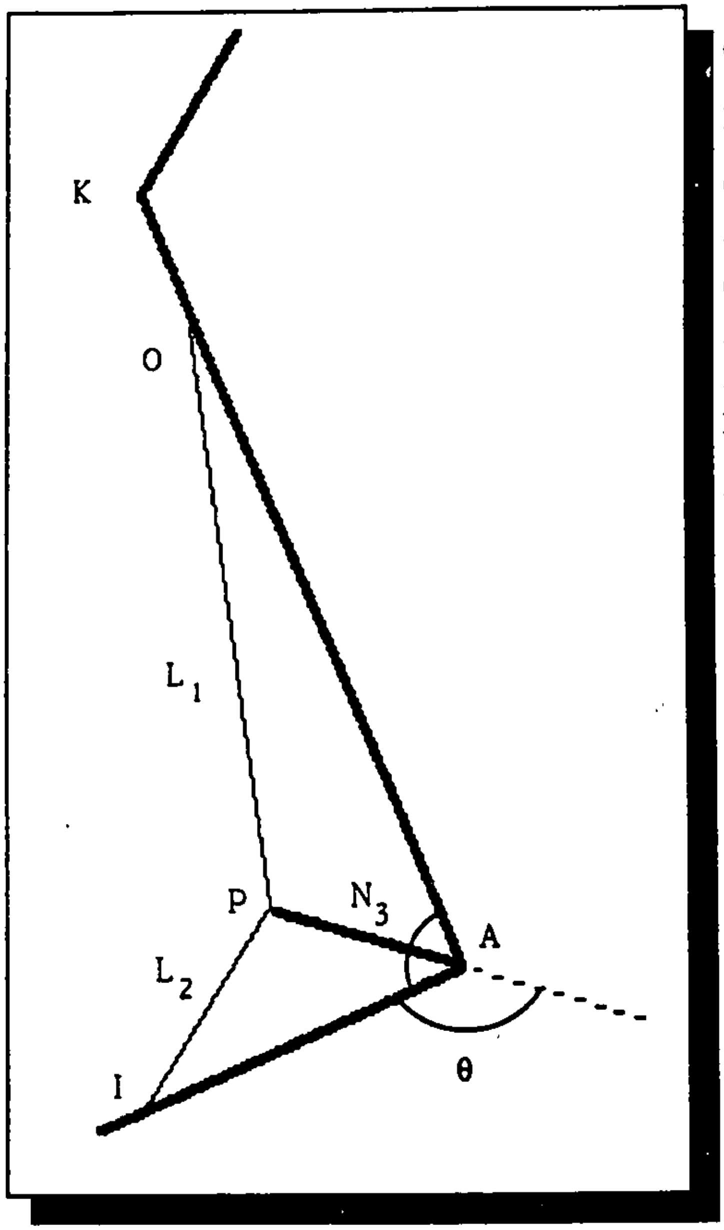
AI = 0.038·RHT

θ = 120°

$\angle OAI = 360° - N_3 - \theta$

$L = \sqrt{((OA)^2 + (AI)^2 - 2\cdot(OA)\cdot(AI)\cdot(\cos \angle OAI))}$

Tibialis Anterior



O = origin: lateral surface of the tibia

I = insertion: base of the first metatarsal

OA = 30.7

AI = 7.0

θ = 120°

AP = 4.0

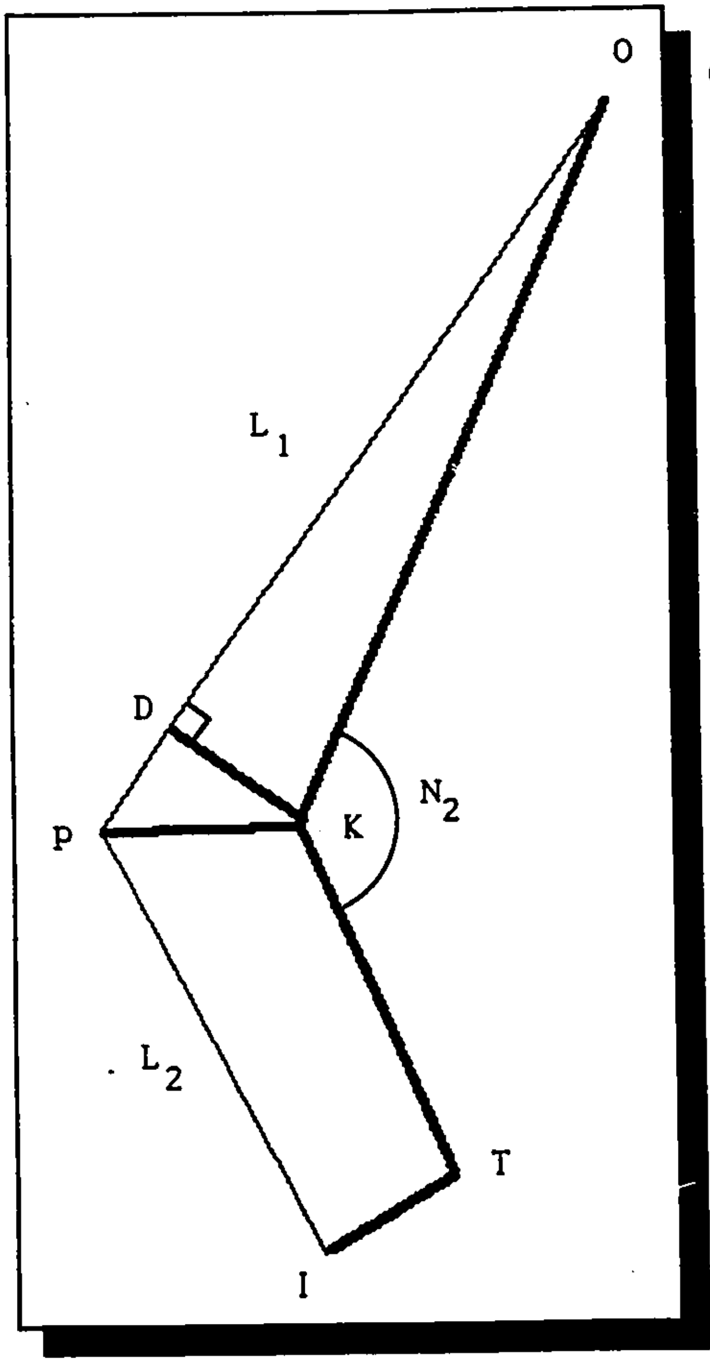
$\angle OAP = \angle PAI = N_3 / 2$

$L_1 = \sqrt{((OA)^2 + (AP)^2 - 2\cdot(OA)\cdot(AP)\cdot(\cos \angle OAP))}$

$L_2 = \sqrt{((AI)^2 + (AP)^2 - 2\cdot(AI)\cdot(AP)\cdot(\cos \angle PAI))}$

$L = L_1 + L_2$

Vastus Lateralis



O = origin: upper half of the anterior inter-trochanteric line, outer lip of the linea aspera.

I = insertion: patellar tendon.

$\angle TKP = 83°$

$L_2 = 0.0460 \cdot RHT$

$OK = 0.1520 \cdot RHT$

$KT = 0.0520 \cdot RHT$

$PK = 0.0180 \cdot RHT$

$DK = 0.0212 \cdot RHT$

$\angle OKP = 360° - \angle TKP - N_2$

$OP = L_1$

$L_1 = \sqrt{((OK)^2 + (PK)^2 - 2 \cdot (OK) \cdot (PK) \cdot (\cos \angle OKP))}$

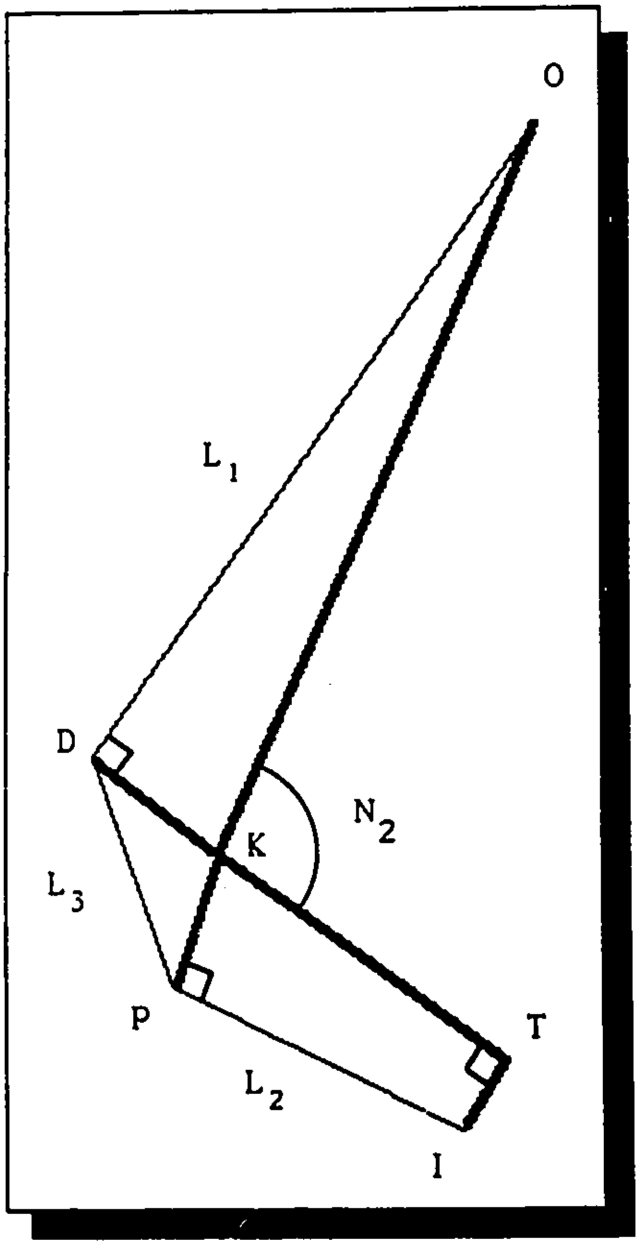
If $N_2 \geq 151°$ Then

$L = L_1 + L_2$

Else

(cont'd)

Vastus Lateralis (cont'd)



$\angle OKD = 82°$

$\angle DKP = 360° - \angle TKP - N_2 - \angle OKD$

$L_3 = \sqrt{((DK)^2 + (PK)^2 - 2\cdot(DK)\cdot(PK)\cdot(\cos \angle DKP))}$

$L = L_1 + L_2 + L_3$

Appendix B

Review of Literature

Review of Literature

Overview of the chapter

To cover all of the material relevant to this research experiment the following topics will be discussed in the order cited. To describe the development of two-joint muscle research, a general review of two-joint muscle function in relation to different classification systems will be described in the initial pages of this chapter. Current findings in deep-knee bending movement kinematic, kinetic, and electromyographical studies will be reviewed subsequently to aid in the description of the expected observations. Justification of the experimental techniques to be used in this experiment, such as, indirect muscle-tendon-unit length determination and electromyographical data collection, will be developed and discussed in the ensuing sections. Muscle length-force and EMG-force relationships will be discussed within these two sections to justify certain assumptions made in the experimental design.

Two-joint muscles

The first attempt at describing two-joint muscle function was reported by Lombard (1903). Through observations made on the extensor mechanisms of a frog leg, Lombard proposed that two-joint muscle function about its joints was dependant on activity of other musculature and the magnitude of the moment arm at each joint. More specifically, he predicted that if one joint was fixed, contraction resulted in agonist activity about the second. When both joints were free to move, the dominant role assumed by the muscle was to create the movement observed at the joint where the muscle assumed the largest moment arm (figure 13.). In the case of the frog, Lombard suggested that a closed path of energy exchanges existed between two-joint muscles during the extension of the hip, knee, and ankle. The flow of energy within this path, a figure eight pattern, was said to be provoked in the direction of the greatest muscle moment arm.

Aware of the differences in musculature geometry of humans, Lombard proposed that these principles were applicable to the biceps femoris and rectus femoris in man during walking. He later on suggested that the simultaneous hip and knee extension observed during walking was produced from simultaneous contraction of the two-joint hamstring and quadriceps muscles. Although this theory was never validated, the function of these particular muscles could be truly determined only by means of monitoring muscle moment arm and muscle length changes during the entire hip-knee extension movement.

