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A BIOMECHANICAL COMPARISON OF FORWARD AND REARWARD HUMAN PROPULSION

presented by

Timothy William Flynn

has been accepted towards fulfillment of the requirements for

Master of Science degree in Biomechanics

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A BIOMECHANICAL COMPARISON OF FORWARD AND REARWARD HUMAN PROPULSION

by

Timothy William Flynn

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ABSTRACT

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The purpose of this thesis was to investigate the stance phase of rearward walking and running and compare these to their forward ambulation counterparts. The saggital plane of the right knee was analyzed. The knee muscle moment, power, and work requirements were determined. EMG signals were captured from six muscles. The ground reaction torque was analyzed. Statistically greater peak negative power and negative work occurred during forward walking and forward running. Significantly different patterns of EMG activity were found between forward and rearward conditions. The ground reaction torque suggested a decrease in the tibial internal rotation and subtalar pronation during rearward walking and rearward running respectively. The use of retropropulsion in the rehabilitation of overuse injuries of the knee is discussed.

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To Susan, David, and Phillip, for the happiness which you bring me and the patience you show for a wandering mind.

And finally, to my father, John Patrick Flynn for continuing to journey with me daily, "Dad, your presence is felt."

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I. INTRODUCTION

During the last three decades there has been a steadily increasing emphasis on maintaining a healthy lifestyle. In an effort to increase the quality of one's life and decrease the risk of disease, the public has become more interested in aerobic fitness. Running is one of the most efficient means of increasing cardiovascular endurance. In this country alone, millions of people are running everyday. Unfortunately, a large percentage of runners and recreational joggers develop debilitating knee pain during their training. James and associates(24) found the knee joint to be the most common cause of pain in runners and the patellofemoral joint to be the most common area of dysfunction. One factor contributing to this is the large forces that enter the body during running. A large amount of muscle energy is expended in an attempt to deccelerate the body. The knee joint demonstrates this via the large eccentric activity of the quadriceps group during the initial stance phase of running. The early stance to mid stance phase of running is considered the time where biomechanical loads give rise to knee dysfunction.

Despite the common occurence of overuse injuries around the knee, there is no common successful prevention method or rehabilitation protocol. One method that is gaining popularity in the rehabilitation community in the treatment of these injuries is backward or retro walking and running. Despite the fact that retropropulsion is becoming a common modality in the rehabilitative process, little research exists that analyzes the unique biomechanics of this activity.

The purpose of this research was to investigate the stance phase of retro walking and running and to compare this to their forward ambulation counterparts. The knee will be the primary joint of interest. The model of investigation will center on the determination of the knee muscle power and work. Kinematic and kinetic data will be captured. Electromyography of six muscles surrounding the knee will give insight to the primary muscle firing pattern during these activities. The functional and clinical significance of retropropulsion will be addressed throughout this thesis.

II. SURVEY OF LITERATURE

The survey of literature will be divided into four sections: retropropulsion, muscle power, electromyography, and clinical & functional biomechanics.

Retropropulsion

Backward walking and running as a form of movement has probably been utilized since man began upright activity. Various daily activities require us to move backwards to position our upper extremities. Numerous sporting activities require backward running. Just as the cromagnum man back-pedaled in self-defense of a wild animal, the modern day athlete runs backward to position himself on the defensive field. Though normal children develop the ability to move backward at an early age, it appears that it requires a high central neural program to perform. Thorstensson's(43) study of five subjects performing backward and forward walking noted that the lower limb followed essentially the same trajectory when performing forward and backward walking. However, these trajectories are in opposite directions, in order to accomplish this dramatic and very specific change most muscles changed their pattern of activity in relation to the different movement phases. Providing that the same neural circuitry is utilized in forward and backward walking, Thorstensson suggested a marked and specific modifiability occurred in the neural network that generates locomotion.

In 1980, a short article by Flodberg(19) reported the rehabilitation value of retro running in the treatment of an overuse injury of the hip. Mackie et al.(29) investigated the effects of a three month backward training program in twenty one subjects following ACL injury. They reported a significant increase in hamstring and quadriceps power, as measured on a CYBEX II testing apparatus. Bobath(8) and Brunnstrom(10) recommend backward movement in the evaluation and treatment of motor control in hemiplegic patients.

Several authors have studied the kinematic differences in forward and backward ambulation. Bates et al.(7) studied nine female runners in forward and backward conditions. The authors assumed running backwards to be kinematically opposite of forward running. Bates et al. reported backward running required significantly less hip joint range of motion and greater knee joint range of motion. The authors' theorized that muscle function would reverse from concentric to eccentric action and vice versa. Bates and McCaw(6) tested two subjects walking forward and backward on a treadmill. The greatest differences were noted at the knee joint. In forward walking, the knee joint exhibited the well documented three periods of load accepting flexion, followed by the extension period of mid stance, and a second flexion period prior to toe off. Backward walking consisted only of a single extensor phase throughout the entire support period.

Shuck(41) agreed with the finding of a single extensor period in backward walking. The author noted that the ankle dorsiflexes sharply after initial toe strike in backward walking, apparently compensating for the lack of the knee flexion phase normally seen in forward walking. The knee appears to be the primary shock absorber in forward walking and running and the ankle in backward walking and running.

Vilensky et al.(47) studied sixteen parameters in four subjects walking forward and backward on a motor driven treadmill. In contrast to Grillner's prediction(20), that human backward walking is achieved by a simple change in phase relationship between the hip and knee joints, Vilensky et al. noted marked changes in the movements of the hip and knee joints and their interactions. The authors were in agreement with Bates and McCaw's(6) finding of a single extensor support phase at the knee during backward walking. Vilensky et al. also noted that backward walking was achieved by a faster cadence, but a decreased stride length when compared to forward walking at the same speed. A decrease in stride length tends to be a protective strategy for gait disturbances(12) such as paresis, pain, and coordination disorders(27). Given the fact that this protective strategy occurs in healthy individuals when stability is challenged, Vilensky et al.(47) proposed that backward walking "threatened" stability.

Ground reaction forces have been measured in backward running. Armstrong et al.(2) briefly described vertical, fore-aft, and medial-lateral force differences during forward and backward running in eight subjects. The lack of a defined first peak in backward running vertical force was noted. Armstrong et al. interpreted this as a more gradual dissipation of force controlled by the eccentric calf muscle contraction. The peak lateral force was greater and lasted for a larger amount of the support time in backward running, when compared to forward running. The investigators' interpreted this as necessary to maintain stablility while running backwards.

Threlkeld et al.(45), in a study of ten runners, concluded that backward running produced lower vertical impulse stress than forward running. The investigators' also noted that an eight week training program of backward running improved concentric knee extensor torque at low speeds on a isokinetic dynamometer. Threlkeld et al. reported backward running could be clinically useful for reducing stress to injured joints and for increasing knee extensor strength.

Kramer and Reid(25) studied one subject walking forward and backward using high speed cinematography and surface electromyography. The authors noted that the lower extremity muscles were electrically active for greater sustained periods of time (consistent activity), and also demonstrated a greater degree of inconsistent electrical activity during backwards walking than during forward walking. Muscles were more active in a pulling and stabilizing function during backward walking than during forward walking.

Muscle power and work

In order to gain insight into the rehabilitative and training siginificance of backward ambulation, a biomechanical evaluation must assess muscular activity. Winter(54) states that only by examining the mechanical powers at each joint can an assesment of the importance of the muscles at the ankle, knee, and hip be ascertained.

In Elftman's classic studies(14,15), he outlined methods for calculating the rate of work done on the leg and further demonstrated this by analyzing one stride of one runner(16). Quanbury and colleagues(37) studied the power flow to the lower extremities during the swing phase of walking.

Robertson and Winter(39) studied two subjects during the complete walking cycle. The authors studied the rate of work done by the muscle moments and termed this muscle power. Where muscle power is the resultant joint muscle moment vector multiplied as a dot (scalar) product with the joint angular velocity. When the resultant joint moment is in the same direction as the joint angular velocity concentric muscle action occurs, if the directions are opposite an eccentric muscle action is occurring. With this model the lower extremity muscles could be viewed as generating mechanical energy or absorbing mechanical energy via concentric or eccentric muscle contraction respectively. Robertson and Winter(39) concluded that the measurements of joint and muscle power were valid throughout the walking cycle for all trials of the three leg segments studied, except the ankle during weight acceptance and late push off. The authors also noted that the assumption of the joints acting as an ideal hinge connection was valid for joint moment and muscle power analysis.

Winter(54) studied the moments of force and mechanical power in eleven subjects performing slow jogging trials. He described five distinct phases of the knee muscle power pattern: an initial shock absorbing peak during weight acceptance, a small generation burst during early push off, a major absorption pattern during late push off, a third absorption peak decelerating the leg and foot prior to impact, and a final small positive burst as the knee flexors rotate the leg posteriorly to reduce forward velocity prior to heel contact. Winter also found that over the entire stride the knee muscles absorbed 3.6 times as much energy as they generated, and the ankle muscles generated 2.9 times as much as they absorbed.

Ae and associates(1) studied five skilled sprinters at increasing running speeds. The authors found that the muscle power patterns were consistent for increasing speed, but that the magnitude of muscle power increased as running speed increased.

Electromyography

Elliot and Blanksby(17) stated by synchronizing electromyography (EMG) with cinematography one can gain a more complete understanding of the integrative pattern of electrophysiological and mechanical parameters in the performance of human locomotor skills.

Numerous investigators have studied EMG activity of the lower extremity musculature during walking. In 1963 Moore and colleagues(32) studied muscle activity in walking with a system which telemetered the EMG signal. This allowed the subject to be freed from a cumbersome umbilical cord which followed the subject through his or her activity. Since that time multichannel telemetry systems have advanced considerably and now allow for maximum freedom of movement and essentially no added weight for the subject.

Yang and Winter(56), studying eleven subjects walking at three different cadences, noted a significant change in the magnitude of the signal at increasing velocity but the shape of the EMG pattern generally remained similiar at the different cadences. Yang and Winter noted that although the EMG pattern changes across subjects revealed a seemingly systematic trend, the individual subject responses varied greatly. The investigators thought that these individual differences could be related to the trade off in function between synergestic muscles, the differences in fiber type, or the kinetic differences in each of the subjects walking gait.

Arsenault et al.(3) attempted to validate the notion of a normal profile of EMG signals during gait. Within a subject all the data obtained from a given muscle were observed to be extremely stable. This indicates that gait might be programmed, if programming is defined as high repeatibilty in neuromuscular output. However, analysis across subjects, demostrated differences between the muscular recruitment profile of several of the muscles investigated. The authors felt these peculiarities for a particular muscle for individual subjects were important, since biologically speaking such peculiarities in the EMG firing pattern would contribute to the production of an overall joint moment history differing from one subject to another.

Arsenault et al.(4) accumulated EMG data over 10 strides in eight subjects. The author concluded that since intra-subject variations were usually small, three strides per subject would offer very reliable EMG data for that subject. Furthermore, the investigators reported three strides per subject to be reliable for inter-subject comparisons.

The EMG pattern has also been studied in running. Elliot and Blanksby(17), studying ten females running on a treadmill, noted that from foot contact to mid stance the lower extremity muscle activity was concerned primarily with stabilization. Lower limb stabilization then gave way to a powerful driving thrust during the mid and late support phases. Similar results were reported by Elliot and Blanskby(18), when analyzing ten male runners.

At foot strike during running, marked activity has been shown in the vastus lateralis and vastus medialis in preparation for the rapid loading which subsequently occurs.(52) This correlates with Komi(26) who reported that muscle activity during the eccentric phase of contact was much greater in magnitude in all muscles than during the concentric phase.

Functional and clinical biomechanics

Mechanics of the foot/ankle complex significantly influences patellofemoral joint mechanics. A brief description of the closed kinetic chain motion of the subtalar joint, and its influence on the knee is now presented.

Root and colleagues(40) describe pronation of the subtalar joint during weight bearing consists of calcaneal eversion with the talus adducting and plantar flexing. Supination is described as calcaneal inversion with abduction and dorsiflexion of the talus. Inman(22) states that the primary function of the subtalar joint is to absorb transverse plane rotation of the lower extremity. Levens and colleagues(28) classic study on transverse rotation of the lower extremity in walking found tibial transverse rotations averaging 19 degrees. The researchers also reported relative transverse rotation of the tibia with respect to the femur to average 9 degrees.

