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CHANGES IN MUSCLE ACTIVITY DURING FAST, ALTERNATING FLEXION–EXTENSION MOVEMENTS OF THE KNEE

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ABSTRACT. The effects of high frequency alternating knee flexion–extension on muscle activity of the quadriceps and hamstring muscle groups has been investigated. Standard loads were used for each subject. The muscle activity in vastus medialis, vastus lateralis, rectus femoris and the lateral hamstrings were recorded by electromyography during increasing velocities. Rectus femoris and hamstrings were found to increase their activities significantly with increasing speed while vastus medialis and vastus lateralis showed no such change. The individual thigh muscles thus differ in function in relation to the velocity of movement.

Key words: Knee, muscle contraction, exercise therapy, quadriceps, hamstrings, electromyography

The quadriceps muscles have been studied under many different conditions, covering a variety of postures, speeds and types of muscle contraction and exercise (including isometric, isotonic and isokinetic). In most investigations in which high velocity movements have been studied, the speed of the movement has been controlled by external means often in the form of an isokinetic dynamometer (12, 13, 17). In such circumstances, inertia of the moving segments (due to their mass and acceleration) is almost eliminated. Normally, when a body segment is moved freely through space, and especially when rapid acceleration and deceleration is required, inertia makes an important contribution to the load. For example, during locomotion, the lower limb must experience inertial forces, particularly when velocity is high.

Investigations of high speed movements against inertia have tended to focus on the smaller joints of the upper limb where the inertia of the moving segments is more easily controlled (1, 2, 5, 8). Little attention appears to have been paid to the study of high speed movements of heavy segments involving large muscle masses. In view of this and of the importance of understanding the nature of quadriceps function in realistic situations, a study of high speed knee movement in which knee muscles must combat inertia seemed warranted.

When considering the control of such movements from a neurophysiological viewpoint, rapid ballistic movement appears to form part of a distinct type of voluntary movement which is more likely to be preprogrammed in the higher centres of the central nervous system and operate independently of sensory feedback (4, 11, 16). In the debate of the relevance of various types of motor control on different movements patterns, it has become obvious, when considering the modern neurological theories (15) that no consideration is given to the possibility that some muscle groups may be more involved in phasic activity under the control of higher centres while others may be more concerned with maintaining and adjusting posture. For this reason one could presume that all the individual components of the quadriceps would take part in rapid ballistic movements of the knee.

Some neurophysiologists, however, have recognized the possibility that different types of muscles may respond in different ways to a particular voluntary command. From their experiments on single motor unit recruitment, Desmedt & Godaux (1978) concluded that the organization of motor commands during voluntary ballistic contraction was different for 'fast twitch' and 'slow twitch' muscles. Although not specifically referring to the thigh musculature, this investigation is nevertheless highly significant as motor control is considered in terms of the type of muscle involved.

Far more explicit theories on the motor control of different muscle types had been put forward in the 1960s by Rood (7, 14, 18) based on clinical observations and the neurophysiological theories of the time. These theories stressed that muscles such as rectus femoris and hamstrings were more under control of the higher centres of the nervous system and were mainly involved in brisk, non-weightbearing activities, especially those requiring a high level of skill. In contrast it was postulated that muscles such as the vasti and more particularly vastus me-

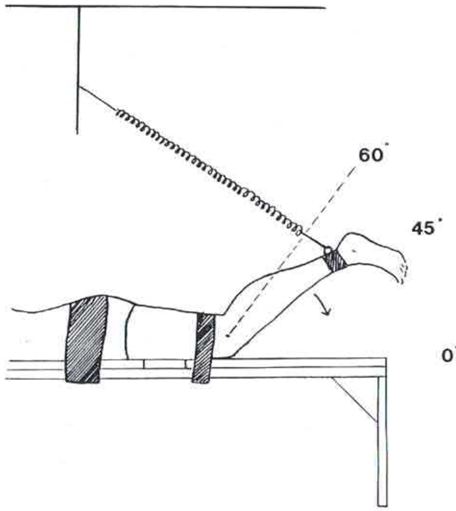

Fig. 1. A diagrammatic representation of the exercise position. Firm stabilizing straps were placed over the lower thigh and pelvis. A space was provided in the foam to decrease pressure on the EMG electrodes monitoring quadriceps activity.

dialis, rely heavily on sensory feedback for their action especially during weight-bearing.