Validation of Lombard's two-joint muscle extensor theory as applied to the frog was attempted subsequently by Lombard and Abbot (1907). By stimulating muscles independently and in pairs, it was concluded that co-contraction of two-joint muscles could produce extension at the hip and knee joints.

a)

b)

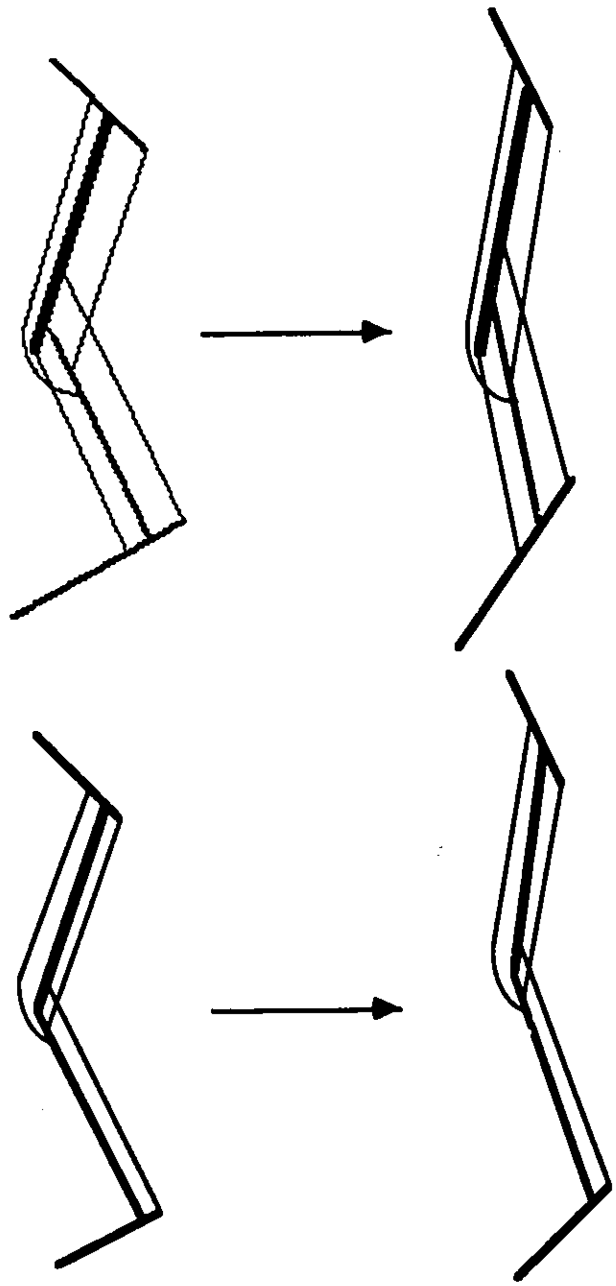


Figure 13. Simultaneous extension of the lower limb joints by the biarticular muscles in (a) frog and (b) human.

To determine neuromuscular mechanisms of lower limb two-joint muscles in human movement, Markee, Logue and Williams (1955) observed local muscle activity in anesthetized and spinally transected dogs. Local muscle length changes and movements about joints were recorded as medial and distal nerve branches of each muscle were individually stimulated. The authors concluded that certain two-joint muscles acted as being reciprocally inhibited. That is to say, different sections of a muscle could contract at different times and in a opposing manner when recruited simultaneously. According to these findings, when both joints were free to move, two-joint muscles could be recruited to act as one-joint muscles without producing resistance at the other joint. Moreover, it was possible at any instant to observe simultaneous concentric activity of medial fibres and eccentric contraction of distal fibres or vice-versa. The investigators proposed that non-homogeneous contraction could be possible in man due to similarities in nerve branching of these muscles.

In an effort to determine nervous control of the lower limb two-joint muscles and to confirm Markee et al.'s (1955) findings in man, Basmajian (1957) utilized EMG signals from medial and distal fibres of selected muscles. Activity was monitored with a series of indwelling electrodes while subjects performed selective knee flexion and hip extension. According to the similarities between medial and distal EMG signals of each muscle during knee flexion and hip extension, the investigator concluded that the two-joint muscles contracted homogeneously as a unit, producing equal magnitudes of force at both joints.

In a similar study, Basmajian and Fujiwara (1975) monitored EMG signals from indwelling electrodes placed in fibres of the rectus femoris, medial hamstrings, iliopsoas, vastus medialis, and gluteus maximus. To isolate these muscles during singular and combined movements, a special "pulley" apparatus was designed in which each subject was placed in a supine position and flexed the right knee and hip to 90 degrees. Controlled forces were applied to the thigh and shank in directions which would cause hip and knee flexion or extension. All of the ten male subjects were then asked to perform a series of combined hip and knee movements. In disagreement with the previous findings of Markee et al. (1955), this study revealed that muscle fibres within human two-joint muscles contracted homogeneously throughout the entire length of the muscle.

Further advances in knowledge of neuromuscular coordination of the lower limb two-joint muscles were also made as a result of the Basmajian and Fujiwara (1975) study. When these muscles were monitored during monoarticular movements, the medial hamstrings solicited a greater EMG response during knee flexion than hip extension. Greater EMG levels were acquired from rectus femoris during knee extension when compared to hip flexion. According to these results, medial hamstrings were designated as being primarily knee flexors while rectus femoris was denoted as a knee extensor. In instances where the two-joint muscles were presumed to act as double agonists, the authors observed that the antagonistic two-joint muscle behaved as though reciprocally inhibited. In this case, medial hamstrings were inhibited during simultaneous extension of the knee and hip flexion and rectus femoris was inhibited during simultaneous knee flexion and hip extension.

It is interesting to note that Basmajian and Fujiwara (1975) reported inhibition of the medial hamstrings during simultaneous hip and knee extension, contradicting predictions made by Lombard (1903) for this movement. However, three out of the ten subjects in the study did in fact show hamstring activity during the performed movement and therefore did not completely eliminate the consideration of Lombard's (1903) hypotheses.

In a differential study of the hip extensor muscles, Flynt (1971) revealed differences in activities of semitendinosus and biceps femoris during hip extension with the knee flexed and with the knee extended. Semitendinosus activity was predominant during the former while biceps femoris during the latter. Furthermore, biceps femoris was discovered to be a very strong lateral rotator of the femur while semitendinosus was dominantly active throughout medial rotation. Even though the medial hamstrings and the biceps femoris are two-joint muscles that cross the posterior of the hip and knee joints, the evidence presented by this investigator indicated differing recruitment patterns during variations in hip movement. Hence, this study suggested that the medial and lateral hamstrings should be considered separate mechanical entities.

Utilizing a similar apparatus to that of Basmajian and Fujiwara (1975), Kazai, Kumomoto, Yamashita, Maruyama, and Tokuhara (1978) attempted to determine the role of the antagonistic two-joint muscles in certain leg movements and obtained in part conflicting results. In their investigation, rectus femoris and biceps femoris IEMG were recorded from a sample of 5 men asked to provide isometric resistance during a series of several leg movements. A system of pulleys was utilized to load each muscle in agonistic-antagonistic fashion at both the hip and knee joints. Both muscles were tested under three general conditions; (1) during monoarticular movements where they were considered agonists, (2) during two-joint movements where they functioned as double agonists, and (3) during two-joint movements where each muscle assumed an agonistic role at one joint and an antagonistic role at the other. When IEMG was plotted as a function of external load, the rectus femoris muscle was predominantly a knee extensor and biceps femoris a knee flexor. During concurrent knee and hip extension, rectus femoris demonstrated reciprocal inhibition as external load on the hip joint increased. Biceps femoris activity patterns varied widely among subjects during the execution of the hip and knee flexion. In the authors summary, it was concluded that the role of the two-joint muscles had no set pattern and were subjected to variation as a result of increased loading at each joint.

Since muscle properties during static conditions vary widely from dynamic conditions, it must be understood that conclusions based on static (isometric) movements may not be applicable to dynamic conditions in which Lombard's (1903) predictions were considered. Unfortunately, having supplied much added insight into two-joint muscle function and their nervous recruitment under isometric contraction, the application of results from these studies to dynamic conditions are limited.

Function of two-joint muscles in dynamic conditions

In the first reported study of weighted deep-knee bending, Basmajian, Harin, and Regenas (1972) analyzed electromyographical signals from the four heads of the quadriceps. According to the report, simultaneous contraction of the biceps femoris and rectus femoris was prominent during the hip-knee extension movement. The ascent was characterized by a decrease in EMG levels of each muscle from initiation to termination. Rectus femoris activity showed a relatively late onset and ceased at a relative knee angle of approximately 130°. EMG levels of rectus femoris and biceps femoris were greatest at relative knee angles of 90°. Vastus lateralis showed greater levels of activity than rectus femoris and biceps femoris at all instances of the movement.

The mechanics of two-joint muscle function during simultaneous hip and knee extension while rising from a chair were first analyzed by Rasch and Burke (1959). When simultaneous contraction of the two-joint hamstring and quadriceps muscles was considered, and since both muscles produced opposing effects about both joints, it was thought that the effort of each muscle would mutually neutralize each others action at both the hip and knee joints. However, when a mechanical model of the lower limb was used to simulate the effects of a two-joint muscle contraction, simultaneous hip and knee extension was observed. This seemingly contradictory situation was named Lombard's paradox. As concluded by Lombard (1903), this result was attributed to differences in muscle moment arms about each joint. Hence, during the rising from a chair movement, the hip flexor effect of the rectus femoris was overcome by the hip extensor effect of the hamstrings while at the knee, flexor effects were overcome by the rectus femoris' knee extension effect. This particular action was supported by the results of a study conducted by Elftman

(1939) who reported that hamstring and rectus femoris muscle moment arms were greater for extension than flexion at both the hip and knee, respectively.

Rasch and Burke (1978) also discussed the possibilities in which one of the two-joint muscles involved could behave in a string or belt-like fashion. It was suggested that energy from one segment was transferred to the segment of muscle insertion when one of the muscles contracted isometrically or was naturally taut. In the example used, if the rectus femoris was replaced by a cord and the hamstrings were made to contract, the hip extension produced by the hamstrings would pull on the rectus femoris resulting in extension of the knee. Since the leverage arm for knee extension was greater than that of the hamstrings for knee flexion, the knee flexion effect caused by contraction of the hamstrings was said to be nullified by the tendon-like action of the rectus femoris about the knee.

Using a model in which one-joint and two-joint muscles were included as contributors to the hip and knee movements, Rasch and Burke (1978) also presented examples where co-contraction of the the gluteal muscles with the two-joint hamstrings and quadriceps produced simultaneous hip and knee extension. In these instances the observed joint motion was said to be accomplished by means of transmitting energy from the trunk to the tibia via "taut" two-joint hamstring muscles and quadriceps muscles. Under these conditions it was assumed that the two-joint muscle group was active since muscle relaxation would prevent function of such a mechanism.

Modelling of two-joint muscles to predict their function

Mathematical modeling of the lower limb two-joint muscles was first reported by Molbech (1965). In an attempt to explain two-joint muscle role during steered movements of the leg, Molbech analyzed the lower limb action using a geometrical model. As seen in figure 14, the mechanical model consisted of an articular system of two equal length rods, representing the femur and tibia, attached to each other by a single hinge type pin joint which served to simulate rotation at the knee. The hip joint was represented by a slider-crank mechanism free to move along a smooth guide in the vertical (y) direction while the inferior joint of the distal rod, the ankle, was modeled as a hinge joint attached to the ground. The direction of a muscle force inserted on the superior segment was represented by a line connecting points C' and C. In this system, movement about the knee joint was governed by the direction of pull of the muscular force with respect to the direction of a line drawn from the instantaneous centre of rotation of the superior rod to point C, the insertion point of the muscle. Molbech denoted this line as being the limiting line (i.e., line OL) for the effect of forces acting on C since forces applied along OL would result in no rotation of the superior segment. Forces acting in directions to the right or to the left of the limiting line would result in knee flexion and extension, respectively. This was determined by the direction and magnitude of the force component normal to OL .

Using the limiting line as the criterion, it was predicted that the action of the two-joint hamstring muscle force would change from knee flexion to knee extension as the relative knee angle surpassed a mathematically determined critical angle. Molbech estimated from his calculations that this angle was approximately 135 ° of the posterior relative knee angle. Since gastrocnemius muscle geometry in man could be represented by the line C'C, he concluded that this muscle extended the knee at relative knee angles greater than 135°. Moreover, the same was said of the biceps femoris and semitendinosus when the mechanical system was oriented in such a way that the superior joint and the inferior joint represented the ankle and hip, respectively.