During the initial stance phase of gait the weightbearing limb is internally rotating. To allow the foot to stay in the line of progression, the subtalar joint absorbs the internal rotation of the lower extremity by pronating.(22,40) James(23) reports pronation reached a maximum at 15% of the stance phase of walking and at 40% of the stance phase during running. After maximum pronation is reached the subtalar joint gradually supinates. James and colleagues(24) state that excessive or prolonged pronation during the support phase creates increased forces not only applied to the supporting structures of the foot but also to the knee. Furthermore, when tibial internal rotation is increased and prolonged with excessive pronation, more transverse rotation must be absorbed in the knee joint with subsequent disturbance of the normal tibio-femoral rotational relationship and alteration in patellofemoral mechanics.

Tiberio(46) states that five extra degrees of pronation occurring during midstance holds more potential for producing pain than five extra degrees occurring during the initial contact phase. He further states that since the subtalar joint should begin supinating during midstance, the extra pronation at this time is actually a much greater functional deviation and will require greater compensation on the part of the femur.

A number of authors have reported a reduction in patellofemoral symptoms by controlling the amount of pronation during the stance phase of running(11,12,34). Foot orthotics and specific shoe design are methods used to decrease pronation.

A method to indirectly measure the tibial rotation and subtalar joint pronation during stance is the ground reaction torque. The ground reaction torque about the vertical axis has been studied in walking and running. Mann(30) and Root(40) have postulated that this torque is a direct response to tibial transverse plane rotation. This appears to be true during walking, as Ramakrishnam and colleagues(38) found the ground reaction torque to be internally directed during the first 40% of stance phase, and then switched to externally directed during the remainder of stance phase. This would coincide with the tibial internal and external rotation motion that is occurring.

Holden and Cavanagh(21) studied the ground reaction torque during running in ten male runners. The authors concluded that the rationale used to explain ground reaction torques during walking could not be applied to running. Holden and Cavanagh postulated that ground reaction torque in normal running acts to resist foot abduction during the first 60-70% of support, when pronation is known to occur since foot abduction is a component of pronation. During the remainder of support phase the ground reaction torque resisted adduction. The authors also noted that when the subjects wore footwear that increased pronation a subsequent increase in the ground reaction torque tending to resist foot abduction occurred. The above explanation of the ground reaction torque allows an investigator to use the ground reaction torque as an indirect method of measurement of tibial rotation during walking and subtalar joint pronation during running.

III. EXPERIMENTAL METHODS

A general description of the experimental methods and techniques used to collect and reduce the data are described in this chapter. All collection was performed at the Biomechanics Evaluation Laboratory, Saint Lawrence Hospital, Lansing, Michigan. Three types of information were experimentally recorded: kinematic activity of the lower limb; kinetic ground reactions; and surface electromyographic signals from six lower extremity muscles.

Equipment

Video data were collected using four solid state, shuttered video cameras. Data collection rate was 60 frames per second at one millisecond per frame. All four cameras were synchronized by a VP-320 model dynamic processor. The cameras were positioned to optimize viewing of the targets placed on the right lower extremity of the subjects.

A calibration space consisting of twelve targets of known position was placed in the field of view of the cameras. The calibration space (see Figure 1) measured 182.88 X 121.92 X 114.3 cm. The calibration structure provided a known coordinate sytem to define the space of the viewing area. The center of the force plate was the origin of the coordinate system. Each target in the calibration space was covered by retro-reflective tape (3M Scotchlite

Figure 1

Calibration space

Corporation). Illumination of the targets was provided by a single flood light attatched approximately two inches from the center of each camera lens. The proximity of the flood light to the lens was required to achieve the maximum intensity of the reflected light. The retro-reflective tape is extremely sensitive to the "observation angle" which is defined as the angle between the incidence light ray, the reflective target, and the reflective ray returning to the camera lens. An increase of one degree in observation angle causes a 16 fold reduction in the intensity of the reflected light.(44)

The calibration space was filmed and each target location was digitized using the Expertvision(33) three dismensional (EV3D) digitizing program. The EV3D digitizing Program calculates the centroid location of each target. Using the method of direct linear transformation developed by Walton(48), the transformation matrices were determined and stored in the environmental operator section of EV3D. The accuracy of the calibrated space was reported as a "norm of residuals" for each camera. The residual values were less than .38 for all testing sessions. This fell below the system requirements for residual values of less than 2.0, indicating an accurate calibration space.

Ground reaction forces (Fx, Fy, Fz) and the ground reaction torque (Mz) were measured using an AMTI Biomechanics Force Platform Model OR6-6. The force platform incorporates strain gages which measure the applied forces and moments, amplify the signal and then send it to the analog to digital converter. The signal was sampled at a rate of 1000 Hz and stored on the Sun 4 work station (see Figure 2). The orientation of the force plate and laboratory coordinate systems are shown in Figure 3.

The electromyographic signals were collected via surface electrodes and telemetered to a Transkinetics receiver. The signals for each muscle were stored on the Sun 4 workstation.

Subjects

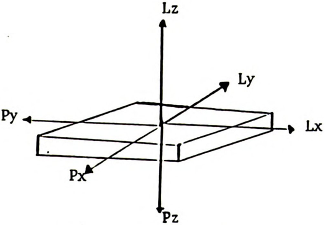
The subjects of this study were male graduate students at Michigan State University. None of the subjects were presently performing long duration backward locomotion on a regular basis. All but one subject was presently engaged in sports (basketball, football, karate) which required backward movement. The subjects were void of previous knee trauma or pathology. Table 1 describes the individual subjects traits. The subjects wore their normal jogging shoes during the testing.

Table 1
Individual subject traits

Subject #	Height (cm)	Weight (kg)	Age (yrs)
F1	192	82.4	24
F2	183	89.8	28
F3	180	69.0	28
F4	172	64.5	33
F5	185	84.5	28
F6	170	63.1	31
Mean $\pm$ sd	180.3 $\pm$ 8	75.6 $\pm$ 11	28.7 $\pm$ 3

Figure 2

Sun 4 workstation



(Px, Py, Pz) = Force plate coordinate system
(Lx, Ly, Lz) = Laboratory motion coordinate system

Figure 3

Force plate and laboratory coordinate systems

Prior to testing the subjects signed an informed consent and were briefed on the testing sequence. Motor points of six muscles on the right lower extremity (rectus femoris, vastus lateralis, vastus medialis, biceps femoris, gastrocnemius, and tibialis anterior) were located. Exact electrode placement was determined using a Chatanooga (Chatanooga Corp., 101 Memorial DR., Chatanooga, TN 37405) Intelect model 500 neuromuscular stimulator (see Figure 4). The points were identified and the area was prepared by shaving the region with an electric razor and then wiping it several times with a dry cloth to remove skin oils. In order to minimize cross talk between muscle groups the electrodes were placed approximately 1-2 cm apart over the motor point of interest.(5) The electrodes were attatched to the transmitters and secured on the subjects right lower extremity with self adhesive tape. Once all electrodes and transmitters were in place a manual muscle test was performed on each of the six muscles while monitoring for the appropriate activity. If the signals were weak or absent, modification was made.

Seven 2.54 cm diameter and two 1.27 cm diameter retro-reflective spherical targets were placed on specific locations on the subjects pelvis, right lower extremity, and right shoe. Target size was chosen to maximize the efficiency and accuracy of the automated digitizing system. The EV3D digitizing system sweeps across a 240 X 256 pixel grid on the video image and computes the average of the

Figure 4

Neuromuscular stimulation of motor points centroid of each target. The larger targets provide a more accurate centroid due to greater pixel surface area in which to average the centroid. Conversely if the targets were too large, merging of two targets would occur and result in the centroid of the merged spheres to be calculated as a single target. Table 2 gives the target number and anatomical location of each target. Target positions are shown in Figure 5.

Table 2
Target number and anatomical location

Target #	Landmark (on right side of body)
1	Anterior superior iliac spine
2	Posterior superior iliac spine
3	Greater trochanter
4	Lateral femoral condyle
5	Anterior tibia (at level of proximal gastrocnemius tendon)
6	Posterior shank (at level of proximal gastrocnemius tendon)
7	Superior calcaneus (superior heel counter of shoe)
8	Inferior calcaneus (inferior heel counter of shoe)
9	Distal 1st metatarsal (on shoe)

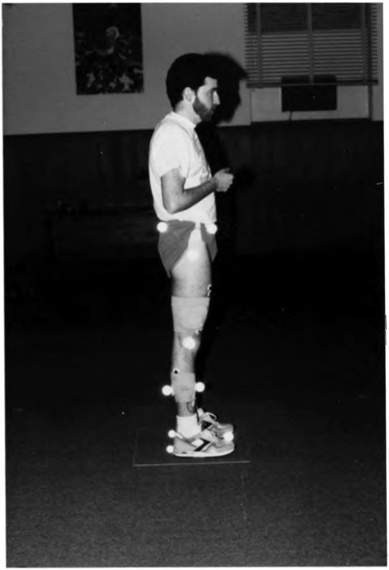


Figure 5

Target locations on subject

Data collection

The subject stood on the force plate in a relaxed position with the knee joint in neutral (neither flexed nor extended). Five seconds of video data were collected and stored. This file was used to calculate the offset knee angle. This step allowed the linkage targets to be independent of an exact vertical or horizontal position.

Three trials of four different conditions were randomly tested. The conditions were: walking forward, walking backward, running forward, running backward. A trial consisted of a subject's right foot landing entirely on the force platform. A mistrial occurred if the subjects stride was unnatural or if the subject altered his stride in an attempt to hit the force plate. The trials were collected using the Beldata software program(49) which allowed simultaneous collection of force plate, kinematic, and EMG data. When the force plate was triggered at the instant of vertical loading, an event marker was placed on each raw video and EMG file to allow synchronization of all components. Immediately after the trial the ground reaction forces and EMG results could be viewed. Three successive trials could be overlayed and viewed. Figure 6 demonstrates the reproducibility of the force data during three successive trials in subject F2 walking forward.

Following successful completion of all trials, the video files were transfered to the EV3D program for digitizing and further analysis.

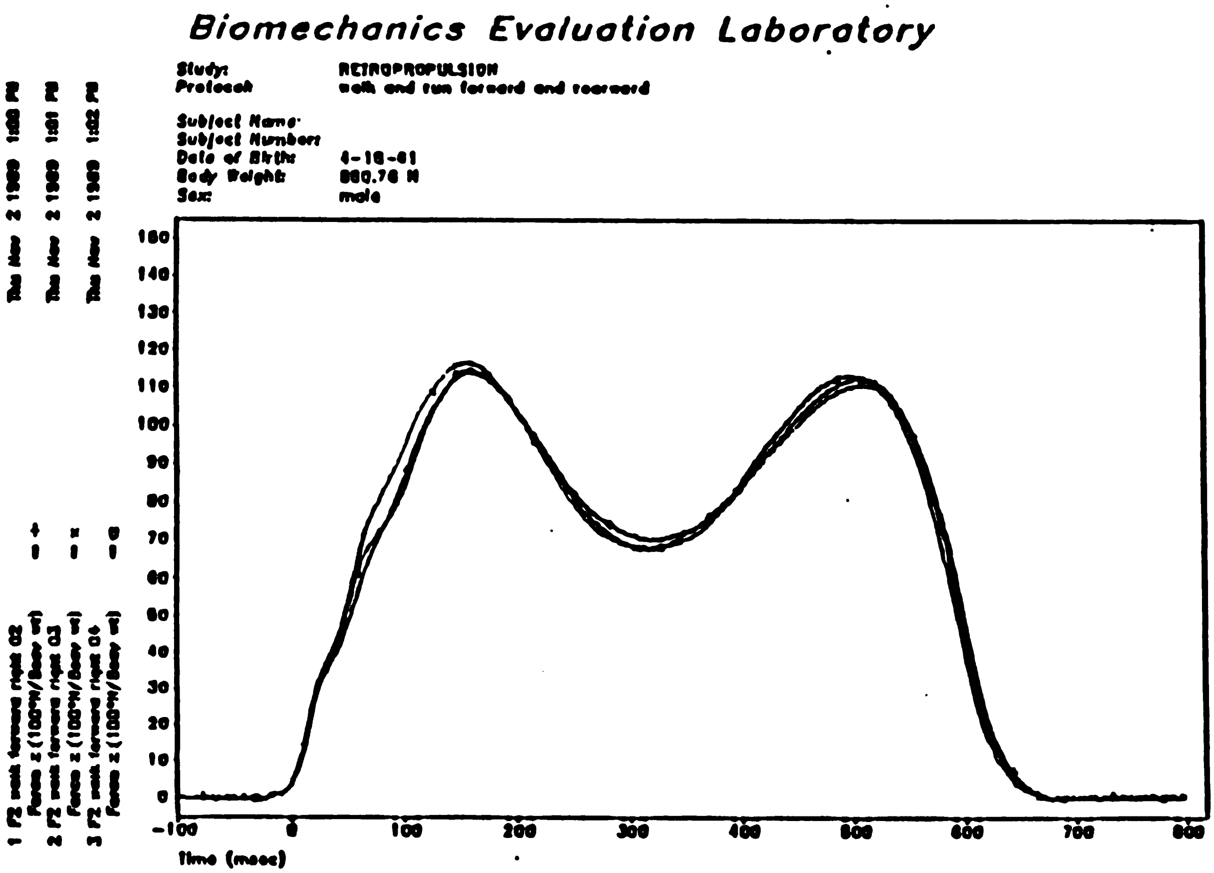


Figure 6

Reproducibility of Z - force data

IV. ANALYTICAL METHODS

Each video file was tracked using the track operator of the EV3D software. The stick figure option was utilized allowing the targets to be tracked as rigid links. Each target had a three dimensional path of motion. The target paths were smoothed using the EV3D track editor operator. The edited files then underwent the following analysis.