In considering the action of the quadriceps during rapid ballistic movement of the knee the hypothesis that rectus femoris may be specifically facilitated during such an action, especially if performed in a non-weight-bearing position, has been tested.

METHODS

This study involved the EMG monitoring of three components of quadriceps (vastus lateralis (VL), vastus medialis (VM) and rectus femoris (RF)) and of the lateral hamstrings during a specified knee extension and flexion exercise, carried out at three different speeds by normal subjects. For each of the four muscles, their activity during each of the three speeds was compared. It was possible to determine whether the amount of muscle activity changed significantly with increasing speed of movement, and whether the four muscles responded to the speed change in similar or different ways. A biomechanical analysis of the specified exercise was also completed to indicate how muscle torque changed with knee angle and to provide a theoretical estimate of how the power produced during the exercise varied through an exercise cycle.

Twenty female subjects, aged between 20 and 35 years, were included in the study. To be included, subjects had to be in good general health with no history of lower limb pathology or injury. For convenience, all subjects were drawn from the student or staff population of the Department of Physiotherapy at the University of Queensland.

To satisfy the demand for a high speed, movement of the knee against inertia, performed in a non-weightbearing position the movement pattern selected consisted of the following: with the subject in prone lying, knee extension to 0° from a 45° knee flexion position against a light spring resistance, and a return to knee flexion (as in Fig. 1). Prone lying was used as this position has been shown to minimise activity of other hip muscles during knee movements (9). It also enabled the hip position to be kept constant so that the function of the multiarthrodal muscles (RF and hamstrings) could be limited to knee movement. Spring resistance was chosen as pilot studies indicated that during fast repeated contractions, a resistance with some recoil gives a smoother action. Springs also allowed higher speeds to be attained than were possible with either free movement of the leg or against a resistance such as a pulley weight system (as usedby Hellebrant et al. (8)). Furthermore, they have minimal inertial resistance, so that only the inertia of the lower limb (due to its mass) had to be ovecome during the exercise. A spring which would just counterbalance the lower leg at 60° from the horizontal was used. Because of the forces applied during these rapid oscillating movements, the spring was attached directly to a stable bar above the subject. To confine movement to the knee joint, the pelvis and thigh were securely stabilised by straps to the supporting bench (as in Fig. 1).

Several instruments were used to monitor knee angles and muscle action and to set the speed of movement during the exercise. A metronome controlled the exercise rate and, to ensure that the exercise was performed through the specified 45° range of motion, a polarised light goniometer (PLG) was used to monitor the changing knee angles throughout. This provided a continuous illustration of knee joint angles which, after calibration, allowed their measurement. Its signal was displayed in two ways. Firstly, as a form of visual feedback, a digitising oscilloscope screen was placed directly in front of the subject, so that the pattern of knee angles produced from the PLG could be observed during the movements. The subject was shown the limits of the oscilloscope pattern representing 0° (when the lower leg contacted the bench) and 45° knee flexion, so that with practice, she could accurately confine knee movement to the specified range. Secondly, the signals from the PLG were recorded on a chart recorder, for correlation with other parameters during analysis. To enable initial calibration of the PLG knee angle signal, a "Myrin" goniometer was attached to the lower leg.

A four channel EMG was used to monitor the action potentials for the VL, RF, VM and lateral hamstrings, both raw and integrated EMG signals being recorded (using a chart speed of 25 mm per sec).