Although this outlook was believed to be theoretically justified, Molbech himself admitted that the model was simple and did not describe the exact behavior of these muscles in man. He recommended that the model be modified to include superior and inferior rods of differing lengths, as well as, a mechanism that provided a better simulation of the knee articulation movement

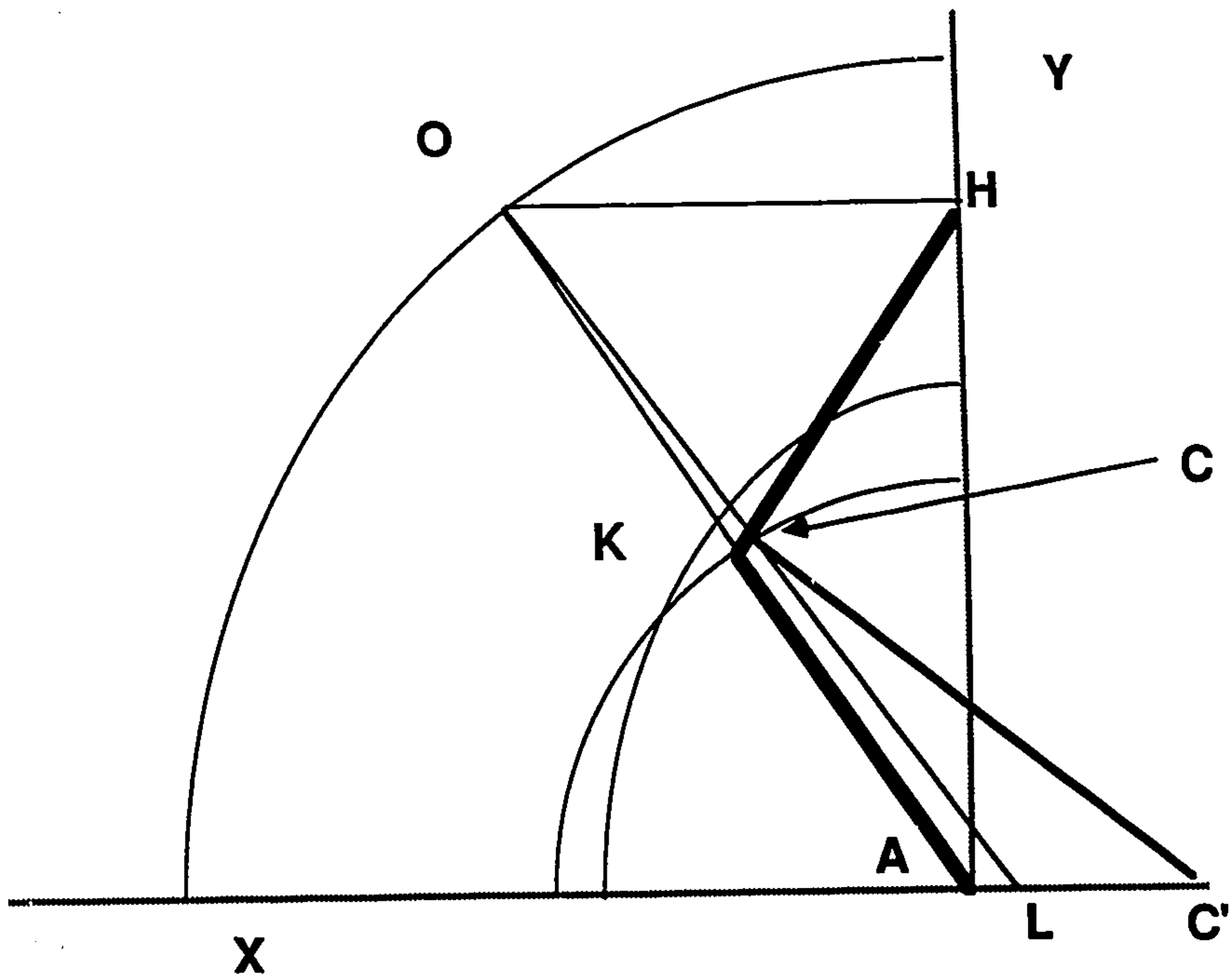


Figure 14. Molbech's instantaneous center of rotation model.

In order to determine roles of two-joint muscles in certain multi-joint movements and in part to justify Molbech's (1965) previous work, Carlsoo and Molbech (1966) examined electromyographical data from two-joint muscles during free movements of the leg, deep-knee bending, and cycling. Raw EMG signals were collected from certain leg muscles including rectus femoris, biceps femoris, sartorius, and gastrocnemius.

When the EMGs were collected during the free movements, rectus femoris was primarily a knee extensor while biceps femoris was primarily a hip extensor. Although this behavior was predicted by Lombard (1903), Carlsoo and Molbech's interpretations differed significantly During deep-knee bending, biceps femoris, rectus femoris and gastrocnemius were observed to contract simultaneously until a certain knee angle was attained. At this point, biceps femoris activity diminished to zero. In the explanation of this observation, the phenomenon was attributed to crossing of the biceps femoris line of pull over the limiting line where its role would change from knee extensor to knee flexor. Hence, it was concluded that at angles greater than 135°, gastrocnemius, biceps femoris and rectus femoris acted synergistically to produce knee extension. At angles smaller than 135° the flexing effects of gravity were counteracted by rectus femoris and other monoarticular knee extensors. Similar conclusions were arrived at in the case for cycling, however, all of the subjects did not solicit such muscular activity patterns.

Intrigued by the predictions made by Molbech (1965), Stanhope (1982) conducted an electromyographical validation of the model during a four second unloaded squat to stand movement. Prior to testing, it was hypothesized that Molbech's (1965) two-joint muscle theory could be supported by observing an increase in biceps femoris activity and a decrease in rectus femoris and vastus lateralis activity during extension of the knee. Twenty-five male subjects whose thigh and shank were equivalent in length were chosen for the experiment. Surface EMGs from the rectus femoris, biceps femoris, and vastus lateralis, as well as, the voltage signal from an electrogoniometer positioned about the knee were simultaneously collected during squatting movements. EMG signals were then integrated through digitization and normalized to 100% maximal voluntary contraction EMG levels (NIEMG). Squatting was performed on a pivoting platform built to eliminate possible effects of the gastrocnemius on the knee joint. The movement was separated into three phases by breaking down the relative knee angle progression ; Phase #I corresponded to 105°-125° of relative knee angle , Phase #II, 125°-145°, and Phase # III 145°-165°.

Despite obtaining very low NIEMG values for the biceps femoris, the experiment revealed co-contraction of biceps femoris, rectus femoris, and vastus lateralis throughout all phases. As movements progressed, the investigator reported a significant decrease in rectus femoris and vastus lateralis activity. No significant change was observed for biceps femoris. From these results, Stanhope (1982) concluded that there existed no electromyographical evidence in support of Molbech's (1965) theory during the squatting movement.

Andrews and Hay (1983) criticized the model used by Molbech from which predictions of two-joint muscle role were derived. As listed in the critique, one of the major flaws of Molbech's (1965) proposal was the lack of instruction in indicating whether the segment on which the muscle force acted was the one where the muscle originated, or the one where it inserted. In the example cited, the authors indicated that biceps femoris could be considered as being a both a knee extensor and knee flexor simultaneously. The equivalent problem was also encountered when monoarticular muscles were considered. A more detailed list and explanation of the anomalies in Molbech's model are revealed in the article.

Realizing the numerous discrepancies in Molbech's (1966) muscle function prediction approach, Andrews (1985) devised a model in which muscle role function was determined based on the kinematic behavior of a linked system subjected only to the action of the muscle of interest when released from rest. In this system, the algebraic sign of the system's generalized velocity at some time shortly after the system is released from rest was used to establish the functional role of the multiple-joint muscle.

In validation of this method, the response of a one degree of freedom mechanical model was observed upon release from several differing configurations. According to the investigator, this method produced results that were considerably different to those derived by Molbech (1965).

Although it was believed that Andrews (1985) model provided biomechanists with a relatively simple means of predicting function of two-joint muscles, its use was still considered to be quite limited since the process of testing muscle function at large numbers of initial rest conditions proved to be very lengthy. Hence, a kinematic based model utilizing Equilibrium analysis techniques was later proposed by Andrews (1987).

Summarizing the description given in the report, the classification of muscle function in this method was based on the assumption that changes in the configuration of a mechanically representative system could only result from changes in length of the muscle in question. More specifically, this method established the functional role of a muscle according to 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 (figure 15.), 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 (i.e., absolute angle of tibia, crank angle, etc..). The arbitrary 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 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 undetermined.

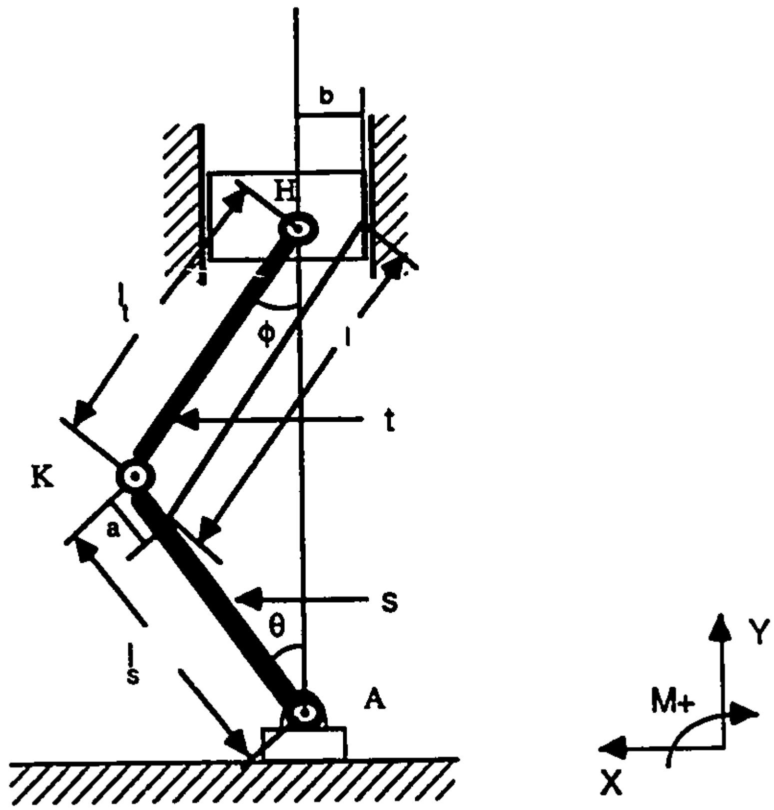
Subsequently, Andrews conducted an investigation to describe the functional roles of the hamstrings and quadriceps at the hip and knee during cycling as determined by this method. In vivo conditions were simulated with a mechanical link system in which joint centres of rotation were modeled as points and the muscles were modeled as straight lines. Anthropometrically determined femur and tibia lengths, as well as, quadriceps and hamstrings muscle origin and insertion points were included in the mechanical representation to ensure simulation of real life conditions.

The results of this investigation revealed that the quadriceps generally functioned non-paradoxically at both the hip and knee during cycling. In contrast, the hamstrings generally acted paradoxically at the knee and exhibited essentially equal regions of paradoxical and non-paradoxical behavior at the hip. Although it was possible to model the squatting movement in this manner, no attempts of this have been reported to date.

Function of two-joint muscles in other activities

Despite the very small number of reported investigations on deep-knee bending, added insight into two-joint muscle function can be gained by examining their activities during similar movements. Although some of the movements discussed in the following are more dynamic than deep-knee bending, each one is characterized by significant periods of simultaneous hip and knee extension.

Kelley, Dainis and Wood (1976) observed simultaneous biceps femoris and rectus femoris activity while studying the mechanics and muscular dynamics of rising from a seated position. During the task, subjects were allowed to lean forward with their trunks and rise from a seat in a physically unrestricted manner. Greater levels of EMG activity were recorded during the early stages of the movement. These levels diminished as the movement proceeded to an easy standing position where muscle activity was fundamentally at baseline values.



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

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

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

Figure 15. Example application of the Andrews Kinematic classification method.

Since the take off phase of sprinting was characterized by simultaneous hip extension, knee extension, and ankle plantarflexion, previous investigators had suggested that the lower limb two-joint muscles would act in synergy to produce this movement. However, according to Simmonsen, Thomsen and Klausen (1985), no evidence of Molbech's phenomena was evidenced in sprinting. In their study, which involved monitoring muscle EMG activity, muscle lengths and muscle moments from four male track athletes during competition-like sprinting trials, rectus femoris lengthened throughout the whole ground phase and no simultaneous gastrocnemius, rectus femoris and biceps femoris activity was noted prior to or at takeoff. Nonetheless, one would question the application of Molbech's predictive methods to running since the hip is not constrained to simple vertical movement.

Simmonsen's et al. (1985) experiment also revealed that two-joint muscles were submitted to larger variations in muscle contraction when compared to monoarticular muscles during sprinting. More specifically, all two-joint muscles showed longer periods of eccentric activity when compared to concentric activity throughout the cycle. The authors attributed this fact to initial limb deceleration and elastic energy storage for execution of the subsequent movement. When observed to behave in this manner, two-joint muscle function at running speed was described as being very efficient. Similarly, work done by Elftman (1966) has suggested that the presence of multi-joint muscles in man involved a potential for increased efficiency.