The knee angle was defined as the angle between the femur rigid link (targets 3-4) and the shank rigid link (targets 5-6). An offset angle was calculated from the standing file allowing the knee angle to be independent of exact placement of the targets. The EV3D angle operator calculated the knee angle. Each knee angle file was then differentiated to produce a knee angular velocity file.

Target 4 was assumed to be the saggital plane knee joint center and the X (X_4) and Z (Z_4) coordinates of this target in laboratory space was required during each instance of foot contact with the force plate. The location of target 4 and the knee angular velocity file (ω) during stance were exported to the Belcalc(50) and Calc_twf(51) programs. The force plate Y force (Fy) and Z force (Fz) as well as the Y coordinate of the center of pressure (COPy) were combined in the following manner to calculate the knee muscle moments.

(1) $\text{Moment} = (-Fy \cdot Z_4) + (Fz \cdot (X_4 - COPy))$

The muscle power was then calculated by equation (2).

(2) $P = M \cdot w$

Equation (3) will yield the work performed by the muscles.

(3) $W = \int_0^t P(t')dt'$

The synchronization of the kinetic (1000Hz) data and the kinematic (60Hz) data was accomplished by taking kinetic data at each 16 millisecond time interval.

The mean peak positive and mean peak negative power was computed from 3 trials of each condition for each subject. Within subject analysis of forward walking-rearward walking and forward running-rearward running was performed with a paired student t-test with a level of significance of $p<0.05$.

The mean negative and positive work was analyzed in the same manner.

The EMG signals for the six muscle groups were subject to the following analysis. First, the mean pattern of muscle activity was determined and graphed in percent of stance phase. Second, the mean duration of total on time was determined and expressed in percentage of stance phase. The final step compared the mean duration of on time within the forward-rearward walking and running conditions utilizing a paired student t-test with a level of significance of $p<0.05$.

The ground reaction torque was subjectively compared for direction and timing during stance phase of each condition.

An overall picture of motion, forces, and muscle activity and the clinical significance of these is then presented.

V. RESULTS AND DISCUSSION

The following chapter is subdivided into a walking and running section. The results of the knee joint muscle moment, power, and work are presented. The EMG activity and ground reaction torque are also presented. Each of the graphs represent the right stance phase of gait. Time zero corresponds to foot strike (0% stance).

An example of the calc_twf program output for subject F5 walking forward right is given in Figure 7. These figures demonstrate the combined force and motion data with resultant knee muscle moment and muscle power numerical output.

WALKING

The knee muscle moment curves for the six subjects walking forward are presented in Figure 8. Each graph shows the three trials from each subject. Extension of the knee joint is positive (+) and flexion is negative (-).

The shape of the forward walking knee muscle moment curves compare favorably with the results of previous authors.(1,38,53,54,57) The knee joint moment demonstrated a momentary flexor pattern during the first few percent of stance, followed by a extensor response which assisted in

Kinematic .vid file name: f5wfrl1.vid
Kinematic .ted file name: f5wfrl.ted
Kinematic .dif file name: f5wfrlK.dif
Force file name: A2.fpt

Target 4: R LAT FEM CON

Time (sec)	x coord (m)	z coord (m)	angular velocity (rad/sec)	y force (N)	z force (N)	y cop (m)
-0.050	-0.3295	0.5455	2.6307	14.2832	-2.2778	0.5201
-0.033	-0.2995	0.5426	3.0303	15.0405	-2.2300	0.5454
-0.017	-0.2631	0.5117	2.8126	15.7905	-2.1841	0.5691
0.000	-0.2354	0.5409	2.2502	21.0205	45.5220	0.1228
0.017	-0.2043	0.5411	2.3693	47.0908	298.4834	0.1221
0.033	-0.1734	0.5127	2.5853	-48.0112	396.8916	0.1083
0.050	-0.1431	0.5452	2.7881	-80.9756	537.0718	0.0964
0.067	-0.1144	0.5474	2.6139	-105.1963	600.5449	0.0835
0.083	-0.0888	0.5485	1.9223	-147.8335	681.0001	0.0673
0.100	-0.0676	0.5480	1.0110	-168.7163	789.3183	0.0563
0.117	-0.0508	0.5468	0.2614	-170.9639	916.4360	0.0497
0.133	-0.0375	0.5457	-0.2463	-158.9243	982.3535	0.0441
0.150	-0.0264	0.5451	-0.6004	-143.1748	995.1884	0.0367
0.167	-0.0170	0.5450	-0.8364	-120.7607	984.6910	0.0282
0.183	-0.0087	0.5450	-0.9634	-106.5474	932.3048	0.0176
0.200	-0.0012	0.5452	-1.0419	-92.2632	856.3193	0.0031
0.217	0.0059	0.5456	-1.1389	-77.9976	783.2587	-0.0126
0.233	0.0127	0.5465	-1.2766	-68.9829	718.7358	-0.0285
0.250	0.0191	0.5477	-1.3837	-56.2176	657.3510	-0.0425
0.267	0.0252	0.5492	-1.3606	-41.9932	622.5347	-0.0531
0.283	0.0310	0.5508	-1.2433	-36.7319	593.4448	-0.0612
0.300	0.0365	0.5521	-1.1767	-29.2471	576.1025	-0.0666
0.317	0.0424	0.5538	-1.2121	-21.0395	561.7482	-0.0709
0.333	0.0489	0.5549	-1.2881	-12.8232	562.2232	-0.0740
0.350	0.0561	0.5556	-1.2763	-3.1289	565.6938	-0.0768
0.367	0.0649	0.5558	-1.1060	7.3257	589.9356	-0.0784
0.383	0.0712	0.5559	-0.8151	16.2983	623.0328	-0.0795
0.400	0.0843	0.5561	-0.5720	28.9932	671.0996	-0.0803
0.417	0.0954	0.5565	-0.3119	43.1611	719.2451	-0.0810
0.433	0.1079	0.5569	-0.0518	58.8037	758.5615	-0.0823
0.450	0.1221	0.5571	0.2413	79.6484	795.2197	-0.0828
0.467	0.1384	0.5568	0.6242	104.2285	829.1372	-0.0837
0.483	0.1570	0.5561	1.0265	125.1157	862.9321	-0.0851
0.500	0.1778	0.5551	1.3721	145.2500	881.8701	-0.0875
0.517	0.2012	0.5537	1.6874	161.6699	885.8716	-0.0898
0.533	0.2275	0.5517	2.0623	176.5273	863.0728	-0.0924
0.550	0.2573	0.5487	2.6355	186.8745	804.4414	-0.0958
0.567	0.2908	0.5446	3.4842	184.5141	703.6207	-0.1009
0.583	0.3284	0.5392	4.4217	163.5517	566.2359	-0.1087
0.600	0.3698	0.5329	5.1587	132.2163	416.4326	-0.1185
0.617	0.4117	0.5262	5.5435	94.1318	275.1055	-0.1258
0.633	0.4622	0.5201	5.6409	57.5034	151.5142	-0.1352
0.650	0.5118	0.5154	5.4955	20.8628	57.4458	-0.1421
0.667	0.5624	0.5129	5.2118	-3.0420	17.4790	-0.1795
0.683	0.6076	0.5126	6.0056	-1.5210	2.8799	-0.0797
0.700	0.6667	0.5128	6.8760	-3.0278	2.7686	0.0441
0.717	0.7128	0.5192	0.0000	-0.7720	-0.0723	3.6659

Figure 7 Program Output calc_twf

Time (sec)	moment (N·m)	power (watts)
0.017	-50.0280	118.5302
0.033	0.2138	-0.5528
0.050	19.0981	-53.2478
0.067	39.0290	-102.0165
0.083	66.4350	-127.7048
0.100	83.5433	-84.4643
0.117	92.4602	-24.1724
0.133	93.1951	22.9521
0.150	88.2622	52.9935
0.167	76.8132	64.2443
0.183	66.3133	63.8875
0.200	51.9697	54.1481
0.217	37.3349	42.5215
0.233	26.3169	33.6337
0.250	15.1316	21.3520
0.267	7.3519	10.0032
0.283	2.2823	2.8376
0.300	-1.1713	-1.3783
0.317	-4.3866	-5.3169
0.333	-6.9918	-9.0061
0.350	-9.8106	-12.5212
0.367	-12.0288	-13.3035
0.383	-12.3738	-10.4614
0.400	-13.4334	-7.6843
0.417	-13.6768	-4.2659
0.433	-13.3565	-0.7318
0.450	-13.1471	3.2123
0.467	-12.6983	7.9265
0.483	-7.6017	7.8027
0.500	-1.0205	1.1002
0.517	9.1719	-15.4770
0.533	19.2773	-39.7558
0.550	27.3519	-72.0871
0.567	33.1616	-115.5407
0.583	36.2279	-160.1871
0.600	34.1916	-176.3831
0.617	29.9517	-166.0361
0.633	19.6480	-110.8326
0.650	10.4816	-57.6014
0.667	8.2533	-43.2619
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0.700	3.5205	-24.2073
0.717	0.0844	-0.0000

Figure 7 (cont'd.).