For each subject, the skin of the left leg was prepared for application of pairs of 8 mm silverchloride electrodes, using techniques described by Gilmore & Myers (6). These were placed longitudinally 2 cm apart to straddle the motor points of VL, RF, VM and lateral hamstrings (located by stimulation with faradic current). After attachment of the PLG and the Myrin-goniometer, the subject was positioned on the exercise bench and the stabilising straps secured over the pelvis and above the knee, and an ankle-strap applied for attachment of the spring (as in Fig. 1). The subject was then asked to sustain maximum isometric contractions of knee extension and flexion while

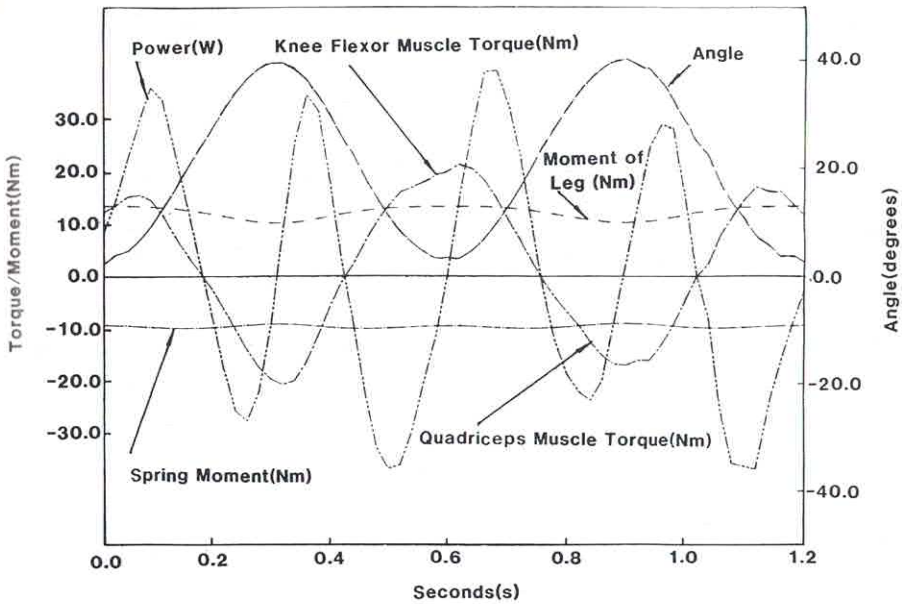

Fig. 2. Computer printout of the biomechanical analysis of the exercise movement performed at Speed 2 (100 beats per minute). All movements were computed about the knee joint, with negative torque representing quadriceps action and positive torque representing knee flexor action. Power was included to illustrate how it varied through the exercise cycle.

the EMG gain was adjusted so that the size of the raw signal was approximately equal on all channels. The integrated EMG output took the form of a full wave area integral with gains normally set at 500 µV per division and resets set at 50 per sec per division.

The subject performed some initial warm-up leg exercises, and then asked to relax the lower leg while the lightest possible spring was attached to ensure counterbalancing of the leg by the spring. For 19 subjects, a '15 lb' spring achieved this purpose, the remaining subject requiring a '20 lb' spring. Calibration of the spring using known weights was carried out to find the spring constant and thus allow estimations of the forces exerted at varying spring lengths. Counterbalancing was considered to be effective when EMG signals of monitored muscles reached minimal levels, signifying relaxation. The position of each subject was then adjusted to meet the standardized requirements of 60° knee angle and 90° between spring and lower leg at the ankle. The 60° knee angle in the resting position was chosen so that when the subject performed the fast repetitive knee extension between 0° and 45° knee flexion, the lower leg would not return to the position of its balance with the spring. Above this point, the movement would have become very irregular as the weight of the leg would no longer have been counterbalanced by the spring. These procedures ensured that despite the varying lengths and weights of subjects' lower legs, the exercise conditions were standardised for all subjects. Using the oscilloscope for visual feedback, opportunity was then given to practise the specified 0° to 45° range of motion at each of the three speeds. Each beat of the metronome represented one complete exercise cycle (i.e. 45° to 0° to 45°). Speed 1 represented a slow frequency of oscillation of 50 beats per min (average angular velocity 75°/sec); Speed 2 an intermediate frequency of 100 beats per minute (average angular velocity 150°/sec) and Speed 3 was the highest possible frequency which could be attained by the subjects after one practice session. This was usually between 120–140 beats per minute (average angular velocity 195°/sec).