In determination of total and segmental energy exchange during running, Elftman (1966) found a stage in the forward swing of the leg, shortly before it contacted the ground, in which hip extensors were required to do positive work at the same time as the knee flexors were required to decelerate the extension of the knee. It was rationalized that at this moment the hamstring muscles received kinetic energy from the momentum of the lower leg at the knee joint while at the same time they were expending energy at the hip joint. Two-joint muscles were claimed to be more efficient at performing this task since they could apply energy derived from the knee to the hip joint. One-joint muscles could also effectuate this role, however, the energy received at the knee joint would be wasted. According to his calculations, the work of walking required an expenditure of 2.61 h.p. by the limb muscles while 3.97 h.p. would be needed if only single joint muscles were used.

Since cycling was considered to be a constrained movement, previous investigators (Carlsoo and Molbech, 1966) proposed that the propulsive thrust generated during the the first 180° of the pedaling cycle was produced by simultaneous co-contraction of the lower limb two-joint muscles. Although previous electromyographic studies (Houtz and Fischer, 1959; Despires, 1974; Goto et al., 1975) have shown that co-contraction of hamstrings and quadriceps was evident during this phase in cycling, further information was required to determine muscle role during the movement.

In an attempt to provide an integrated description of cycling, Gregor, Cavanagh, and Lafortune (1985) also revealed co-contraction of knee flexors and extensors during periods of extensor activity. Net joint moment data revealed that the second half of the thrust phase was characterized by simultaneous hip and ankle extensor moments, and a flexor moment at the knee. When EMG information was coupled with joint moments, the authors concluded that conditions leading to Lombard's paradox were not evident in cycling. In their explanation, it was said that the hamstring muscles generated the flexor moment measured during thrust to resist the extensor moment produced by the external force exerted on the foot by the pedal at this time. The fact that the knee was extending would lead one to assume that an extensor moment was produced at the knee, when in fact the net moment was flexor. In a closing statement, the authors suggested that muscle activation pattern in this case was modified as to resolve the problems posed in Lombard's paradox.

Muscle function classification systems

According to Andrews and Hay (1983), the manner in which an investigator distinguished between paradoxical and non-paradoxical two-joint muscle function at a joint was influenced by many factors. These included (1) the method used to determine when the muscle was active,(2) the determination of the criterion used to establish the existence of paradoxical behavior, and (3) the classification method used to establish the functional role of the muscle. Despite having gained a greater understanding of recruitment patterns of the two-joint muscles in various movements, the problem of agonist-antagonist and flexor-extensor classification procedures used to identify paradoxical function has created much confusion about when, where, and if this phenomenon does occur in the lower limb extension movement.

One of the earliest attempts to define a system for classifying muscle function has been that proposed by Rasch and Burke (1959). According to their definitions, a muscle was said to function as an agonist to the movement when it underwent a concentric contraction while an antagonist was simply a muscle whose contraction tended to produce a joint action exactly opposite to that of another specified muscle. Under these same guidelines, a muscle functioned as a stabilizer when it contracted isometrically to anchor a bone in providing a firm base of pull for another muscle. Although these definitions were quite simple, this system did not take into account the conditions in which a muscle acted eccentrically to: (1) control a movement against gravity such as the biceps brachii during controlled extension in lowering the hand, or (2) to prevent joint soft tissue injury such as in biceps brachii function during a rapid extension of the elbow.

Realizing this problem, Andrews (1985) devised a joint kinetic based classification method in which the functional roles of both one-joint and two-joint muscles could be determined systematically. This system was appropriately named the Standard Kinetic (SK) method of classifying muscle function.

As a prerequisite to using the SK method, joint centres were modeled as specific points which at the same time acted as the point of origin of an orthogonal reference system whose axes were coincident to those considered in flexion/ extension, adduction/ abduction , and internal rotation/ external rotation as seen externally. Individual muscle moments M_m whose magnitude were determined by the cross product of the exerted muscle force and the moment arm about the specific joint centre, were considered with respect to (A) the resultant moment M_m produced about the joint centre by all muscle forces acting on the concerned segments or (B) the resultant moment M_F about the joint centre of all forces acting on the concerned segments. The use of either of the two moments described, produced fundamentally the same results. However, since the description given in instance (B), M_F, corresponds to that moment which is measurable externally (i.e., the net joint moment), further explanations will be made with reference to this term.

According to the SK method, a muscle was described as being an agonist to the movement about each respective axis if and only if the algebraic sign of its individual moment M_i was the same as that of the net resultant muscle moment M_F. Conversely, antagonist function was assigned if the algebraic sign of the individual muscle moment was opposite to that of the net resultant moment.

Furthermore, primary, secondary and tertiary roles of each muscle were determined by comparing magnitudes of the three vectorial components, m_x, m_y, and m_z of the individual muscle moment. As an example of the use of this classification method, consider the actions of a muscle m_i about a joint centre J in the x direction (flexion= positive, extension= negative), the y direction (abduction= positive, adduction = negative) , and the z direction (internal rotation = positive, external rotation = negative). If the individual components of the moment created by m_i about J were the following;

M_i= resultant moment created by muscle m_i about J.

$= m_x\, i + m_y\, j + m_z\, k$

and

$$m_x = -10\ \mathrm{N{\cdot}M}$$

$$m_y = 1\ \mathrm{N{\cdot}M}$$

$$m_z = -5\ \mathrm{N{\cdot}M}$$

while the simultaneous resultant muscle moment MM_j of all forces about J was ;

$$MM_j = MM_x\, i + MM_y\, j + MM_z\, k$$

where

$$MM_x = 20\ \mathrm{N{\cdot}M}$$

$$MM_y = -5\ \mathrm{N{\cdot}M}$$

$$MM_z = -4\ \mathrm{N{\cdot}M}$$

Muscle m_i in this instance would be defined as being primarily an antagonist-extensor, secondarily an agonist-external rotator, and a tertiary antagonist-adductor of the segments about J. In the case of two-joint muscles, individual muscle and net force moments would be computed about the centre of rotation of the second joint and the above procedure repeated.

In contrast, the classification scheme used by Carlsoo and Molbech (1966) in their validation study was based on EMG and joint kinematics. According to the method of analysis chosen, a muscle acted as an agonist (antagonist) if the mechanical action it produced was concurrent (opposed) to the joint motion. Electromyographic data was collected to determine state of activity and joint relative velocity was measured to indicate joint motion. These measurements were in turn combined in analysis to determine if a muscle was acting in favor or against the joint motion. Consequently, a muscle was said to be a joint flexor or extensor if it was active while flexion or extension of that joint was being observed.

Problems in Carlsoo and Molbech's (1966) function classification method became evident when it was assumed by the investigators that the resultant joint moment was concurrent to the joint velocity. Hence, if the algebraic sign of the knee relative velocity was positive and this corresponded to knee extension, then it was assumed that a net joint extensor moment was in effect. According to the theory behind net joint moment calculations reviewed by Winter (1979), evaluation of the direction of joint moment with reference to joint velocity was not valid and hence, presented anomalies in the interpretation.

Although the Standard Kinetic based classification system presented a very suitable method of systematically determining muscle function at a given joint, problems with its use have occurred when paradoxical function of two-joint muscles was considered. As mentioned previously, paradoxical function defined by Lombard (1903) was characterized by isometric contractions of antagonistic two-joint muscles. Since the SK method did not take into account the level of recruitment (i.e., EMG activity) and instantaneous length changes of the muscle, paradoxical function of the hamstring muscles could not be detected with this method. Therefore, EMG activity and muscle length velocity guidelines should have been included in the SK classification method. This modifications would be required to create a system in which muscle function is described in a more detailed manner. Although such modifications have been included in the proposed investigation, no such modifications have been reported, to this date in the literature.

While Lombard (1903) was the first to describe conditions in which two-joint muscles functioned paradoxically, there exists no solid evidence that this paradox is in fact existent during dynamic movements. The majority of related studies have concluded from electromyographical analyses.alone that paradoxical activity of lower limb musculature exists during simultaneous hip and knee extension Despite the limited gains of insight into two-joint muscle function generated by the initial studies, this information has been plagued by many discrepancies in the results and methods used. Molbech (1965) proposed a model that predicted joint configurations at which function of the two-joint hamstring and gastrocnemius muscle changed from antagonistic to paradoxical during simultaneous hip and knee extension. However, as pointed out by Andrews and Hay (1983), researchers were prevented from using this model due to the discovery of several major inconsistencies. In a review of muscle paradoxical function, Andrews (1985) declared that discrepancies in interpretation of results were due to several factors, including the muscle function classification method used. A systematic joint kinetic based method (i.e. the SK method) for determination of muscle function was then proposed. Although the results obtained from this method provided a more precise description of two-joint function about its associated joints, it was concluded that the method alone could not be used to detect paradoxical function of muscles as described by Lombard (1903). Finally, it was proposed that the EMG and instantaneous muscle length change be incorporated within the kinetic based method to aid in detection of muscle paradoxes, as well as, provide a more detailed description of muscle function.

Biomechanics of weighted deep-knee bending

It should be brought to the attention of the reader that weight lifting is a general term which encompasses many varying types of lifting techniques. Some of these are the dead lift, the snatch lift, the power clean, the parallel and non parallel squat, etc. While all techniques are related by the common goal of lifting a weighted bar, they are considered to be mechanically discrete as a result of the differences in the symmetry, sequencing, and patterns of lower limb movement. Being aware of these differences, we can not justifiably transgress conclusions about the mechanics of other much investigated weight lifting techniques to that of deep-knee bending (i.e., commonly known as squatting). Despite the lack of related literature, only parallel squatting research will be discussed in the ensuing discussion.

Kinematics of deep-knee bending

A squat or knee bend is a movement characterized by a bending of the knees and lowering of the upper body to a desired level followed by the return to a straight standing position. With a weighted iron bar resting on the shoulders, this activity is frequently used as an exercise to strengthen the lower part of the body, specifically the quadriceps (McLaughlin, Dillman and Lardner, 1977).

The movement consists of two phases, the descent and ascent. During the descent there is simultaneous flexion of the knee and hip joints. This action is continued until the desired depth of the squat is achieved. In international squatting competition, the depth of the squat is defined as the point at which the tops of the thighs are below parallel with the platforms (International Powerlifting Federation rules, 1974). Once the desired depth is achieved, the performer simultaneously extends the knee and hip joints until an easy standing position is assumed.

In effort to compare highly skilled and non-skilled squatting technique, McLaughlin et al. (1977) developed a kinematic model for power lifters performing a parallel squat. Kinematic data was obtained from 24 male internationally classed performers from which linear and angular velocities and accelerations of the trunk, thigh, shank, and bar were derived. Vertical bar velocity was chosen as the modelling criteria in formulation of the kinematic model. Movement speed in terms of joint angular velocities were not reported. The kinematic analysis indicated that bar vertical velocity behaved generally in a sinusoidal fashion throughout the squatting cycle. Peak bar velocity during the ascent and descent were approximately 0.5 $m \cdot s^{-1}$ (1.6 $ft \cdot s^{-1}$). Perhaps the most significant finding was the locating of a period during the ascent phase where bar velocity dropped to a minimum. This period, labeled the "sticking point", occurred at a posterior knee angle of 105 degrees, and was found to be consistent for all performers at different barbell loads.

Kinetics of deep-knee bending

In the initial standing position of the squat, there is minimal amount of stress placed on the muscles and ligaments (Steindler, 1955). This can be explained through examination of the line of gravity in respect to the skeletal system. If the line of gravity of any body segment falls through its supporting joint, the lever arm is zero and gravitational torque non-existent.

At the onset of a squat movement, a gravitational moment is applied about the supporting segment joint system resulting from the anterior or posterior displacement of the lines of gravity of the body segments. To maintain equilibrium in this position, muscular forces must act to create a net moment equal in magnitude and opposite in direction of that created by gravity. Descent occurs when the muscular moment is less than that of gravity whereas the ascent is created when the summation of muscular moments are greater than that of gravity.

The greatest gravitational moments are created when the line of gravity falls furthest from the joint(s) centre(s). In the squat this genc.ally occurs when the maximum depth is achieved. This however, is not necessarily the case at all joints. The human body is comprised of a series of segments which interact by influencing the forces of each other.