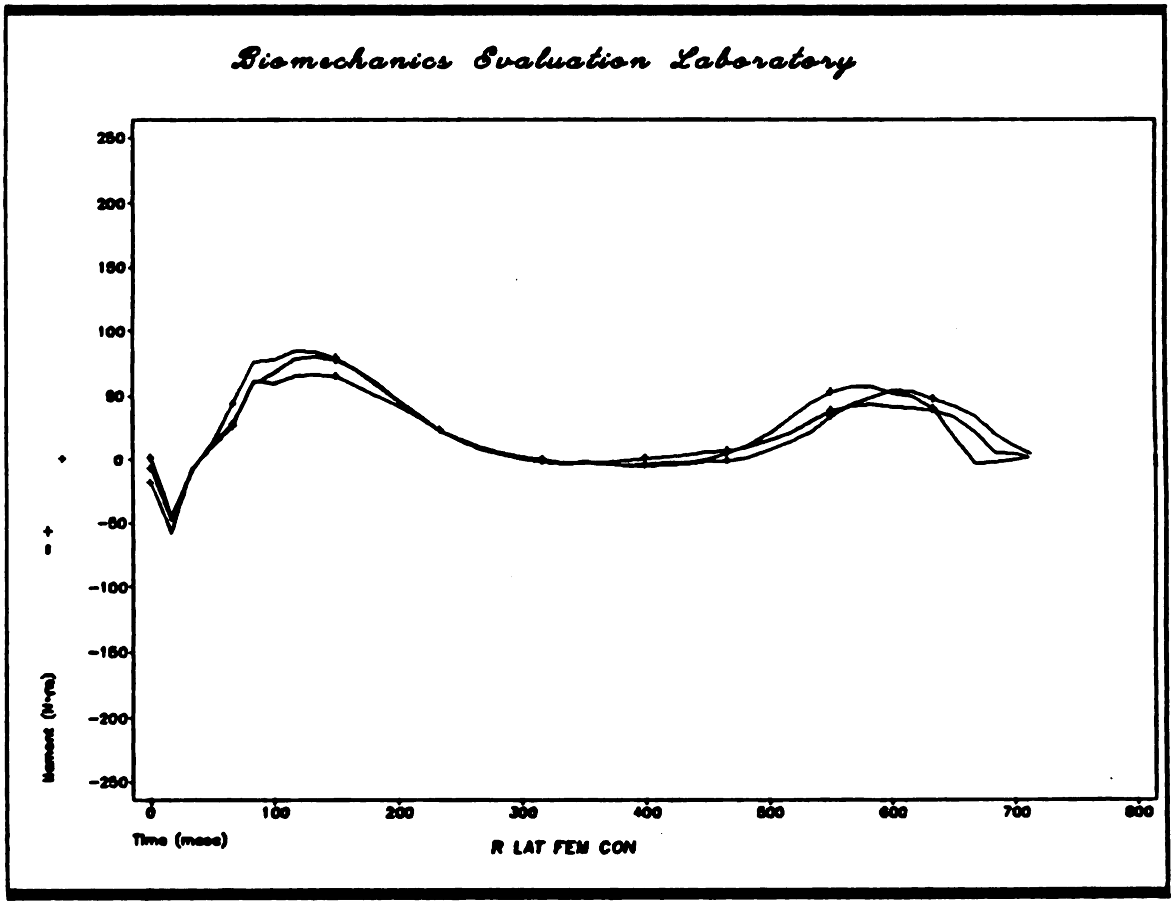

Figure 8a Subject F1

Figure 8 Forward Walking Knee Muscle Moment Curves

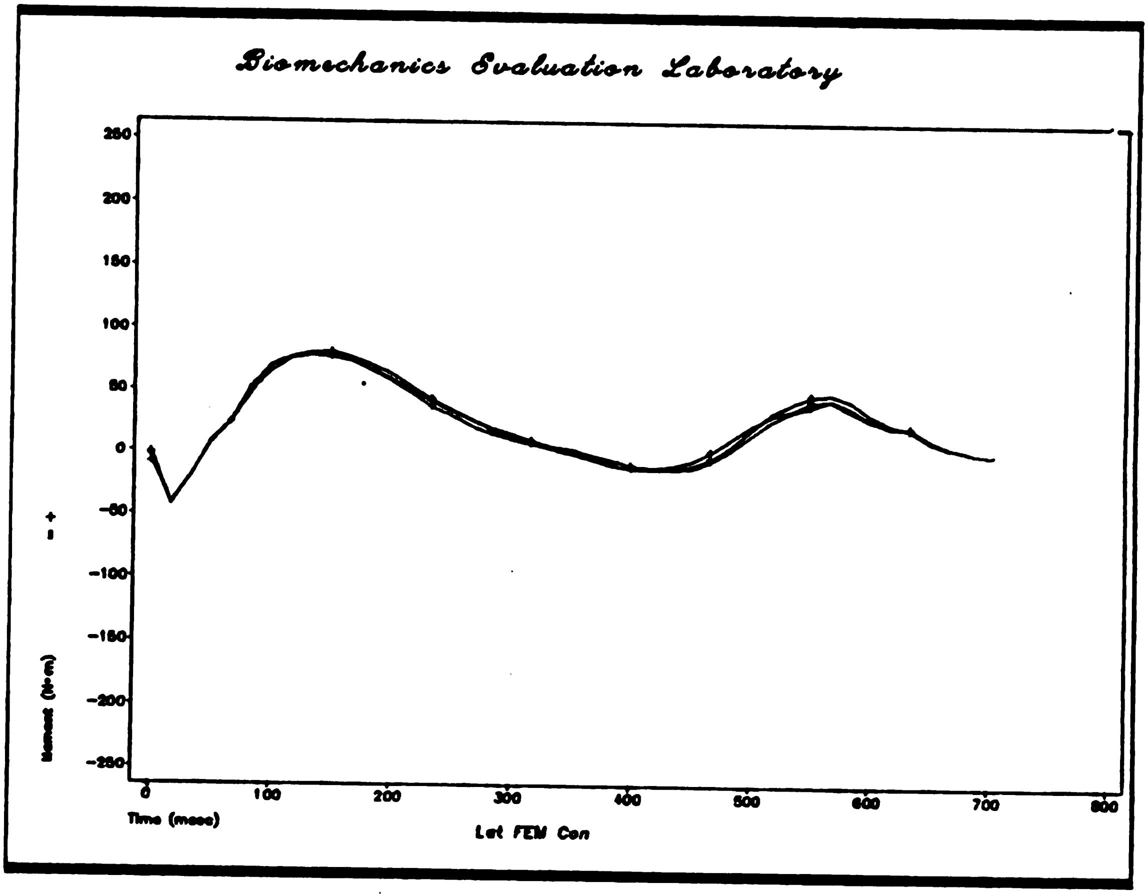

Figure 8b Subject F2

Figure 8 (cont'd.).

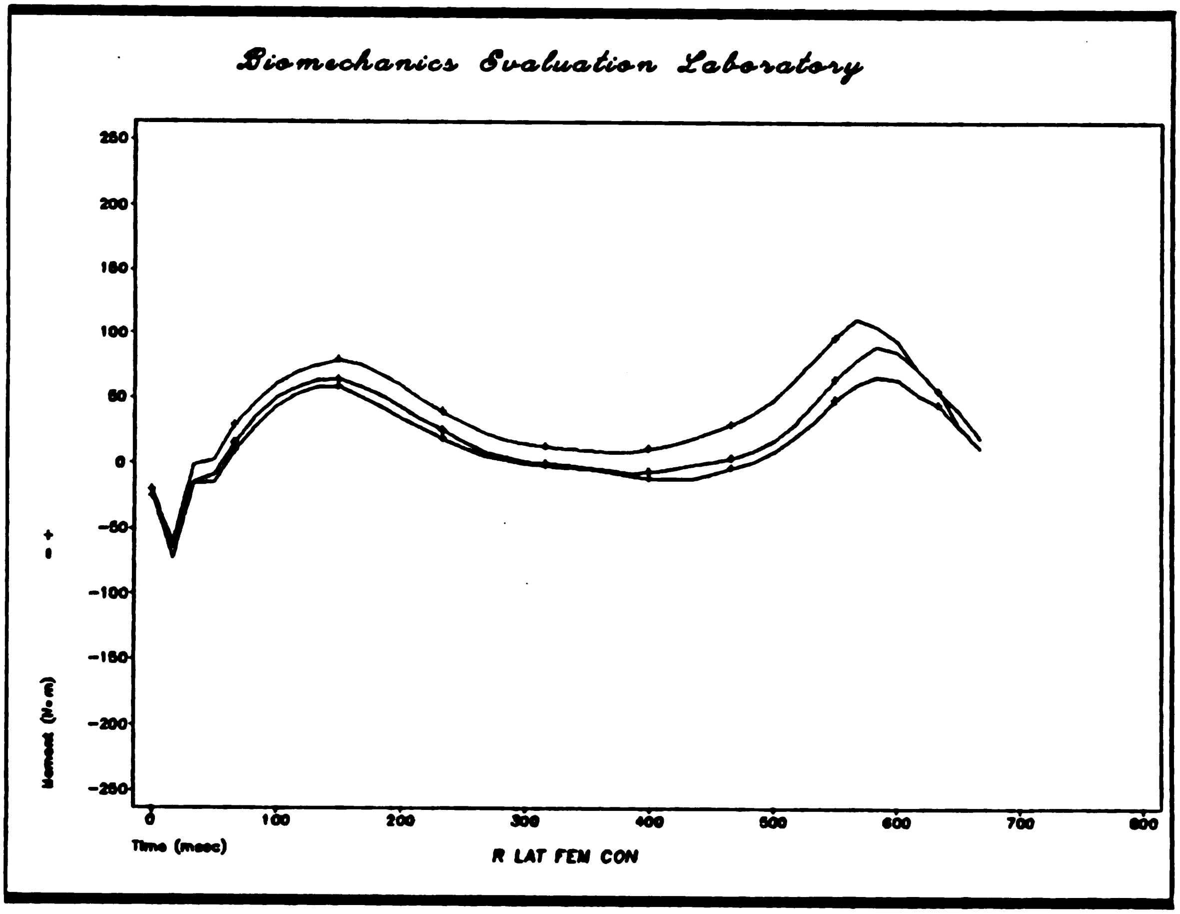

Figure 8c Subject F3

Figure8 (cont'd.).

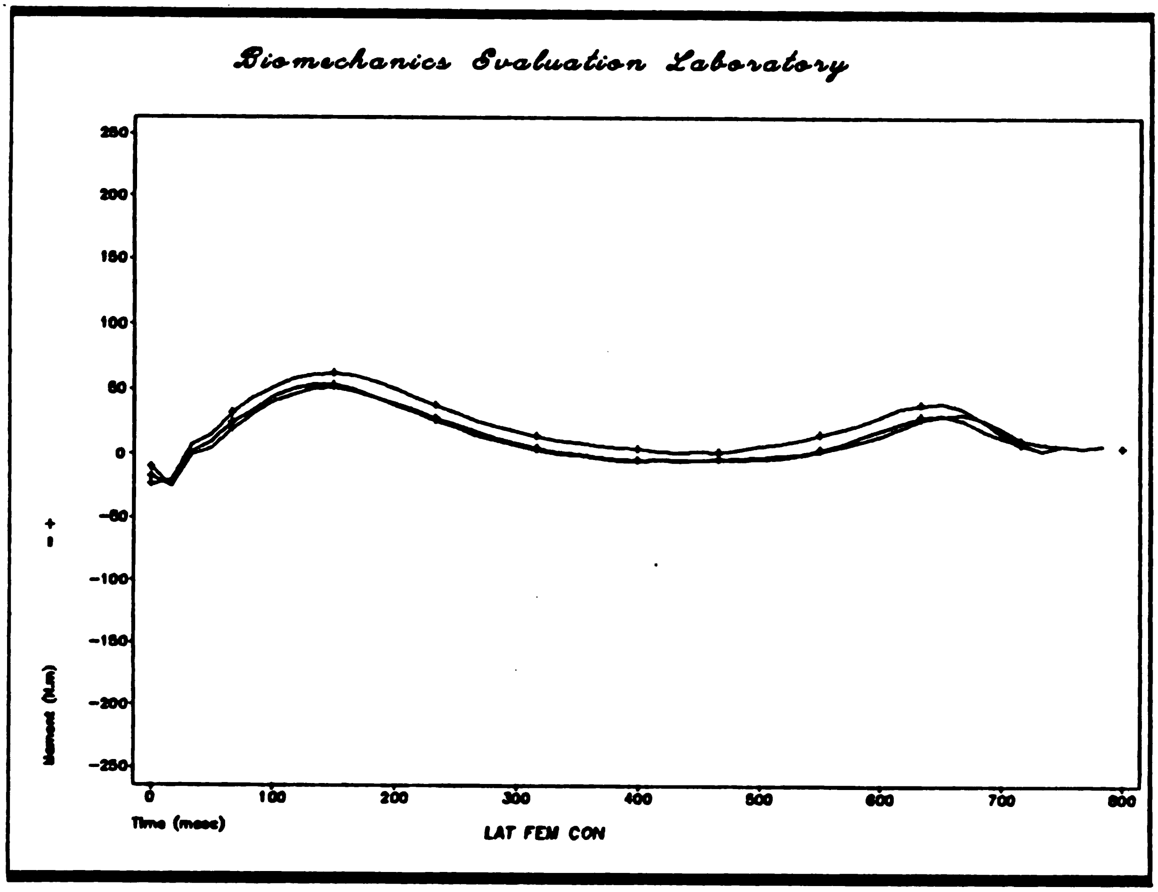

Figure 8d Subject F4

Figure 8 (cont'd.).