Analysis of results relied on a comparison of EMG recordings made under varying exercise conditions. It was therefore important to establish that these were repeatable between successive attempts at any one frequency. To this end, the subject performed the exercise twice for each speed, data being recorded on the EMG once the correct joint range and exercise speed (as shown on the oscilloscope) were peceived by the researcher to have been attained by the subject. Data for approximately ten cycles were recorded, three of these being used for later analysis. To control for such factors as fatigue and learning effect, the three speeds were presented to subjects in random order, with one minute rest interval between trials. The EMG signals were examined for accuracy of joint range and speed during each exercise. Data were not included for analysis if total joint range varied more than 10% or if the time taken to complete three cycles at each speed varied more than 4% from the designated time. In fact, 19 of 20 subjects maintained range and speed within these narrow limits.

Prior to completion of the experiment, measurements were taken of the length of the extended spring in the 60° position, length of the lower leg and distance from the knee axis to the spring attachment. These, together with

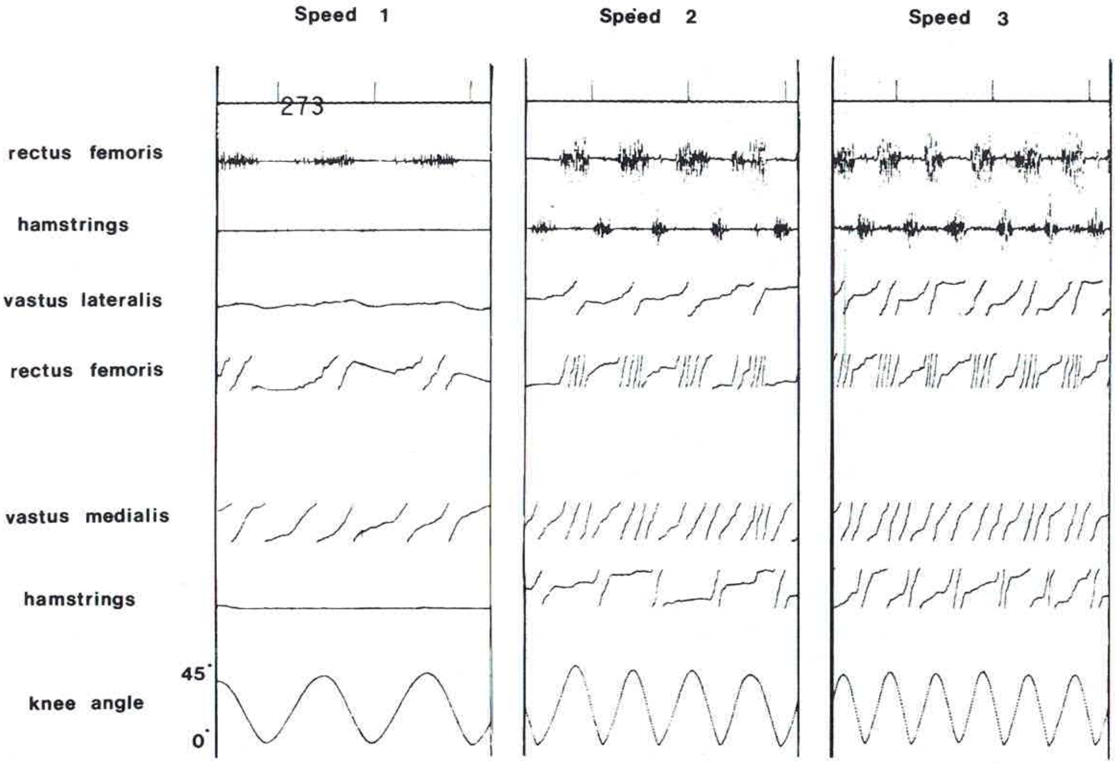

Fig. 3. EMG and PLG recordings at each speed. Recordings from Channels 1–2 represent Raw EMG, while Channels 3–6 represent the integrated signal (each IEMG reset = 5 μVs), with PLG recordings on Channel 7. The time base interval of 1 sec is shown at the top of the readout.

values of the spring 'constant', approximations from tables of the centre of mass of the shank and foot (from Winter (19)) and the data from the P.L.G. on the rate of change of displacement, were used in the development of a biomechanical model of the exercise on a digital computer. The analysis allowed estimations of muscle torques, spring moment, and moment of leg to be plotted for the exercise cycle. Power values were also included to show how these changed through the exercise cycle. A summary of these appears in Fig. 2 with the biomechanical analysis described in detail in Appendix 1.