In an investigation into the kinetics of the knee joint during deep knee bend exercises with heavy loads (load varied from 375 to 650 lbs), Ariel (1974) reported ankle, knee, and hip, dorsiflexor, extensor, and extensor moments, respectively throughout the entire movement cycle. Joint moments were derived via cinematographical techniques and joint reaction forces were determined from moment arm data obtained from digitization of several X-ray negatives (i.e no force platform was used). Compressive and shear forces were revealed to be greatest during initial flexing of the knees. When the knee bend was performed with a bounce at maximum depth, joint reaction forces and extensor moments were greater at the instant of bounce when compared to the no-bounce condition. It was concluded from this study that the bouncing movement posed a greater threat to the knee articular system as a result of greater joint reaction forces.

Using the same performers as in their 1977 study, McLaughlin et al (1978) analyzed the kinetics of the parallel squat. Results from this study were in part agreement with those published by Ariel (1974). As shown in figure 16, peak moments for the ankle were flexor while that for knee and hip were found to be extensor. The joint moment histories indicate that descent phase is in general characterized by sinusoidally fluctuating extensor moments about all three joints. A slight exception to this is seen at the ankle where the extensor moments were very low initially and then changed to flexor for a brief period of time. Extensor moments were assumed thereafter for the remaining duration of descent.

As expected, peak extensor torques for the hip and knee were observed during ascent. The former appeared during the initial phase while the onset of the latter appeared during late ascent. Ankle and hip moments signatures behaved in a parallel fashion, however, transition to flexor moments about the ankle occurred at a slightly faster rate and the moment remained flexor for a longer period of time after the transition. The magnitude of all joint moments diminished to zero in the late phase of the ascent.

In the comparison between skilled and less skilled performers, McLaughlin et al. (1978) found the less skilled athletes to have a greater hip extensor and knee flexor torque during the ascending phase of the task. He attributed these moments to the greater forward lean exhibited by the less skilled performers. By leaning forward, the less skilled athletes placed a greater force on the trunk extensors (i.e. hamstrings). This increased force was said to be transmitted through the hamstrings to the knee joint where it became flexor dominant. The result of this transmitted force was a decrease in extensor dominant thigh torques about the knee.

One must question the validity of McLaughlin et al.'s (1978) results, since in a review of the investigator's techniques revealed that modelling of the linked system did not include horizontal reaction forces and the location of the centre of pressure on the foot. In the lower limb model used, ground reaction forces were configured to be acting through the ankle axis of rotation and not as a distributed force on the foot, hence the lower limb was essentially represented by a two segment system. While omission of the foot segment in mathematical modelling of the kinetics of activities such as cycling have been justified (Gregor et al, 1985), this procedure is highly questionable in that of the squatting movement due to the critical mechanical role played by the plantarflexors and dorsiflexors during this motion. Furthermore, due to the interdependence of the moments applied to the distal joint of a segment with those applied at the medial joint, this fault in technique would not only falter ankle moment data, but would also significantly influence moments measured at the knee and hip as well.

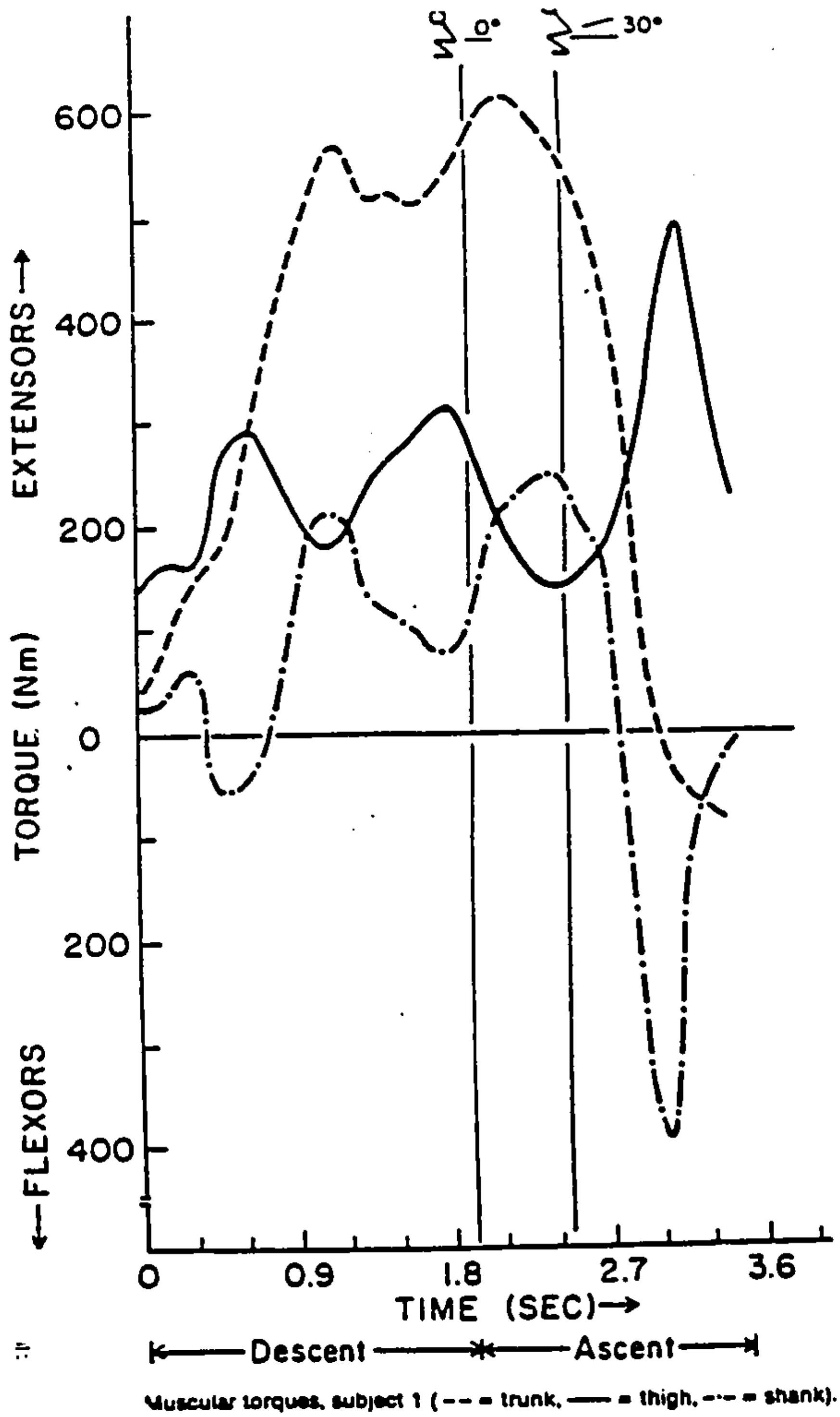


Figure 16. Lower limb joint moment data reported by McLaughlin (1978)

Despite these criticisms, McLaughlin reported that the moment data was in agreement with that of Ariel (1974) and Garhammer's (1976) moment data on the snatch lift. This is not surprising since these investigators also ignored measurement of the horizontal foot-ground reaction force as well as the location of the centre of pressure on the foot .

Regarded as a very similar movement, the mechanics and muscular dynamics of rising from a seated position were investigated by Kelley et al. (1976). Unlike the previous studies, Kelley's subjects performed the task without carrying an additional load. The subjects starting position was sitting with a posterior knee angle of 75 degrees. The task consisted of rising to an easy standing position in the specific rise time of 2 seconds. The muscular torques (extensor) for the knee and hip joints were at maximum values once knee and hip extension began. As the movement progressed, they diminished smoothly until an easy standing position was achieved.

To date there exist no valid data of net joint moment analysis of the squatting movement. However, according to the general findings of these studies, the ascent phase of squatting seems to be characterized by extensor moments at both the hip and ankle while that of the knee seems to vary between extensor and flexor. While many investigations exist for the snatch or clean lifts, data from these studies can not be used to extrapolate to squatting conditions due to the wide discrepancy in the dynamics of the movements.

Electromyography of deep-knee bends

In an analysis of the olympic squat and power cleans, Lehr and Poppen (1979) monitored the electromyographical activity of the vastus lateralis, erector spinae, latissimus dorsi and trapezius muscles in one experienced and two inexperienced subjects. Data revealed that vastus lateralis activity was significant, as a knee extensor, throughout three phases of the squat clean; (1) initial pull, (2) second pull, and (3) catching the bar and rising to the upright position. Since phase three (3) was considered to be essentially equivalent to the ascent of squatting one could predict significant activity of this muscle during the deep-knee bend movement.

In a comparison of experienced to non-experienced athletes, significant differences in sequencing and configuration of the raw EMG were apparent. However, this contrast was not apparent in the vastus lateralis but was only restricted to that of the upper girdle muscles, trapezius and latissimus dorsi. As expected, the EMG activity of all muscles increased in intensity between highly loaded and lightly loaded lifts.

As described previously, Stanhope (1982) collected surface EMGs from the rectus femoris, biceps femoris, and vastus lateralis during squatting movements from a sample of 25 inexperienced subjects. Despite the very low NIEMG values of biceps femoris, the experiment revealed co-contraction of biceps femoris, rectus femoris, and vastus lateralis throughout all phases.

In critique of his own work, Stanhope suggested that perhaps the use of a pivoting platform during the squatting trials resulted in changes of muscle recruitment pattern from that of normal squatting. In this instance, the disturbance in anterior-posterior posture equilibrium created by the pivoting platform produced a mechanically different situation in which the squatting movement was to be performed. Lack of external loading was also mentioned as another possible intervening factor. Significantly different results may have been revealed in loaded squatting movements.

Cerquiglini, Figura, Marchetti and Salleo (1973) studied the change in gastrocnemius and quadriceps femoris (did not specify which one) raw EMG signal characteristics of weight lifters during and after power squat training. Initial tests revealed that both of these muscles were highly active throughout the movement when the load lifted was 80% of the subject's personal maximum. After a two month training period, the authors claimed that a higher voltage and frequency of EMG could be obtained from isometric contractions of the two lower limb muscles. These conclusions were arrived by means of a spectral frequency analysis of the raw EMG signal.

Despite the lack of electromyographical data and the different techniques used by existent studies, one can with most certainty predict that co-contractions of the hip, knee, and ankle extensor muscles will be existent throughout the squatting cycle. Co-contraction of the lower limb two-joint muscles during simultaneous hip, knee, and ankle extension has instigated much research into the mechanics of these muscles during locomotion and a wide variety of other activities.

Muscle-tendon-unit length measurement

Muscle length modelling techniques have been subject to much investigation in the recent decade as a result of the increased interests in modelling and simulation of human kinetic parameters. Many methods have been utilized in muscle force modelling; most of these being concerned with muscle forces about the hip and knee joints.

Some of the earliest reported work performed to determine muscle line-of-action was that of Inmann (1947) using wires to represent hip abductor muscles. Similarly, Merchant (1965) studied hip adductor muscle forces by attaching strain gauge mounted chains to a dried male articulated pelvis and femur. Although muscles have been represented by many other types of devices, elastic bands (Sorbie and Zalter, 1965), springs and force transducers (Evans, 1961), these were considered limited due to differences in material characteristics. As a result of the demand for a more representative modelling technique, researchers have progressed towards using geometrical techniques of modelling. The straight line and centroidal line (Jensen and Davey, 1975) modelling methods form the majority of these.

When quantifying muscle-tendon-unit lengths under the straight line theory, the muscle's line of action is modelled as a series of straight and curvilinear lines joining the midpoint of the areas of the origin and insertion. Centroidal line muscle modelling is more complex since it represents muscle force line of action as the locus of muscle cross-sectional centroids between the centres of the areas of muscle attachment on the two adjacent joint segments.

In one of the primary muscle length studies performed in walking, Morrison (1970) approximated the change in length of certain lower limb extensor and flexor groups by the product of the angular displacement and the perpendicular distance from the joint centre to the line of action. The majority of muscles were represented by a straight line joining the assumed origin and insertion, however, in the case of the gastrocnemius certain adjustments in the model had to be made due to the deflection of this muscle over the condyles of the femur. Changes in the model were also included for the quadriceps to compensate effects induced by patellar movement. Although the methods utilized did not allow for inter-subject comparison, the main investigator did not comment on the accuracy of the method used.

Using a straight line model of the human lower limb, Seireg and Arvikar (1973) attempted to determine forces exerted by 29 muscles about the hip, knee, and ankle joints in the sagittal plane. According to their complex geometry and multiple movement effects, some muscles were represented by more than one line of action. Muscles such as those crossing the ankle joint were modelled as two adjoining lines. Although the authors claimed that this model was much more realistic than in previous studies, the calculations derived were considered quite limited since muscle origin-insertion points were obtained from two dimensional skeleton illustrations. Other investigations such as Dostal (1979), and Dostal and Andrews (1981) have utilized textbook drawings to obtain this information.