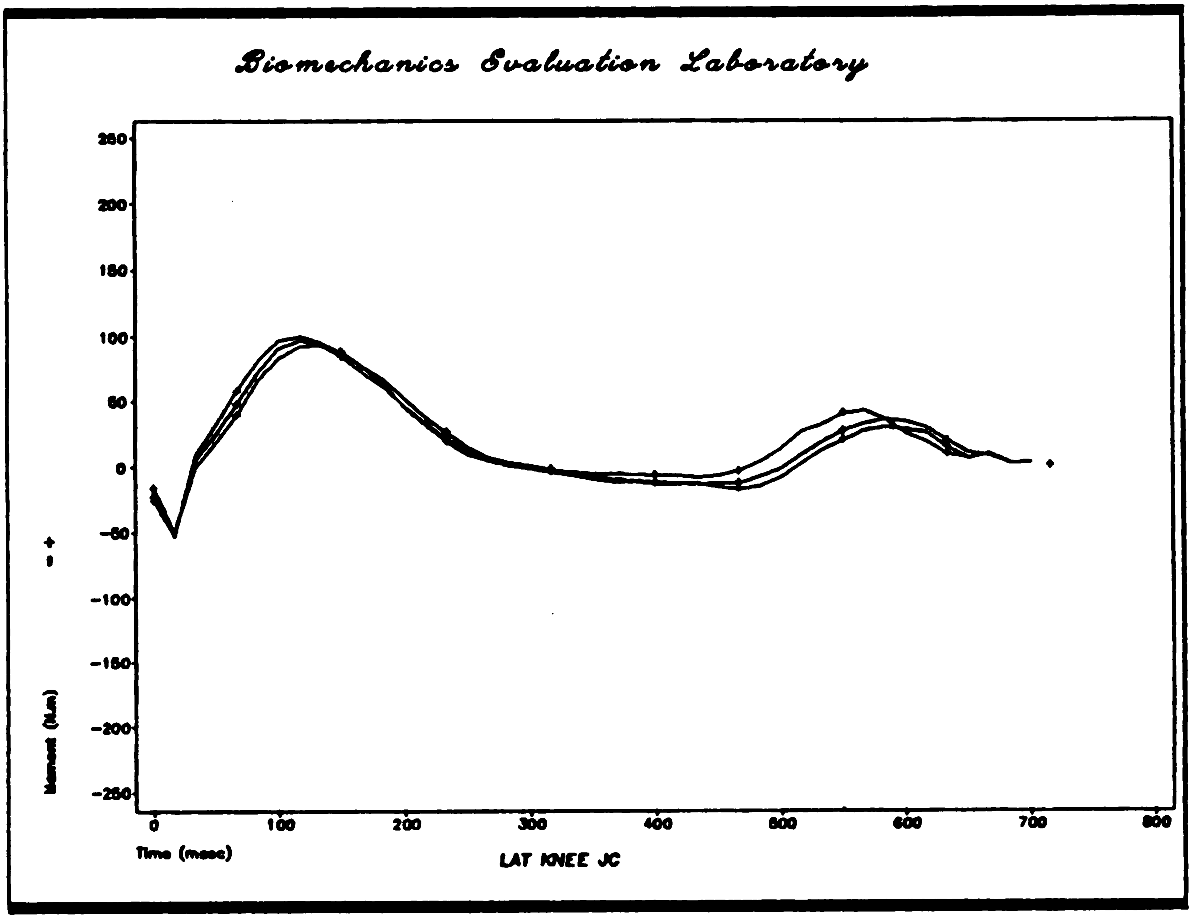

Figure 8e Subject F5

Figure 8 (cont'd.).

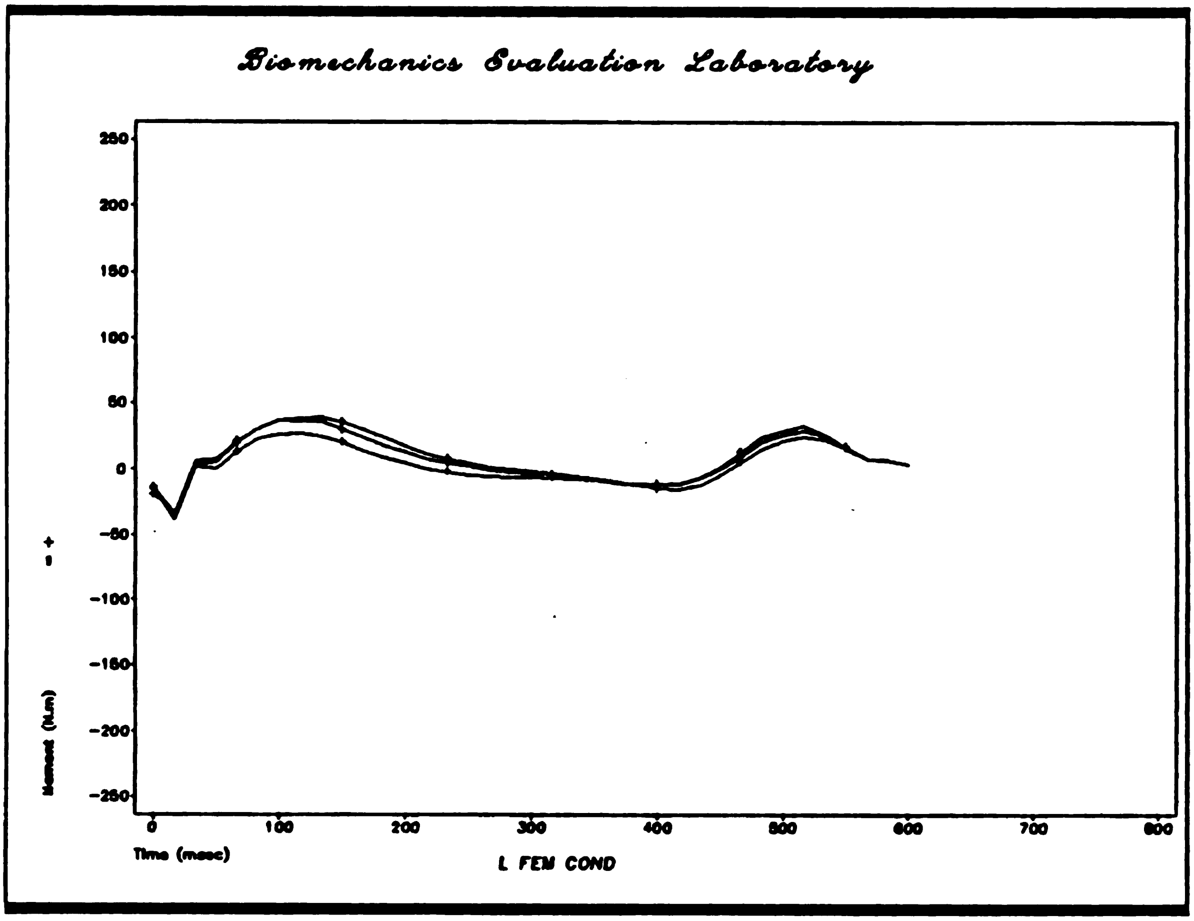

Figure 8f Subject F6

Figure 8 (cont'd.).

arresting knee flexion as full weight bearing occurred. During late stance a slight flexor moment occured before a second extensor moment was seen during terminal stance. The forward curves demonstrated low intrasubject variability. Higher variability was noted across subjects. Winter (54) described the increased intersubject variability to be higher at slower speeds. He postulated that this was a result of the fact that one's natural cadence is accomplished at a subconscious level and well within the extremes of forces possible at each joint. As speed increases higher joint forces are achieved, and a conscious over-ride of the loose walking patterns is necessary.

The knee muscle moment for the six subjets walking rearward are presented in Figure 9. Each graph shows the three trials from each subject for the respective condition. Extension of the knee joint is positive (+) and flexion is negative (-). In rearward walking the knee muscle moment demonstrated an extensor dominance throughout stance phase in all subjects walking rearward. The magnitudes of the muscle moment were consistently lower than those noted in forward walking for the same subject. The rearward walking knee muscle moment patterns demonstrated consistency within subject trials with increased variability across subjects.

The representative sample of the knee muscle power curve for each of the six subjects walking forward are presented in Figure 10. A positive value indicates concentric knee extensor muscle activity or eccentric knee

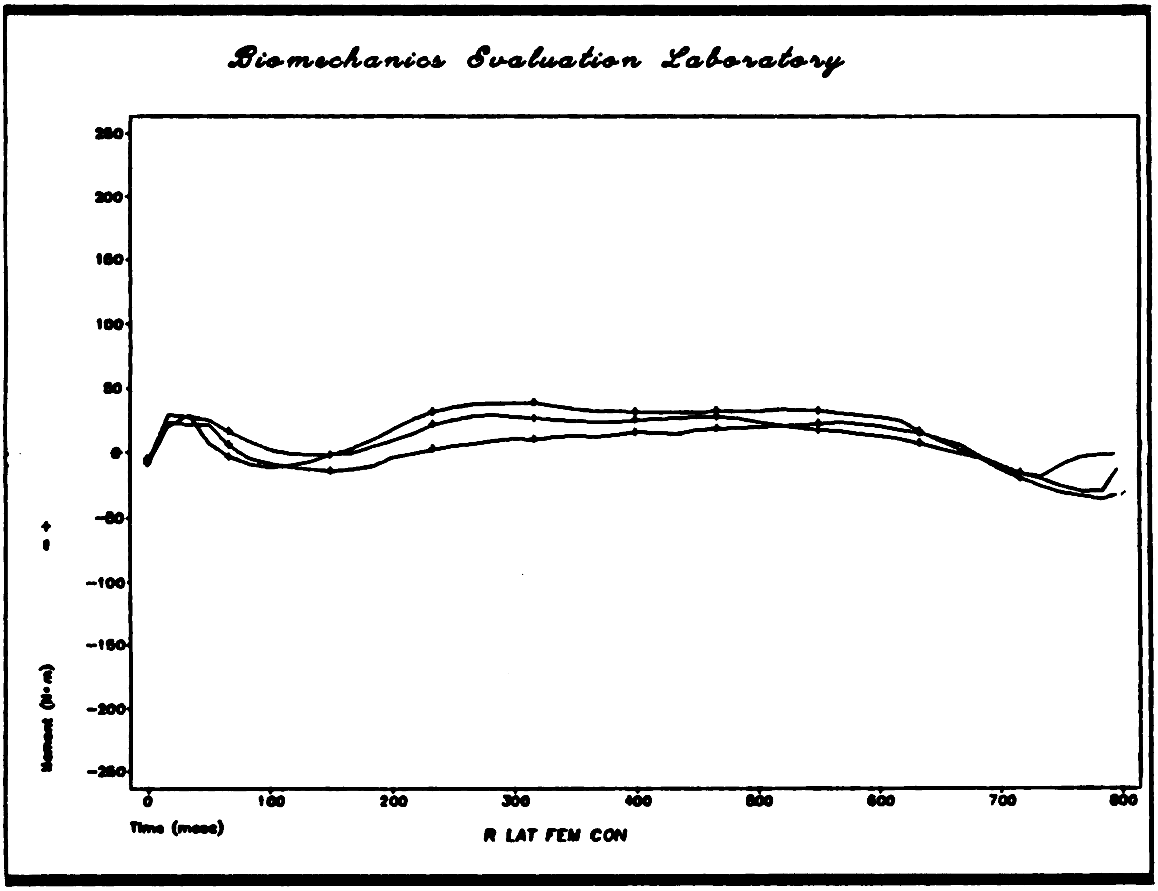

Figure 9a Subject F1

Figure 9 Rearward Walking Knee Muscle Moment Curves

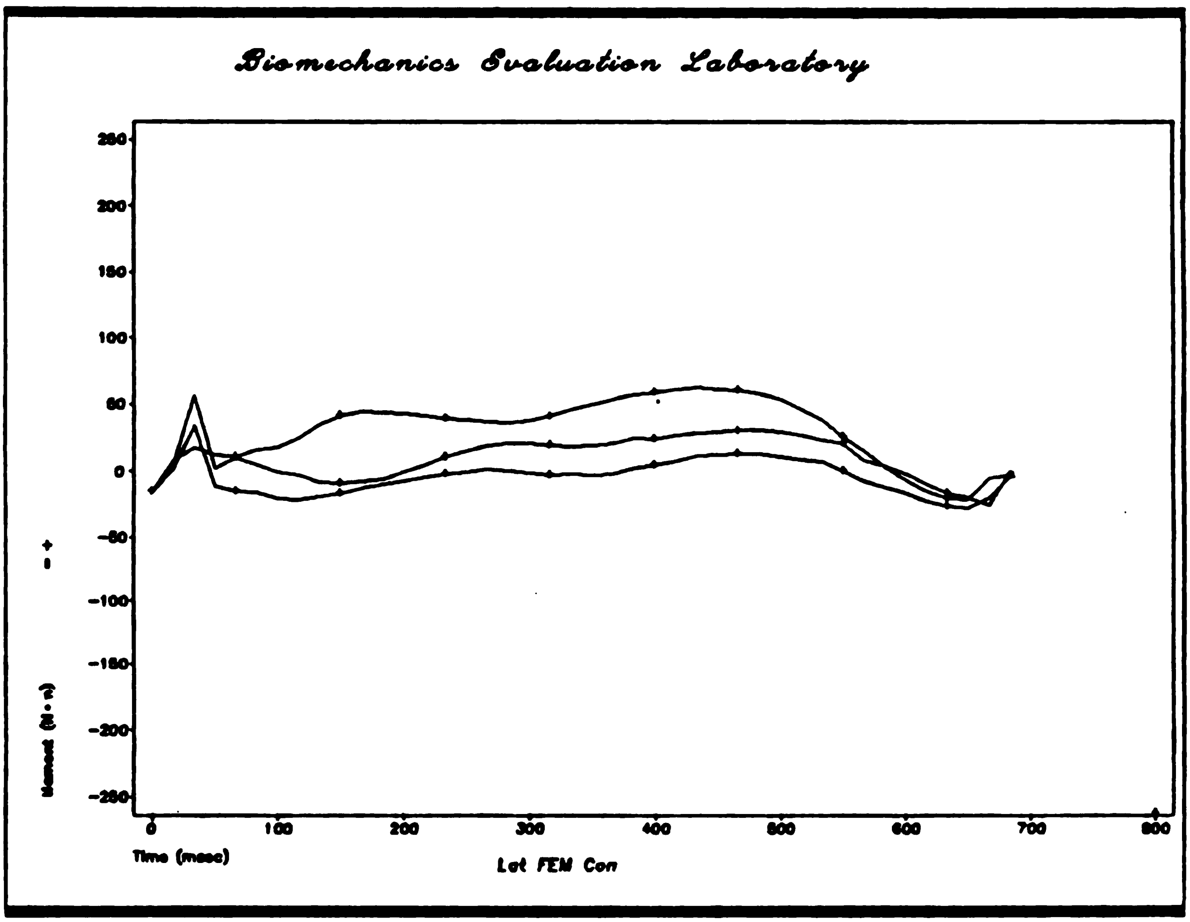

Figure 9b Subject F2

Figure 9 (cont'd.).