RESULTS

Although both raw and integrated EMG signals were recorded (as in Fig. 3) the latter (i.e. IEMG) were used as a basis for analysis. For each exercise speed and each muscle, the measures of total activity during three exercise cycles were collated and means and standard deviations for the 19 subjects calculated (as shown in Table I).

Analysis consisted of applying a three way analy-

Table I. *A comparison of muscle activity at varying exercise speeds*

Muscle activity expressed in terms of mean IEMG readings (μVs) over 3 exercise cycles for 19 subjects. n = number of subjects, $\bar{X}$ = mean, SD = standard deviation

	Speed 1		Speed 2		Speed 3	
	$\bar{X}$	SD	$\bar{X}$	SD	$\bar{X}$	SD
Vastus lateralis	26.60	2.35	22.90	2.25	29.75	2.20
Rectus femoris	23.00	2.35	49.15	2.25	69.90	2.20
Vastus medialis	35.50	2.35	34.85	2.25	42.25	2.25
Hamstrings (lateral)	7.15	2.35	18.70	2.30	42.20	2.30

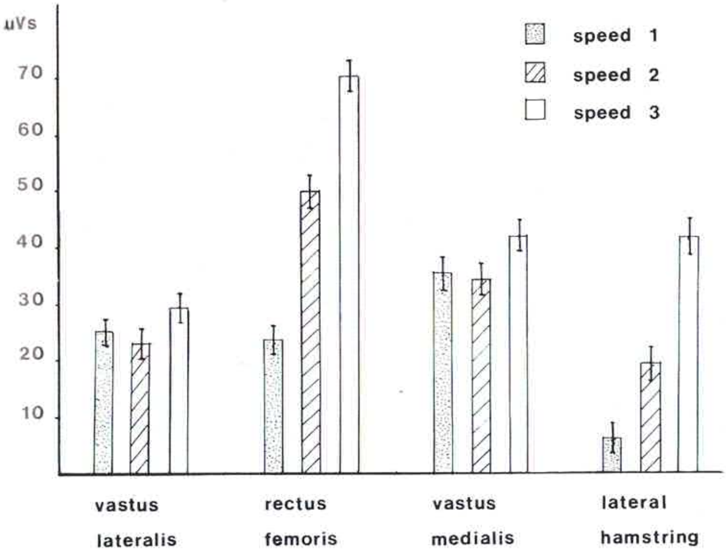

Fig. 4. Changes in muscle activity (measured in microvoltseconds) with increases in speed.

sis of variance (ANOVA) to the factors affecting EMG recordings, namely the four muscles, three speeds and the two trials (i.e. two repetitions of the exercise program at each speed). From Table II it can be seen that there was no significant difference in muscle response for the two repeated trials (M×T interaction), showing that the muscle function for each specified condition was repeatable. This analysis also revealed that certain muscles reacted differently to changes in speed, as shown by the level of significance of the M×S interaction in Table II. As the ANOVA referred to responses of all muscles collectively, further analyses, in the form of *t*-tests were undertaken to determine more explicitly which of the four muscles responded by increasing activity with increasing speed. For each muscle, the means and standard deviations of EMG recordings for speeds 1 and 3 were compared, as shown in Table III. It can be seen that muscle function varied between speeds but that it did so in different ways for each muscle. Fig. 4 illustrates this graphically.