In a study of muscle kinematics during various locomotor activities, Frigo and Pedotti (1977) utilized straight line modelling to derive equations for eleven muscle lengths of the lower limb. The lower limb was represented by a three link system in which muscle lengths were trigonometrically expressed as a function of the joint-associated relative angle (hip, knee, and ankle). Intrinsic measurements were derived from dry skeleton data. The instantaneous muscle-tendon-unit length was determined by inputting the relative angular orientation(s) of the joint(s) into the model.

Despite not having normalized the skeletal data, the authors reported that the errors inherent to the model were negligible. They also noted that errors increased with increased angular variations of the limb but did not present explanations. In effort to reduce variability, Frigo and Pedotti's (1977) equations have been adjusted using the data of five male cadavers (MacKinnon, personal communication). This method has been applied to various movement studies with apparent success (Frigo et al., 1979; Hubley, 1981; Simonsen et al, 1985).

Although straight line models have been considered very practical, Jensen and Davey (1975) have questioned the validity of these models. They suggested that a centroidal-line model represented a more valid measurement of muscle-tendon-unit length than the straight-line model since it accounted for the curved paths assumed by muscles. The results of a comparative investigation revealed that some of the orthogonal components of the hip muscle moment created by gluteus medius, rectus femoris, and sartorius muscle were up to 50% higher in the straight line model than those derived centroidally. Moreover, straight line estimates of the resultant hip moment were found to be consistently lower (up to 12%). Despite lacking a control comparison technique, the results derived from centroidal techniques were claimed to be more representative because the determined muscle paths were qualitatively more consistent with direct observation.

Even though the centroidal technique has given appearance of being more realistic, Andrews and Hay (1983) presented several arguments condemning its use. They claimed it much less desirable than straight line modelling because (1) it required a large data base containing numerous muscle configuration parameters, (2) it could produce unreasonable and non-unique results due to difficulties in clearly defining a truly transverse cross section, and (3) it was based on cadaver data where lack of muscle tone may have been significantly different from in vivo measurements.

Despite the claims made by those in favor of the centroidal technique, investigators continued to use straight line representation of muscle force in three dimensional applications. Seireg and Arvikar (1975) modified their previous model of the lower limb (Seirig and Arvikar, 1973) to include three dimensional origin-insertion point data of 31 muscles. Deviations in the muscle path created by other musculature and bony prominences were accounted for in this model. Once again the accuracy of this method was subject to scrutiny for its use of origin and insertion coordinates obtained from the author's earlier work.

In response to the lack of accurate muscle moment arm data, Brand, Crowninshield, Wittstock, Pedersen, Clark and van Krieken (1982) have used radiographic negatives to develop a systematic means of determining three dimensional muscle points of origin and insertion of 47 lower limb and pelvic muscles. Given certain anthropometrical measures taken from the individual, the locations of these points and straight line muscle length were determined through mathematical transformation. Segment specific reference systems, defined from radiographically identifiable landmarks, permitted the calculation of muscle length given any orientation of the segments. A sensitivity study showed that errors in estimating most muscle moment arms as a result of mismarking origins and insertions by 10 mm were generally less than 10%. The greatest average errors occurred in those moment arms determined about the ankle and knee. A right to left side comparison of origin and insertion coordinates revealed an average difference of approximately 15 mm. One of the sources cited as being a major contributor to these errors was the estimation of the effective location of a muscle when a straight line from the origin to insertion did not truly represent the muscle's action. Despite these errors, the authors have stated that it was a suitable method of approximating the muscle origin, insertions, and moment arms in living subjects.

Subsequently, in the design of a muscle force prediction model, Pierrynowski (1982) presented a method of measuring lower limb muscle length from three dimensional coordinates of palpable bone landmarks. Coordinate data obtained from a single disarticulated cadaver were normalized to femur length and transformed to fit the configuration of those segments on a living subject by means of segment specific reference systems. Muscles were generally modelled as straight lines from the centroid of the area of origin to that of the insertion. In conditions where a muscle was constrained to pass, four additional points were defined at these locations. Once established, these points were joined by combinations of straight and curvilinear lines. Straight lines were used whenever a muscle ran freely from point to point, and a curve was fitted when a muscle was forced to alter its course. Muscle length was then obtained by summation of the lengths of these individual sections. In the case of the quadriceps muscles, the patella and the tibio-patellar ligament were modelled as a rigid body of constant length. This length was determined for each subject from in vivo sagittal plane X-ray photographs of male knee joints reported by Morrison (1967). Under-estimation of moment arms about the hip, knee, and ankle joints due to underlying musculature was compensated for by means of digitizing individual muscle layers on the cadaver and extrapolating these values to the subject through transformation.

In discussion of the co-comitent error terms included in this method, Pierrynowsky (1982) suggested that the errors in performing the three dimensional transformations presented the greatest contribution. Although no effort was made to quantify the errors in length measurement, the system was said to have a maximum absolute error of 5 mm in location of the skeleton palpable points. Specialized three dimensional digitizing equipment and lengthy anthropometric measuring sessions limited the application of this method to practical kinesiological laboratory settings.

In summary, the complexity of the musculoskeletal system has resulted in several representations of the muscle-tendon-unit (MTU). These geometrical models include the straight line and centroidal models, and models in two and three dimensions. While the centroidal and three dimensional models allow for greater accuracy, they require more complex data acquisition and reduction. This representation, therefore is limited as investigators must maintain a balance between accuracy and simplicity. It is for this reason and others stated previously that Frigo's and Pedotti's (1977) straight line model was chosen for MTU length measurement in the proposed investigation.

The EMG signal and its relationships to muscle function

Description of an EMG signal

The end result of transmission of a nervous action potential via Alpha motor neurons in the production of a muscle twitch is the depolarization of the transverse tubular system and the sarcoplasmic reticulum, commonly described as the Muscle Action Potential (m.a.p.). In this process the motor end plates of the associated nerve branches can be described as being the electrochemical epicentre of the depolarization "wave" which propagates along the direction of the muscle fibres. This depolarization wavefront and the subsequent repolarization wave are the phenomena "observed" by the recording electrodes. Subsequently, the raw EMG signal is simply an amplification of the summation of motor action potentials (m.a.p.'s) as sensed by the recording electrodes. The raw signal can be treated by varying types of amplification circuits to produce various versions of the signal (i.e. rectified EMG, linear envelope, and IEMG) which are considered more practical for research purposes.

In general, EMG recording electrodes can be classified as being of the surface or the indwelling type. The much more popular surface electrode consist of approximately 1 cm diameter metal discs, usually silver/silver chloride, and are used to detect the average activity of superficial muscles. The signal recorded from these electrodes are influenced to a great extent by the electrode surface area, and the perpendicular distance between the electrode and the muscle in question. In the case of bipolar systems, inter electrode distance is of major importance. On the other hand, indwelling electrodes are nothing more than a fine hypodermic needle with an insulated conductor (electrode) located inside and bared to the muscle tissue at the open extremity of the needle; the needle itself forms the other electrode. These electrode types are required to record from deep muscles as is done in the assessment of fine movements.

Although surface electrodes are more commonly used in contemporary biomechanics investigations, most early electromyographic studies have been performed with intrinsic fine wire electrodes. Indwelling techniques were believed to be more direct measurements of muscle activity and seemed to be sufficiently reliable. In an attempt to determine reliability of this technique, Jonsson and Reichmann (1968) reported that the reliability of the indwelling measurement was far superior between recordings made at 15-20 minute intervals than when made between separate experiments.

Due to the lack of reliability between test sessions, increased time of preparation, and possible health risks imposed on the subject, surface electromyography was soon adopted by many investigators. As it was revealed by Komi and Buskirk (1970) and Zuniga, Truong, and Simons (1970), the surface electrode measurement proved to have greater repeatability between tests separated by intervals of significantly long duration.

Zuniga, Truong and Simons (1970) studied the effects of mono-polar skin electrode positioning on the amplitude of averaged EMGs from the biceps brachii during elbow flexion and forearm supination. Inter-electrode distance represented 12.5 percent of the muscle-tendon length. Results showed that signal amplitude varied parabolically as the electrode position was brought further away from the elbow. This relationship was also found to be parabolic in nature when the electrodes were positioned transversely to the longitudinal axis of the muscle. This was attributed to the fact that electrode position near the middle of the muscle was at a smaller average distance from all potential sources within the muscle than one located away from the middle. The fusiform geometry of the biceps muscle was also discussed as being another possible intervening factor.

Coincidently, when the amplitude of the averaged EMG versus isometric tension relationship was determined for varying electrode positions, it remained curvilinear. However, at distal positions, EMG-tension curves were slightly lower than that determined from the optimal (middle of belly) position. When surface electrodes were placed on the opposite side of the upper arm, the authors noted that a weak contraction of the biceps could be sensed. This phenomenon, more commonly known as "cross-talk", introduced significant limitations in the differentiation of the respective activities of underlying muscles. Hence, it was recommended that results derived from surface electromyographical techniques should be interpreted with caution.

Komi and Buskirk (1970) established reliability coefficients for surface electrode and intrinsic wire readings of the biceps brachii during different isometric and isotonic contractions. Determination of test-retest reliability for the intrinsic and surface electrode IEMG measurements at intervals of ten minutes on three different days revealed a superior coefficient value for the surface electrode (average r= .88) when compared to that of the indwelling method (average r= 0.62). IEMG measurement reliability with test-retest intervals of 3 and 55 days using surface electrodes was also measured for surface techniques and were considered very good during three types of contractions. In the 55 day interval condition, the following average coefficient values were obtained ; maximum isometric (r=.95), eccentric (r=.91) and concentric (r=.97).

Reliability of surface IEMG recordings was also determined for different changes in muscle length. It was found that during eccentric contraction, reliability was greater at shorter muscle lengths whereas measurements made during concentric contractions were more reliable at optimal muscle lengths.

From these results, it was concluded that the surface electrode technique was appropriate for long term studies of electromyographic muscle activity. However, the authors noted that the chemical method used to mark sites and ruler to provide exact inter electrode distance were critical to obtaining good reliability. Reasons for the less reliable intrinsic measurements were suggested to be due to difficulties in re-inserting electrode into the exact same place, and controlling the amount of exposed electrode surface to the muscle between sessions.

Yang and Winter (1983) determined the intra-class correlation coefficient of the EMG signals within and between days to assess the reliability of electromyographical measurements and reported opposing results. Discrepancies were attributed to differences in methodology, contraction levels and EMG processing techniques. In the closing statement, Yang and Winter (1983) advised that in order to improve EMG reliability: (1) all treatments should be completed within one day, with no electrode manipulation and (2) several trials per condition (a minimum of three) should be used in calculation of an average signal.

In an effort to determine a proper signal averaging technique, Yang and Winter (1984) evaluated the effects of four EMG normalization methods on the inter-subject variability of electromyographic profiles in normal gait. EMG signals from five lower extremity muscles were divided by: (a) the mean EMG over three 50% of isometric maximum voluntary contractions (MVC), (b) the EMG per unit of isometric moment of force, (c) the maximum of the subject ensemble average, and (d) the mean of the subject ensemble average. A coefficient of variation for each method was determined for 11 subjects. Results indicated that methods (c) and (d) produced smallest variability whereas methods (a) and (b) resulted in larger inter-subject variabilities. Lowering of the inter-subject variability by means of normalizing to peak EMG amplitude recorded during the gait cycle was also reported by Knutsson and Richards (1979).

Although some variability was still evident, this could have been attributed to conditions specific to the testing environment: electrode impedance, position of the electrode and skin temperature were listed as examples. Subject-specific characteristics such as muscle fibre-type, amount of subcutaneous fat and manner of the motor unit recruitment, all of which were beyond the control of the investigator, may have also contributed to the inter-subject variability (Komi,1973).

In terms of normalizing to amplitude at 100% MVC, similar results were obtained by Dubo et al. (1976). These investigators normalized the linear envelope (LE) EMGs in gait to the EMG amplitude during a 100% maximum voluntary contraction. From their results, inter-subject variability was very large after this type of normalization.

The results of Yang's and Winter's (1984) study also recommended that normalization of the linear envelope signal to either the peak ensemble or the mean ensemble average as compared to normalizing by an isometric value provided a greater sensitivity for detecting phasic abnormality between different types of contraction actions. Efforts to normalize the EMG signal have not only aided in reducing the inter-subject variability, but have also provided some insight into the factors influencing the nature of the variability.

Although normalizing by a peak or mean ensemble EMG reduces the amount of inter-subject variability, this technique is limited in the sense that the reference is peculiar to the EMG amplitude of the trial. Thus the degree of activation is not known relative to the subject's absolute capability. This may be important in condition comparisons and as stated by Yang and Winter (1984), other measures such as magnitude of the mean EMG (i.e., in millivolts) or results from muscle strength tests should be additionally reported.