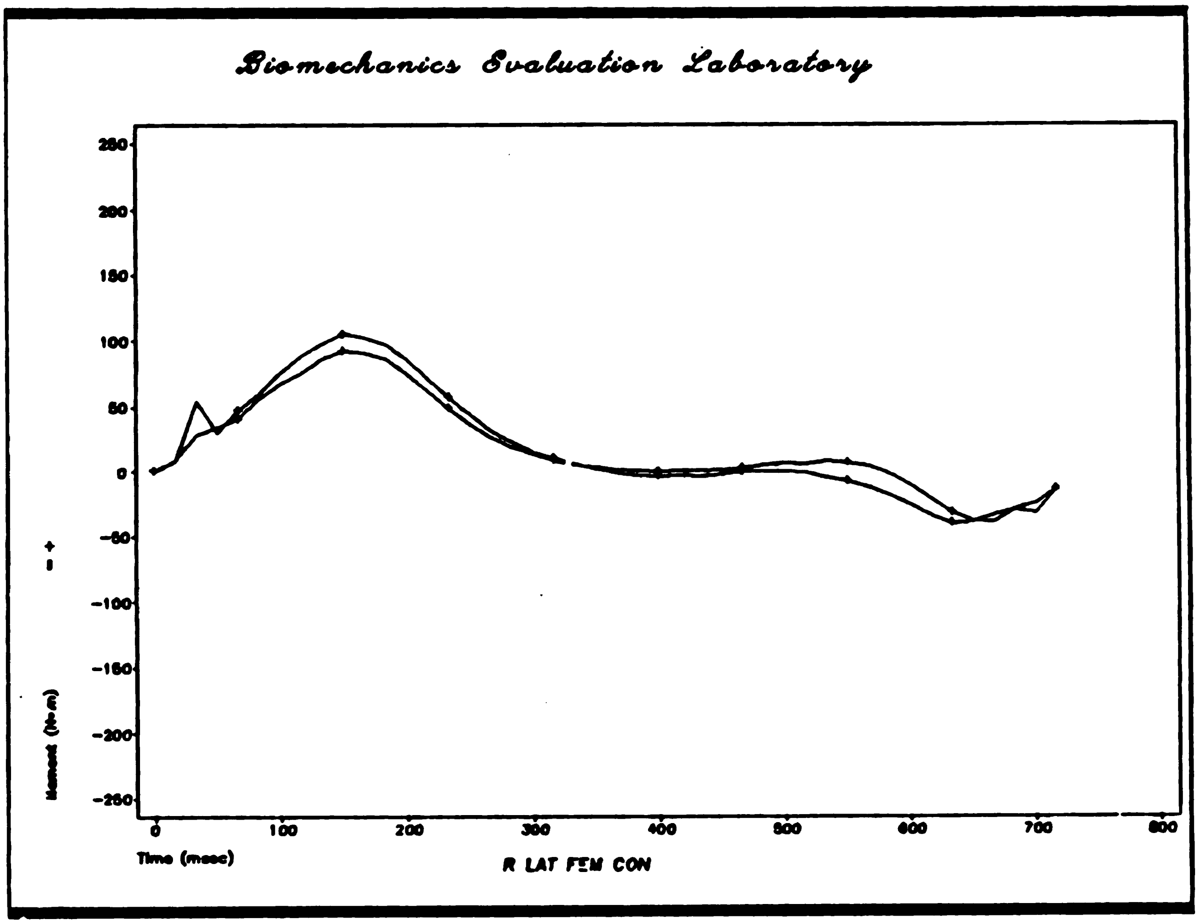

Figure 9c Subject F3

Figure 9 (cont'd.).

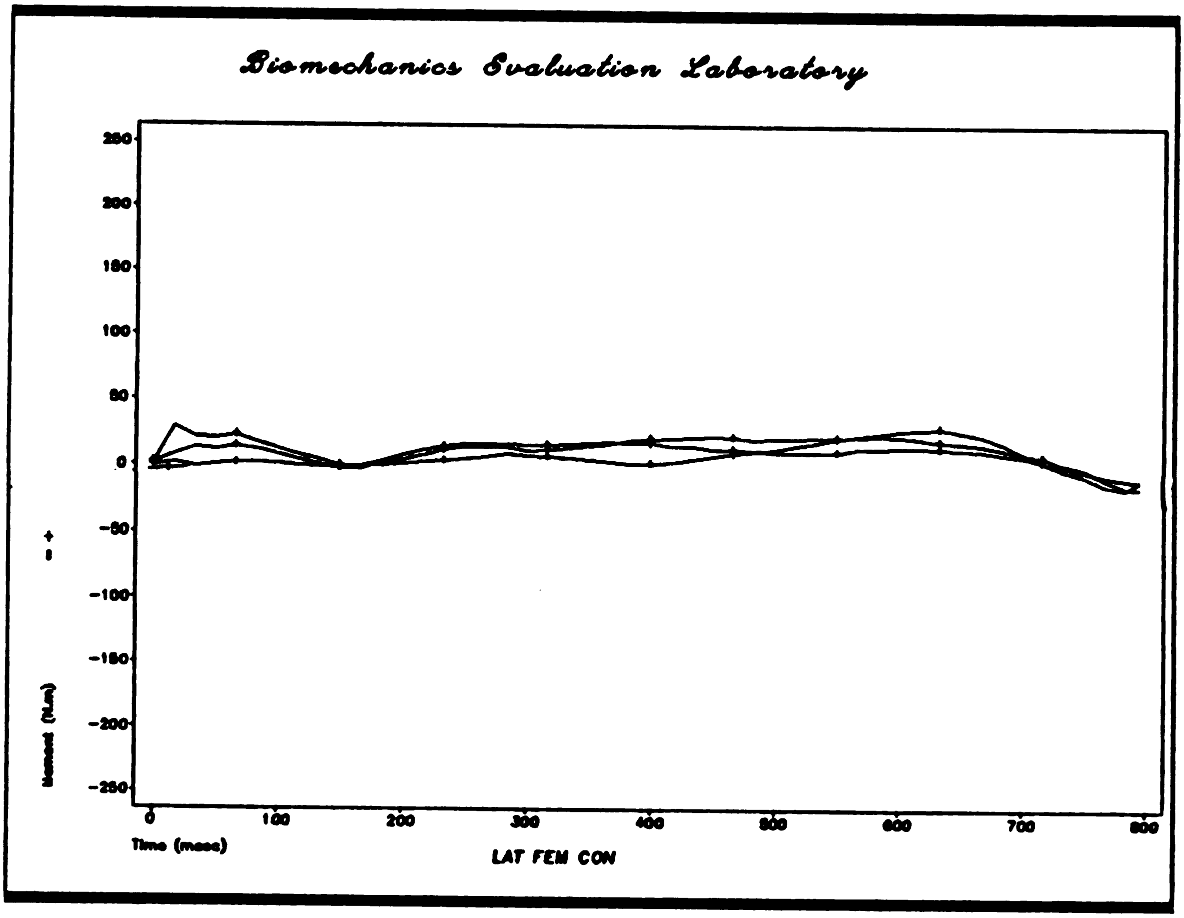

Figure 9d Subject F4

Figure 9 (cont'd.).

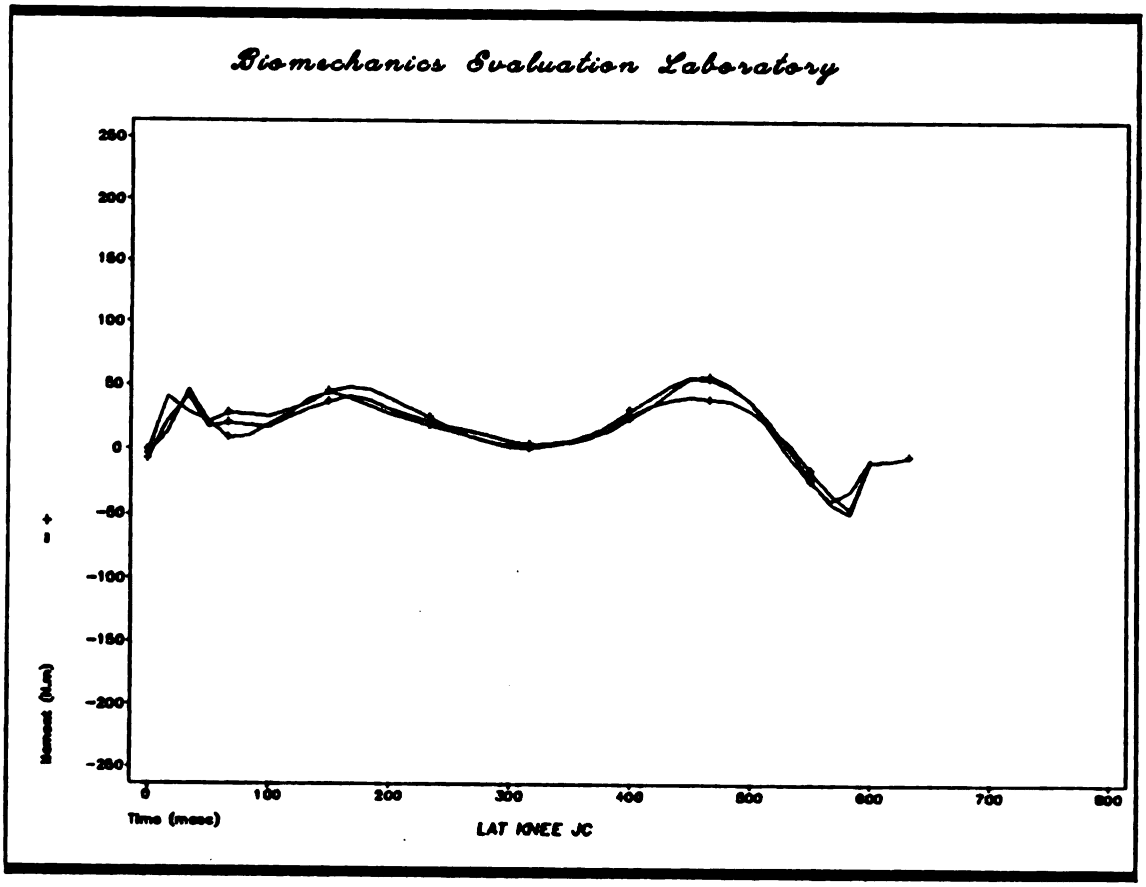

Figure 9e Subject F5

Figure 9 (cont'd.).