In view of these results, multiple comparisons were undertaken using the Sheffe method (10). A comparison of changes in muscle activity between RF and hamstrings and between VM and VL revealed no significant differences in either case. (RF and hamstrings: F_s=4.996; critical value of F=8.04. VM and VL:F_s=1.097; critical value of F=8.04.) Further comparisons showed that the average change in muscle activity of RF and hamstrings was significantly greater than the average change in activity of VM and VL (F_s=213 with critical value of F=18.9 p<0.01). In terms of magnitude of muscle activity, this experiment showed that high speed oscillating knee extension movements demand a very pronounced increase in activity of RF and hamstrings over that required for

Table II. *Results of 3 way analysis of variance*
df = degrees of freedom, SS = sum of squares, MS = mean square, F = F ratio, p = significance

Source	df	SS	MS	F	p
Subjects	19	261 727.75	13 775.25	38.39	<.005
(M) Muscle	3	123 360.00	41 120.00	116.08	<.001
(T) Trials	1	72.25	72.25	.20	NS
(S) Speed	2	117 863.00	58 931.50	166.36	<.001
M × T	3	58.00	19.25	.05	NS
M × S	6	77 526.00	12 921.00	36.47	<.001
M × T × S	6	438.00	73.00	0.21	NS

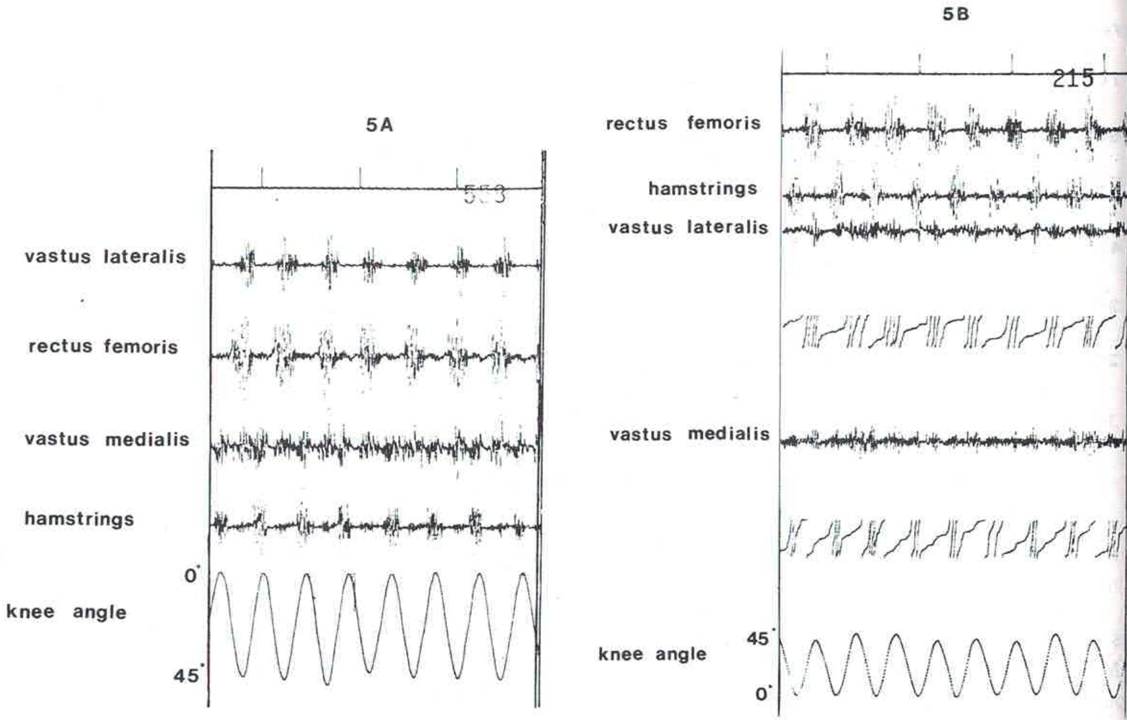

Fig. 5. The sequence and pattern of muscle activity at Speed 3. 5 A and 5 B represent two different subjects.

slower speed movements, but that no such changes occur in VM and VL.

The timing and sequence of muscle activity throughout the range was examined also. Fig. 5 represents examples of raw EMG for RF, VM, VL and hamstrings for two subjects at the fastest speed. Similar consistent patterns of muscle activity were found for all subjects. In 100% of cases, RF and hamstrings displayed phasic 'on and off' activity (as in Figs. 5 A and 5 B). In contrast, VM showed a tonic, continuous pattern in 90% of subjects, while VL displayed this pattern in 45% and worked phasically (as in Fig. 5 A) in 55% of subjects.