In an electromyographical study of cycling, Gregor et al (1985) normalized LE EMGs to the maximum signal of the movement cycle and utilized 50% of this value as a criterion to determine whether or not a muscle was active. This value was assumed to be a conservative indicator of significant muscle recruitment. Even though the results were discussed with reference to this arbitrary signal level, LE EMG amplitudes from maximal isometric or dynamic efforts were never reported.

Applications of the EMG Signal

Although the relationship between EMG and force is not completely understood, EMG techniques have a major role in the estimation of muscle force production. In an overview paper, Crowninshield and Brand (1981) have considered the use of EMG in three general ways.

The most significant application of EMG signals is the direct assessment of muscle force. Although the use of EMG as a determinant of absolute muscle force has been criticized in the past (Solomonow et al.,1986), White and Winter (1986) proposed an EMG-dri en model which used a measure of neural input and calculated muscle-tendon-unit lengths in effort to solve the muscle force estimation problem. They found that a third degree polynomial adequately described the force-length-excitation relationship for the gastrocnemius during gait.

Many limitations exist in implementing this type of model, simply because the relationships between EMG and muscle force during complex movement is still unclear. Yet it does provide a reasonable means of using EMG in an attempt to understand individual muscle function.

In instances when a mathematical approach to solving an intermediate problem is used, EMG signal have also served as a measure of temporal constraint. The level of EMG can be used to reduce the number of unknowns in a human multi-segment model. Such an application has been used in an individual muscle force prediction model presented by Pierrynowsky (1982).

The EMG signal has also been used for temporal validation of muscle force. This application is imperative when considering changes in muscle length during movement. In these instances, it is important to know whether the muscle is active while shortening or lengthening during the motion, thus indicating whether or not an active tension is being created.

Despite its wide range of applications, the use of the Linear Envelope signal as a temporal indicator of force production has been questioned. One of the major criticisms of this use is the time lag introduced by the digital smoothing process. Although this delay is non existent in the raw EMG signal, the time lag is approximately equal to the time between activation of muscle fibres and the onset of tensile force measured at respective origins and insertions. It has been suggested that during this brief time period, contractile force contributes to stretching of the series elastic component, as well as, in overcoming of the initial shortening velocity. Being a function of the characteristics specific to the muscle and the movement created, quantification of the EMG-force time lag for biarticular muscles of the lower limb during deep-knee bending have to this date never been reported.

The EMG-Force Relationship

Lippold (1952) first proposed the use of EMG patterns to directly estimate muscle force during isometric contractions. This was based on the assumption that a quantitative relationship existed between the EMG and muscle force. Although the existence of such a relationship was questionable, this research prompted much exploration into the use of the EMG signal as a means of predicting force.

Whereas all previous EMG-Force related investigations had been performed using animal muscle tissue, Lippold (1952) attempted to quantify this relationship in human gastrocnemius muscle. When the raw EMG integrated over 1/6 sec and measured external force were plotted, a linear relationship (i.e., $0.93 \leq r \leq 0.99$) between these variables was revealed. This observation was also reported by de Vries (1965).

In a study done by Close, Nickel and Todd (1960), a linear relationship between the count rate of action potentials and the integrated EMG was demonstrated. This, however, was not a surprising discovery since the count rate increased with muscle tension in linear fashion.

When the elbow flexor EMG was monitored during isometric contraction at varying loads, Zuniga and Simons (1969) reported a curvilinear (quadratic) relationship between averaged electromyogram levels and muscle tension. Averaged EMG magnitude increased progressively at submaximal intensities whereas during maximal contraction states, little or no increase in tension was observed with marked terminal increase of EMG potential.

Similar curvilinear relationships have been reported by Vredenregt and Rau (1973) and Nightingale (1960).

In their explanation of the curvilinear trend observed, Zuniga and Simons claimed that all previous investigations had denied attempts to fit EMG-force data to a curvilinear line; only linear curve fitting techniques were utilized. Hence, they concluded that the linearity of the IEMG-force relationship reported from these studies resulted from the "straightening out" of the quadratic relationship.

Although these contradictory results seem to suggest that different relationships exist for different muscle groups, Moritani and de Vries (1978) attributed these to: (1) differences in experimental methods, (2) the need for observations over the entire force domain, and (3) lack of control for fatigue. According to de Vries (1968), increases in IEMG per unit of force were apparent during fatigued muscle states. In a study where all of the above factors were controlled, Moritani and de Vries (1978) reported the existence of a linear relationship between IEMG and isometric force. The determined correlation coefficient for these data was $r = 0.990$.

Gottlieb and Agarwal (1971) examined the dynamic relationship between isometric muscle tension and the EMG signal in the soleus and tibialis anterior muscles. They found that the functional relationship between the soleus' EMG and the voluntary isometric plantarflexion torque could be adequately described as a simple linear second-order differential equation.

Hof and Van den Berg (1977) examined the linearity between the weighted sum of the EMG's of the triceps surae and the total plantarflexor torque, during slowly varying isometric contractions. The mean rectified EMG's from the muscles involved (soleus, gastrocnemius: medial and lateral head) were linearly proportional to the torque developed. However, they noted that in more dynamic situations the results could have become more non-linear or even non-reproducible because the synergistic muscles were not always activated in a proportional manner. This study suggested that if a linear relationship was found for one muscle during dynamic activity, it could be assumed that a synergistic muscle would be activated in a similar manner.

During isometric contraction it has been assumed that tension generated at the myofibrilar level is simultaneous to the emergence of the EMG signal. However, under dynamic contraction conditions, the tension has been seen to lag behind the EMG. This delay has been attributed to the fact that the twitch corresponding to each m.a.p. reaches its peak 40 to 100 milliseconds afterwards (Hayes, Norman and Winter,1979). Thus as each motor unit is recruited the resulting summation of twitch forces will also have a similar delay behind the EMG signal. Hence, EMG interpretations to force during ballistic type motion must be considered separate to those made from static isometric conditions. Moreover, since it is known that the instantaneous length and velocity changes of a muscle appreciably influence its force output, it is important to see how well the EMG can predict tension under the more realistic conditions.

Bigland and Lippold (1954) looked at the relationship between force, rate of contraction and integrated (1/6 sec.) raw EMG activity in the gastrocnemius during plantarflexion. A linear tension-EMG relationship was elicited when the level of tension was varied during constant rates of contraction, the slope of this relationship being less during lengthening. Similar results were found at different rates of contractions, however, the slope of each individual curve was rate dependent.

Extrapolating from Hill's (1938) equation for the muscle force versus velocity relationship, Bigland and Lippold (1954) predicted that at lower velocities under constant load, the EMG activity was a linear function of the velocity of shortening. When tension was kept constant and rates of contraction were manipulated, different results were reported for shortening and lengthening actions. During shortening, EMG activity increased with increases in the rate of shortening whereas in lengthening, the electrical activity remained independent of the rate of change of the muscle length. It appeared then that the degree of activation of a muscle required to produce tension while lengthening was less than that required while shortening.

Komi (1973) monitored EMG activity of elbow flexors while subjects did both positive and negative work on an isokinetic muscle testing machine. Each subject was asked to generate maximal tension while the biceps brachii muscle lengthened or shortened at controlled velocities. The basic finding of this study was that the EMG amplitude remained fairly constant in spite of decreased tension during shortening and increased tension during lengthening regardless of velocity. This relationship is illustrated in figure 17.

Results from Komi (1973) and previous studies supported the theory that the EMG amplitude indicates the state of tension of the contractile element only, which is quite different from the tension recorded at the tendon. They also indicated that the EMG activity required to do negative work was less than that required to do the same amount of positive work.

Due to the ease of data collection with surface electrodes, investigators have also tried to quantify the relationship of EMG signal with various types of mechanical variables. According to Bouisset (1973), magnitude of the linear envelope EMG varied linearly with peak angular acceleration. When the moment of inertia of the segment was greater and/or when the motion was arrested by striking a barrier rather than stopping voluntarily, the relationship was denoted by a greater slope. When the kinetic energy of the limb system was determined by squaring the angular acceleration term, the EMG' s relationship with kinetic energy was found to be linear and independent of inertia.

A few very recent studies have attempted to partition the contribution of individual muscles to the total torque output during dynamic muscular contractions as time functions. Essentially, these investigators have tried to use the EMG as one of the inputs in equations designed to follow individual muscle forces in a group of muscles during a certain movement.

Hof and Van den Berg (1981) have researched the possibility of predicting ankle torque based on EMG , angular displacement and angular velocity. This EMG to force processing was done by means of an electrical analogue to Hill's (1938) muscle model. They used a calf ergometer, as well as a modified spring-flywheel apparatus, to examine various isometric and eccentric-concentric contractions of the plantarflexors. This model was found to be an accurate predictor of the force and work of a single muscle or a synergy of muscles. The plantarflexion and dorsiflexion movements were still very slow and well controlled. Hence, application of this model to more dynamic contractions was advised to be done with caution.

The preliminary indications derived from muscle force-EMG related research suggest that the EMG signal can be used to measure individual muscle forces, if the effects on muscle torque output of muscle length, velocity of shortening and lengthening, cross-sectional area, and moment arm length changes with joint angle are all appropriately accounted for. It has also been indicated that the EMG signal reflects only the active contractile component of muscle force and work output. Other factors such as series elasticity are equally or more important force modulators.

In summary, during isometric or slow movement conditions, the majority of investigators have found that the EMG-force relationship can be described as a linear function. As the movement becomes more dynamic, the EMG-force relationship becomes much more difficult to describe. Many factors have to be taken into consideration when trying to relate the muscle EMG activity as we measure it and muscle-tendon tension during dynamic or ballistic movements. Type of recording electrode, inter-electrode spacing, electrode site chosen, as well as, signal averaging technique used in processing are factors which can greatly influence the accuracy with which one predicts muscle tension indirectly via the EMG signal. Although the LE EMG signal combined with muscle length, shortening/lengthening velocity and cross sectional area can be used to quantify force exerted by individual muscles, this information with the exception of muscle cross sectional area was used in this study to detect muscle force on a qualitative level.

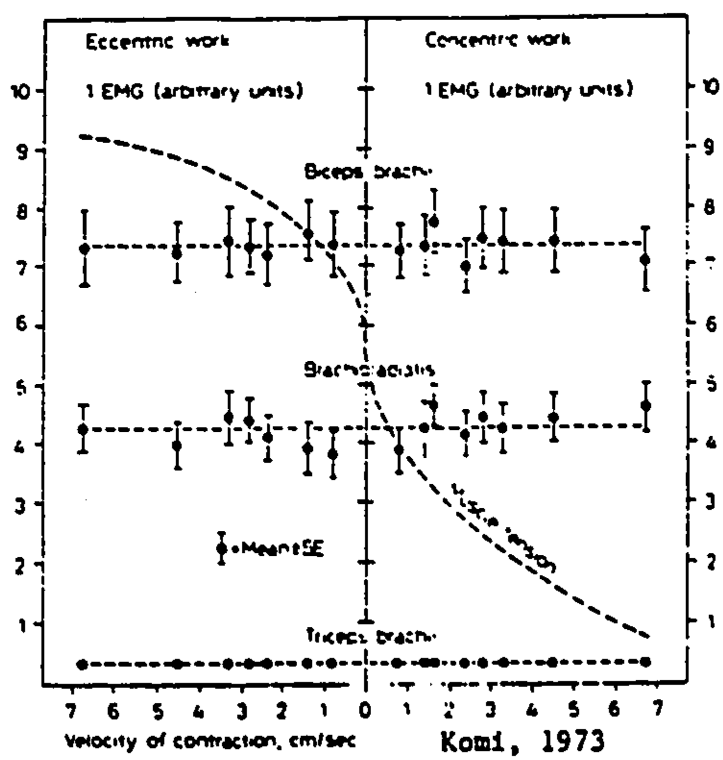


Figure 17. The EMG-contraction velocity relationship according to Komi (1973)

The muscle force vs length relationship

According to present theories, the mechanical structure of mammalian muscle is modelled as being composed of three components. These are the contractile component, the lightly damped series-elastic component and a parallel-elastic component. While the mechanical properties of the latter two components have been determined by microscopic arrangement of structural units, that of the former according to the sliding filament theory, is very much dependent on the interaction of myofilaments in the formation of cross-bridges. According to the studies of Hanson and Huxley (1953), Ramsey and Street (1940), and Gordon et al. (1966), tension developed by the contractile component increased with the number of cross bridges formed.