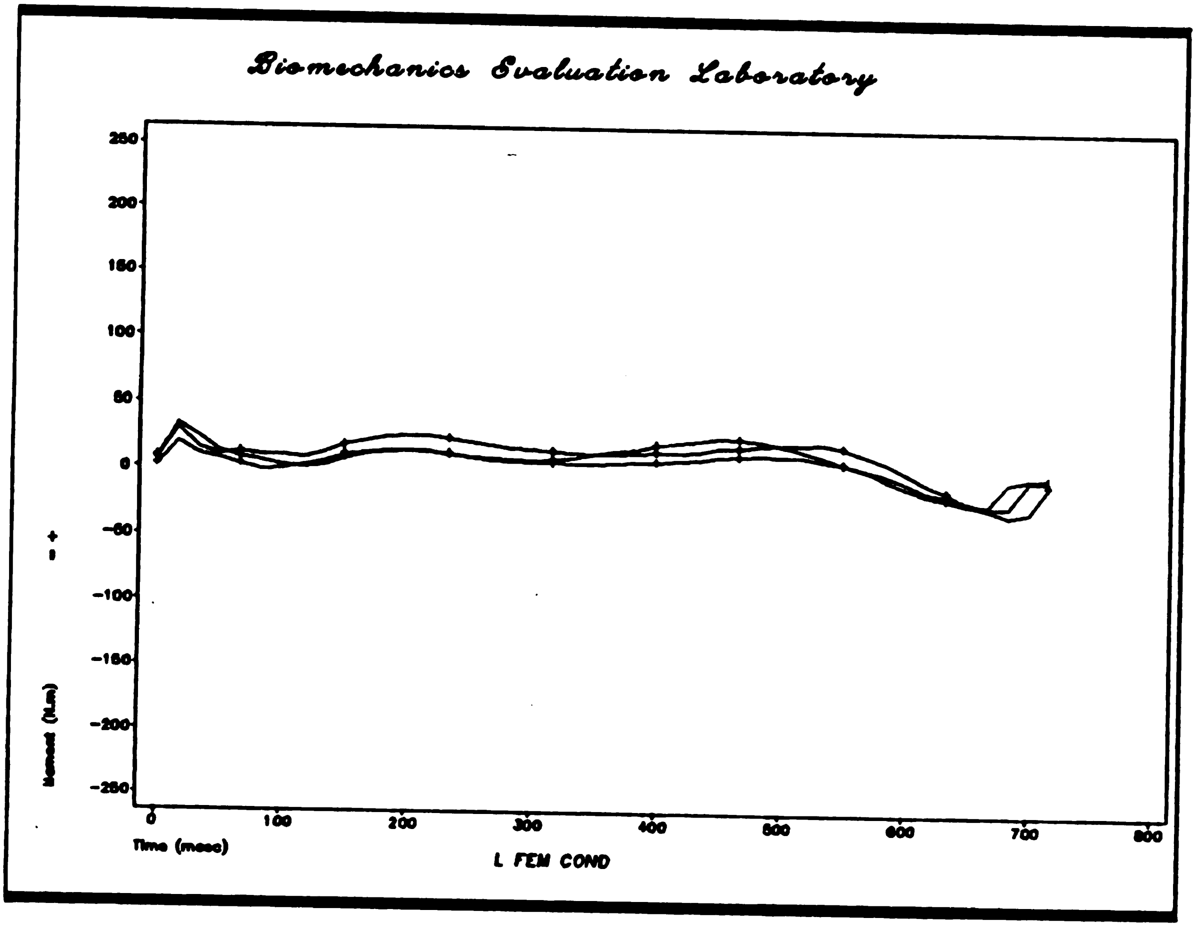

Figure 9f Subject F6

Figure 9 (cont'd.).

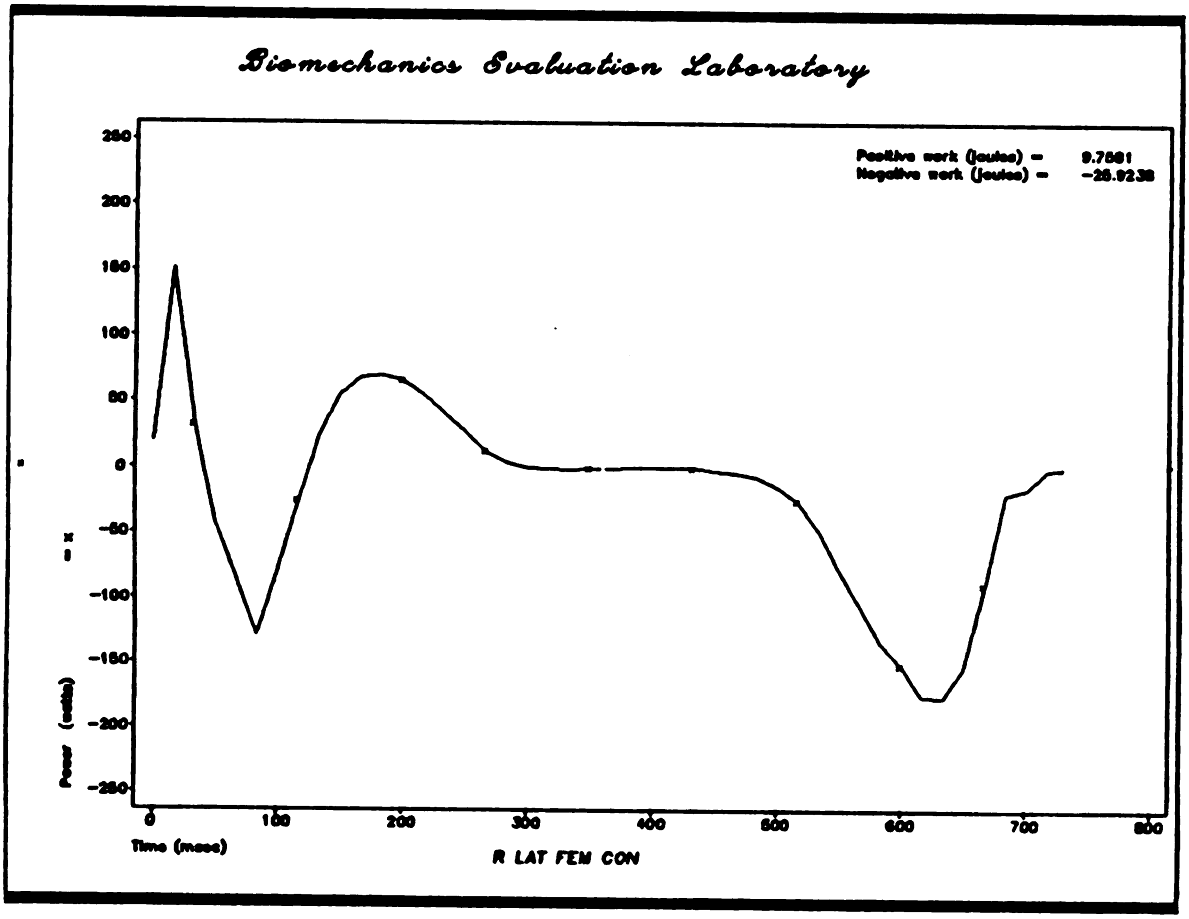

Figure 10a Subject F1

Figure 10 Forward Walking Power Curves

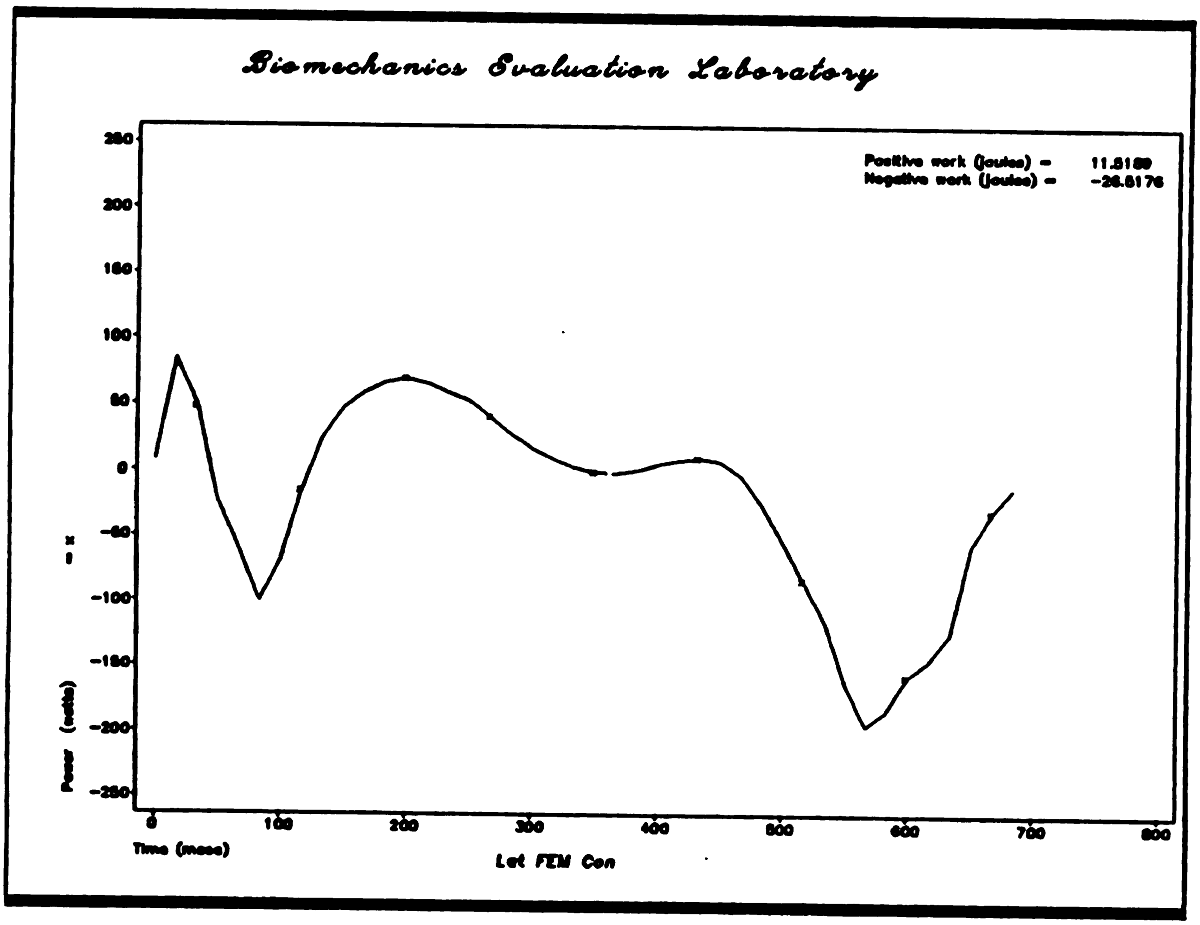

Figure 10b Subject F2

Figure 10 (cont'd.).

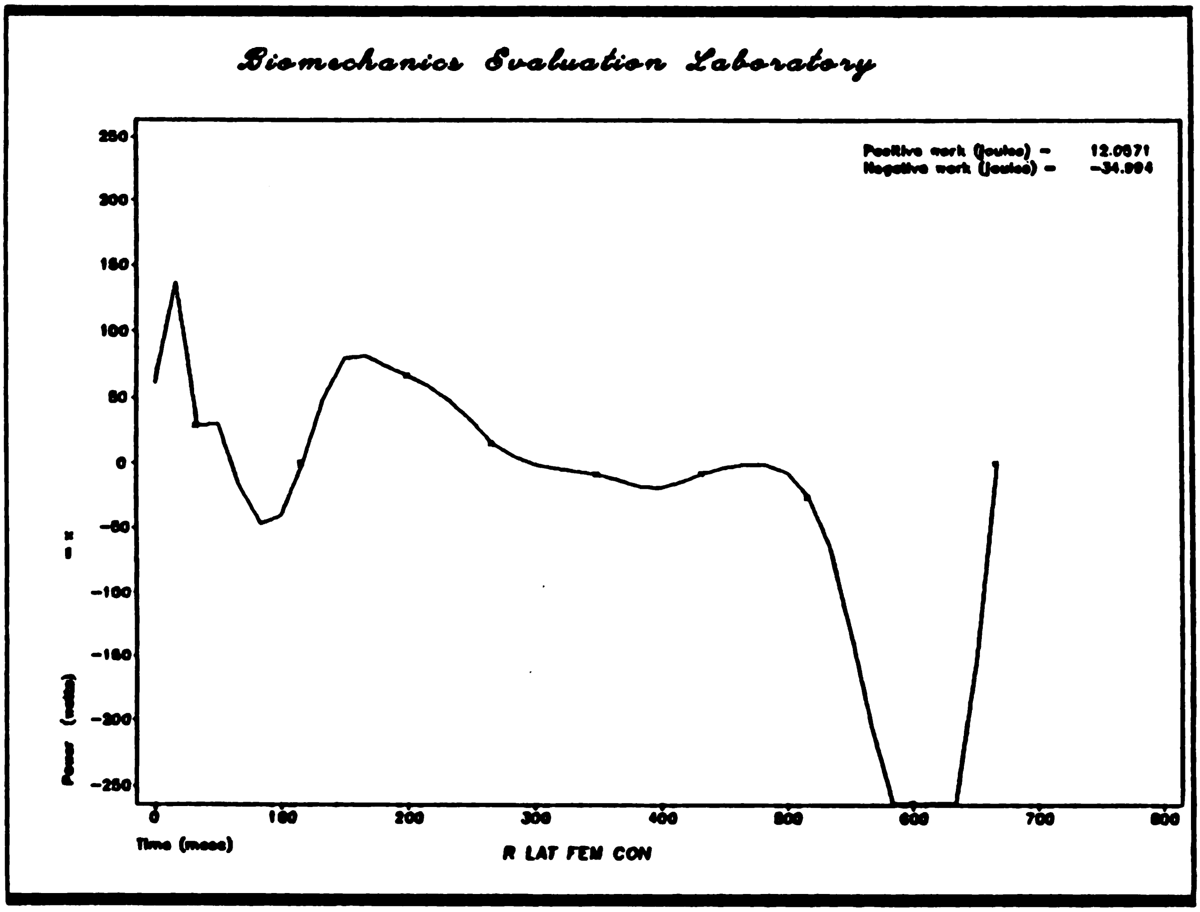

Figure 10c Subject F3

Figure 10 (cont'd.).