DISCUSSION AND CONCLUSIONS

During the execution of this rapid oscillatory movement in prone, the pattern of rectus femoris and hamstrings was different from those of vastus medialis. The rectus femoris and hamstrings operated reciprocally at different parts of the exercise cycle to accelerate and decelerate the lower leg as expected of the primary flexors and extensors of the knee joint. Vastus medialis and lateralis generally acted tonically in a manner which would stabilize the knee joint and restrain the patella during the movement.

The most important finding was that, as the speed of the movement increased, highly significant increases in the activity of rectus femoris and hamstrings occurred in comparison to the vasti. These results are limited to the prone position and thus care needs to be taken with generalizations. The hypothesis that rectus femoris is facilitated more

Table III. *Comparison of muscle function between Speeds 1 and 3*

Muscle	Change in muscle function	*t*-value	*p*
Vastus lateralis	No change	0.667	
Rectus femoris	Significant increase	14.430	<.001
Vastus medialis	Increase just signif.	2.070	<.050
Hamstrings (lateral)	Signif. increase	10.470	<.001

than the vasti in high speed, highly skilled, non weight bearing activities was tested. Significant differences were observed in the predicted direction so the null hypothesis was rejected.

There are some important implications of this study in relation to the design of exercise in rehabilitation. High velocity, low load activity, particularly in the prone position, appears to be inappropriate exercise when quadriceps power needs to be improved. This is because muscle imbalances would occur due to the specific facilitation of rectus femoris and hamstrings with the apparent inhibition of the vasti. The technique may prove useful however if rectus femoris and hamstrings require maximal facilitation.

This study of the effect of increasing speed on muscle function has revealed that individual muscles of the knee musculature respond differently when subjected to high speed alternating exercise movements in the prone position. This finding may contribute to basic scientific knowledge on muscle function, and has clinical significance in relation to the design of quadriceps exercise programs.

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APPENDIX I. *BIOMECHANICAL ANALYSIS*

The angular displacement θ of the leg is approximately sinusoidal with a mean displacement and an amplitude of 22.5°=0.393 radians.

The circular frequency $\omega = \frac{2\pi B}{60} = 0.105B$ (1)

here B = no. of beats per minute

$\therefore\ \theta = 0.393 \cos [0.105\ Bt] + 0.393.$

From Fig. 1 it is obvious that the torques produced by the quadriceps M_Q, by the knee flexors M_H, by the spring force F_S and by the mass M of the leg all combine to produce a resultant torque about the knee which produces the rotation, i.e. the inertial acceleration. $I\frac{d^2\theta}{dt^2}$, where I is the inertia of the leg about the knee joint.

Thus $\Sigma\ M_m = [M_H - M_Q] + F_s A \sin\delta - M_g C \cos\theta$ (2)

where

A =distance from knee to attachment point of spring

δ =angle of inclination between leg and spring

C =distance of centre of mass of lower leg (shank and foot) from knee
=0.606 L (Winter, 1979)

L =length of lower leg.

Now $\Sigma\ M_m = I\frac{d^2\theta}{dt^2}$ (3)

where

I = Inertia of lower leg = Mr^2

M = mass of the leg = $2F_sA/C$, calculated at θ = 60° the equilibrium point

r = radius of gyration = 0.735 L [Winter 1979]

$\therefore I = 1.08\frac{F_sA}{C}L^2$

The power P is the product of the torques and the angular velocity

$$P = I\frac{d^2\theta}{dt^2}\frac{d\theta}{dt} \quad (4)$$

It should be noted that the frequency of the power is twice that of the displacement θ. In addition, whilst the torque M_Q–M_H oscillates at the fundamental frequency, the form is nonsinusoidal due to the spring forces and the mass moments. The computer analysis product the variables plotted in Fig. 2. Any inaccuracies due to the assumption of sinusoidal variation in θ are negligible compared to the difficulties in estimating the mass and inertia of the lower leg.

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