From experiments performed with frog muscle, Gordon, Huxley, and Julian (1966) revealed that the isometric tension developed by the contractile unit varied parabolically with increasing length of the sarcomere. More specifically, between sarcomere lengths of 1.27 m and 2.0 m tension increased to maximum in two linear stages (inflection point occurring at 1.67 m) and reached a plateau between about 2.0 m and 2.25 m. When the sarcomere increased in length between 2.25 m and 3.65 m , tension decreased to zero in a linear fashion. While a smaller number of cross bridges was responsible for the reduction in sarcomere tension at shorter than normal length, tension in this range of length was proportional to the degree of overlap of the thick and thin myofilaments. Although these measurements were derived from dissected frog musculature, the same has been concluded for mammalian muscle.

Several authors have mathematically described the force-length relationship of muscle under isotonic contraction conditions. Bahler (1968), using experimental data from 28 rat gracilis anticus muscles, fitted a parabolic equation to the force-length data from 0.7 to 1.2 resting length of the muscle fibre. Similar results on single frog fibres were reported by Gordon et al. (1966). In a revision of the methods, Hatze (1975) admitted that this data was accurate for the range indicated, yet, claimed that muscles frequently exceeded these limits during human movements. Hence, the relationship was modelled from 0.58 to 1.8 resting length.

According to Pierrynowsky's (1981) muscle model, the force-length relationship of the contractile element was described using Hatze's (1977a) equation. The normalized muscle force-length relationship of the contractile element used in this model was presented in figure 18. Although it was assumed that this model could be used to represent that of the whole muscle, it was thought to be the best available in the literature.

The muscle-tendon-unit in the proposed study was assumed to be exerting force on the segments of origin and insertion when significant electromyographical activity was evidenced. Due to the nature of the movement, the lengths assumed by the concerned musculature during squatting would most likely not approach the force limiting values listed in figure 18 (i.e., 0 N tension at 0.6 resting length). Assuming that muscle lengths remained within these limits in the deep-knee bending movement, the assumption that all active muscles exert significant force at their associated origins and insertions was considered justified.

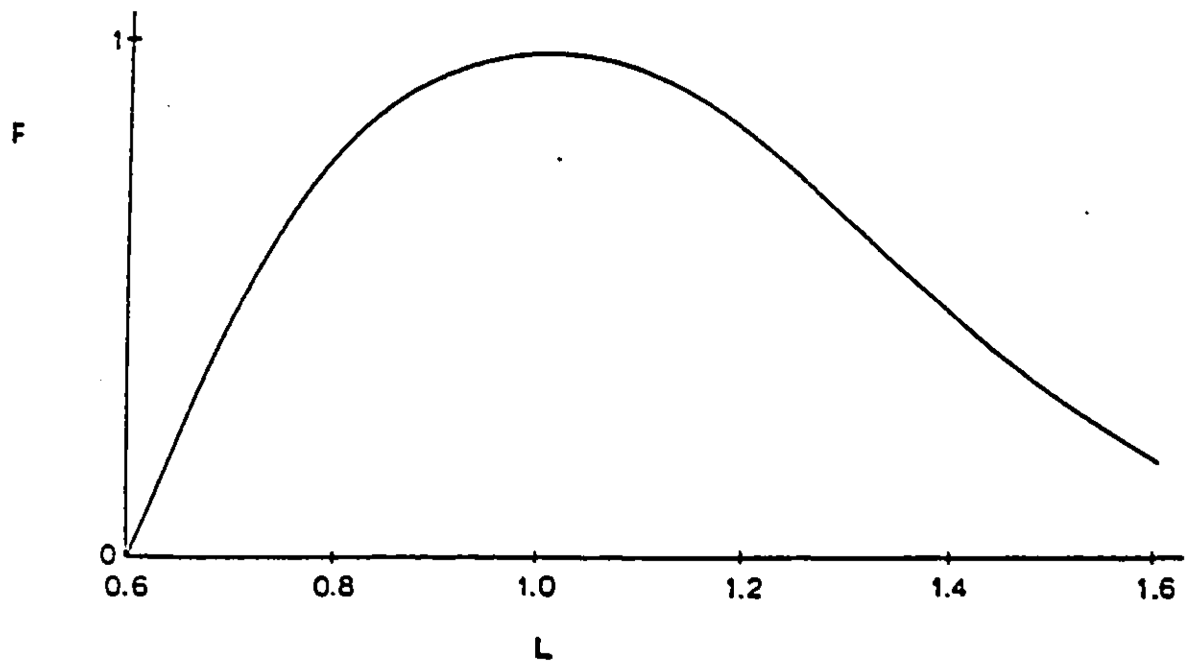


Figure 18. The normalized Force-Length relationship of the contractile element.

The muscle force-velocity relationship

Another factor that heavily influences the amount of tensile force created by a muscle contraction is the velocity at which the contraction occurs. The force-velocity relation for shortening has been determined by numerous investigations from after loaded isotonic tetanic contractions of human muscle (Fenn and Marsh,1935; Hill,1922; Levin and Wyman, 1927; Wilkie,1950). According to Hill (1938) the equation fitted to this inverse relationship could be represented by the following equation:

$$(P + a)\,V = b\,(P_o\text{-}P)$$

where V was the velocity of shortening, P_o was the maximum isometric tension, P was the load, and 'a' and 'b' were physiological constants proportional to the cross-sectional area of the muscle and its fibre length.

As shown in figure 19, this relationship differs between fibre types. Hill's (1938) constant 'a' has the units of force and is higher for fast-twitch than slow-twitch muscles. An 'a' value of 0.20 and 0.35 is used to describe, respectively, the slow twitch and fast twitch force-velocity relationship illustrated in figure 19 (Pierrynowsky, 1982). The constant 'b' has the units of velocity and is approximately six times greater for fast than a slow muscle (Pierrynowsky, 1982). Values of 0.40 and 2.25 resting length are respectively assigned for the slow and fast twitch fibre relationships of figure 19. The model utilized in the proposed study will assume no mechanical difference between the two fast fibre types.

Hill's equation suffered from a number of deficiencies, the major one being that it did not define the relationship during eccentric contraction. Even though a large proportion of muscle activity is eccentric (Abbott et al.,1952), little is known about the maximum force output of a muscle undergoing an eccentric contraction.

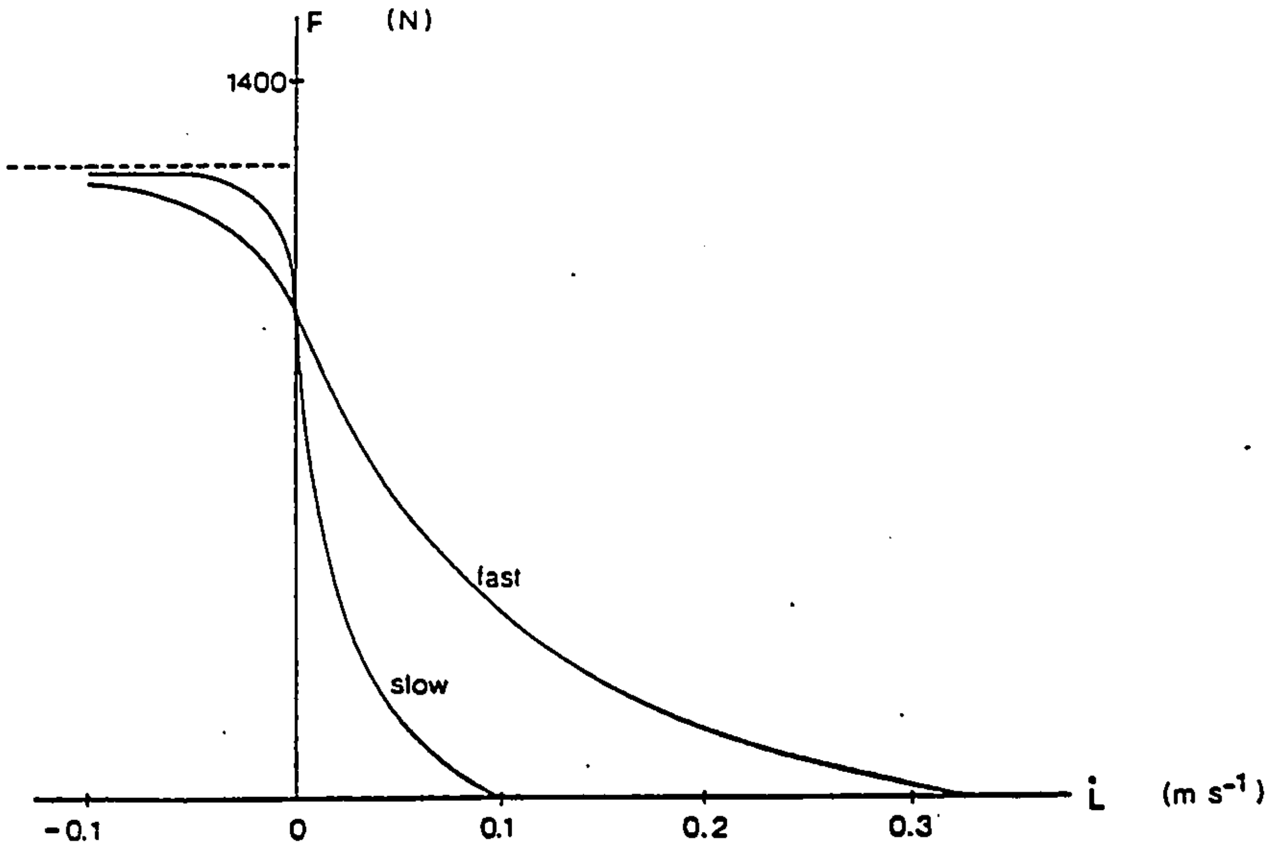


Figure 19. The Force-Velocity relationship of fast and slow twitch fibre populations.

Sugi (1972) studied the force changes during eccentric contractions and has clearly demonstrated that as stretching velocities increased, the force reached a plateau at above isometric tension levels. This is shown in the left section of figure 19. It can be seen that Hill's (1922) equation provides a good description of the relationship between force and velocity only for muscle shortening. Once the muscle is lengthened, the relationship between force and lengthening velocity follows a hyperbolic relationship, albeit, one different than the shortening condition. The maximum force developed by a muscle in eccentric contractions was defined by Joyce et al. (1969) and Katz (1939) to be 1.25 of the maximum isometric force generated.

Although the relationship described in figure 19 represents that of the contractile unit, this can be assumed to be the same for the entire muscle-tendon-unit. Therefore, the concerned muscle was said to be generating significant tensile force if the contractile velocity $V \leq 0.35\ m\cdot s\cdot^{-1}$ (Pierrynowsky, 1982).

Summary

In summary of the chapter, the information reviewed has suggested that paradoxical phenomena explained initially by Lombard (1903) has never been verified in man. Although numerous studies have provided electromyographical and kinetic evidence, none of these have considered simultaneous monitoring of muscle length kinematics. Since one of major assumptions made in Lombard's (1903) model was the isometric contraction of the bi-articular quadriceps and hamstrings during simultaneous hip and knee extension, such a recommendation would verify the existence of paradoxical muscle function as he initially described it. The author chose to examine this phenomenon on subjects during a weighted deep knee bend movement in which there is simultaneous ankle, knee, and hip movements.

A biomechanical review of squatting and other related movements indicated prominent periods of lower limb biarticular muscle co-contractions during ascent phase of the movement. Although no direct evidence was presented, according to data reviewed from similar movements it was suggested that periods of co-contraction increased at elevated movement speeds and external loads. In terms of deep-knee bend movement kinetics, joint moment curves were found to be reproducible between subjects, however, methods used to quantify these were not considered valid due to limitations in the models used.

Pedotti's (1977) method of measuring muscle lengths in two dimensions was chosen for the movement observed in this study. Although it was not believed to be more accurate than three-dimensional or centroidal line modelling techniques, it was considered to be adequate to fulfill the purpose of the proposed investigation.

To justify the use of surface electrodes for the recording of muscle activity, the advantages and disadvantages of using these electrodes compared to indwelling electrodes were described. It was concluded that surface electrode EMGs represented a valid measurement under the conditions presented in this experiment.

The use of the LE EMG signal for detection of muscle force production was discussed and proposed as a valid method for this particular experiment. Fifty percent amplitude of the averaged and time normalized linear envelope signal was proposed as the criterion to detect significant muscle activity. Amplitude averaged and time normalized linear envelopes were chosen as a means of reducing inter-subject variability in the generation of an average representative LE EMG signal.

In the formulation of the proposed methodology, certain assumptions concerning the magnitude of muscle force and its relationships with muscle length, velocity of shortening were discussed. In terms of muscle length and shortening velocity, distinct values obtained from previous works were proposed as being conservative limiting values for significant muscle force production. These values were used in the analysis of the muscle length and EMG data to assess individual muscle force production on a qualitative level.

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