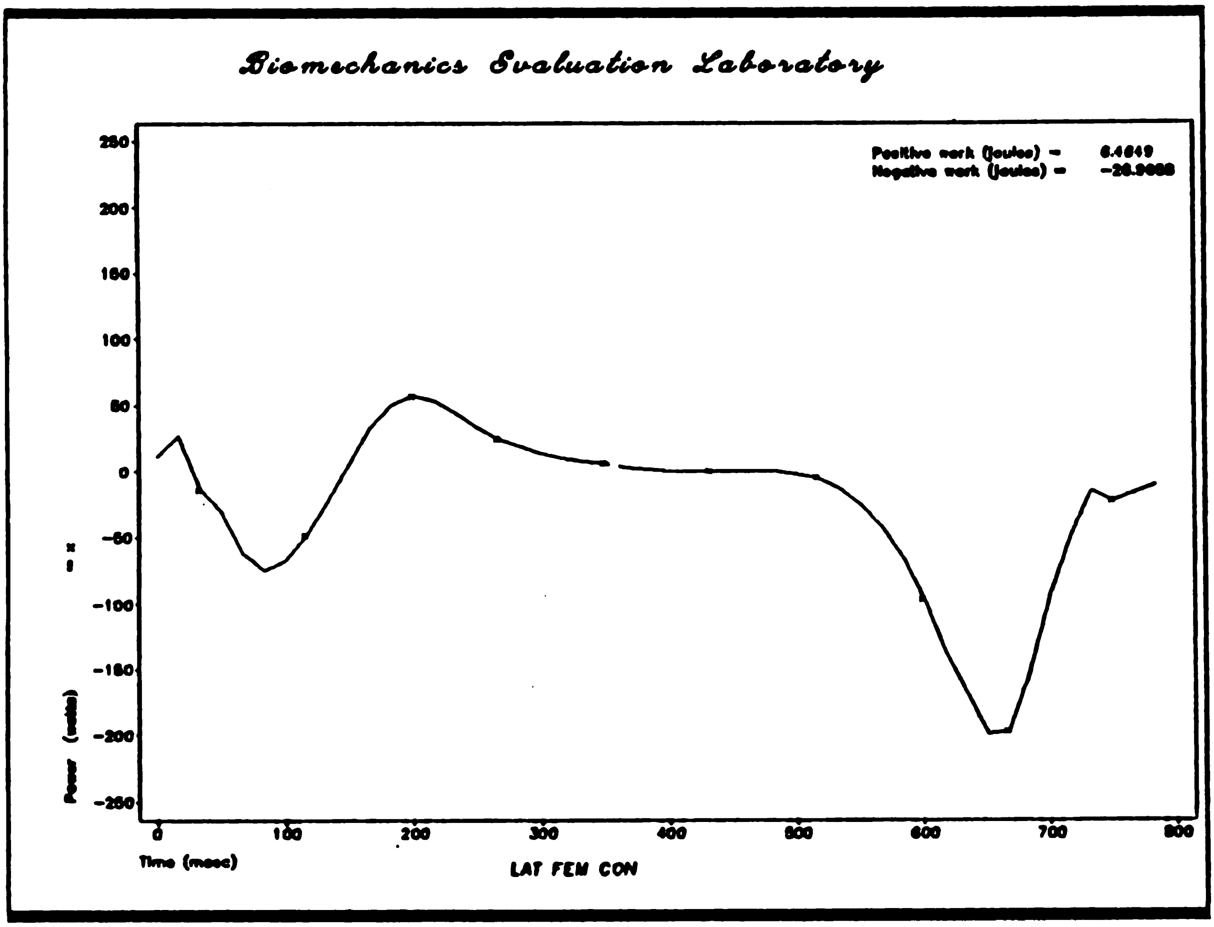

Figure 10d Subject F4

Figure 10 (cont'd.).

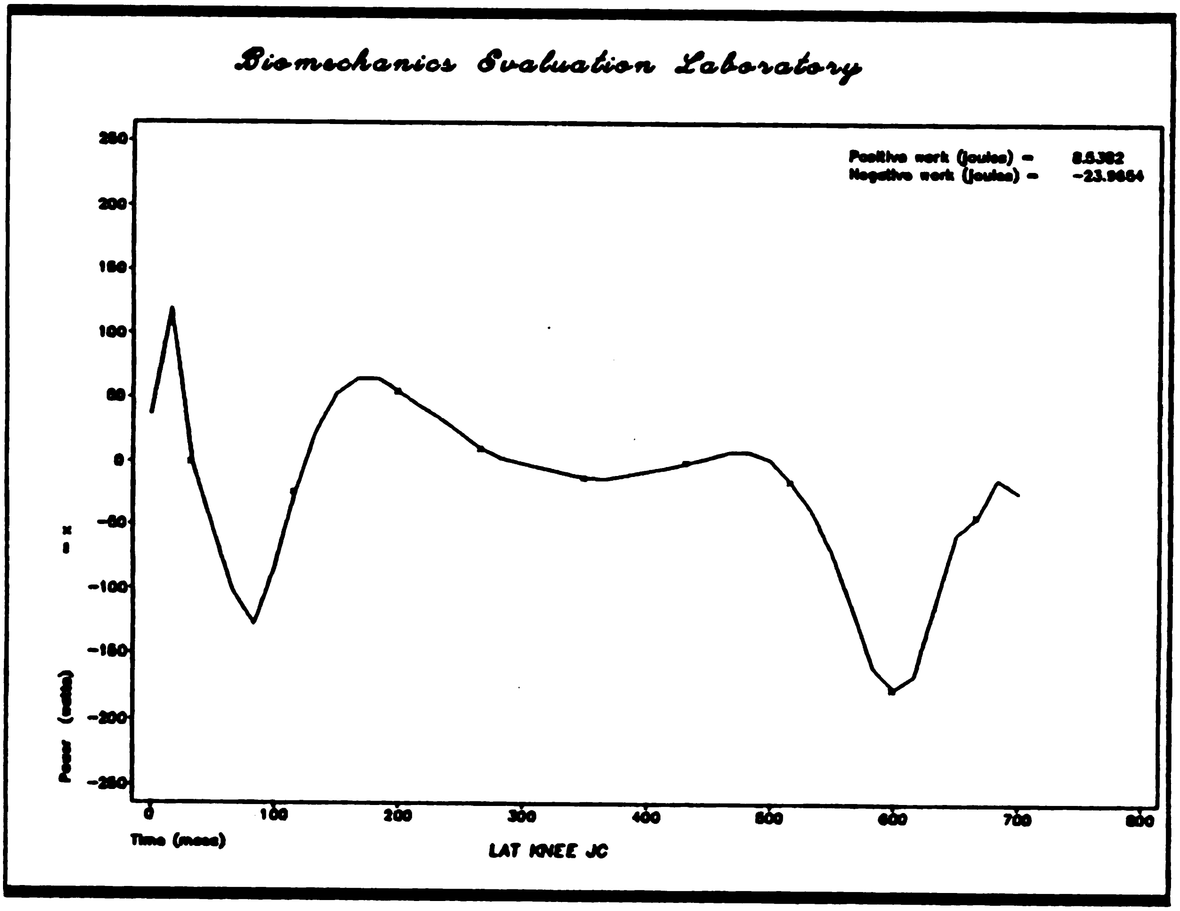

Figure 10e Subject F5

Figure 10 (cont'd.).

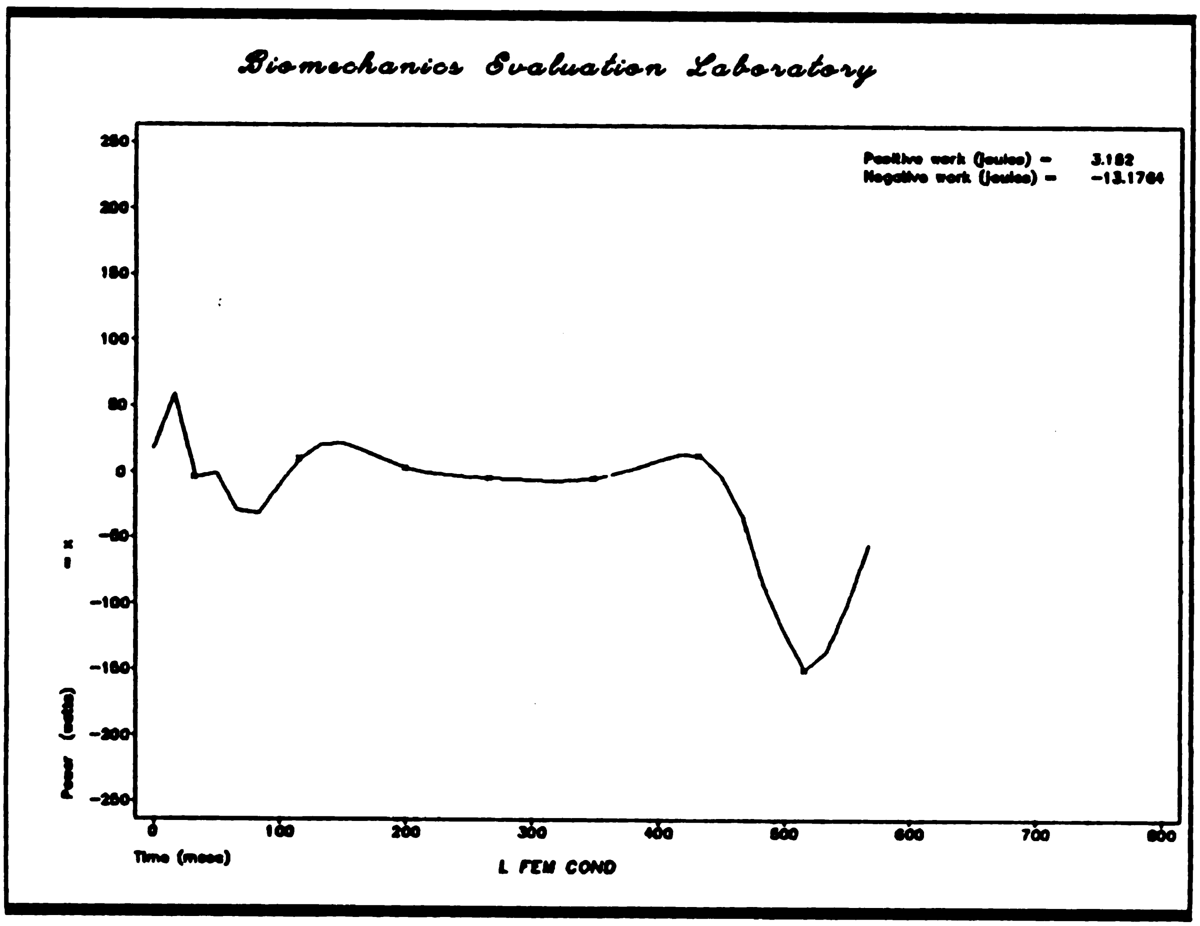

Figure 10f Subject F6

Figure 10 (cont'd.).

flexor muscle activity. Conversely, a negative value indicates eccentric knee extensor muscle activity or concentric knee flexor muscle activity.

The shape of the forward walking power curves compared favorably with Zarrugh(57) who predicted joint power based on kinematic data only. Generally, there were low power requirements in the knee during forward walking. The power curve begins in a positive direction then demonstrated a negative braking phase during early stance, which was followed by a moderate positive phase corresponding to the swinging forward of the contralateral limb, finally there was another relatively large negative phase which corresponded to the flexion of the knee in late stance.

The representative sample of the rearward walking knee muscle power curves for each of the six subjects are presented in Figure 11. A positive value indicates concentric knee extensor muscle activity or eccentric knee flexor muscle activity. Conversely, a negative value indicates eccentric knee extensor muscle activity or concentric knee flexor muscle activity.

The rearward walking power curves showed low power requirements throughout stance. Five of the six subjects demonstrated two positive peaks, one during initial stance and one during late stance. The knee muscular contribution to braking in rearward walking was apparently with concentric quadriceps activity which contrasts with the

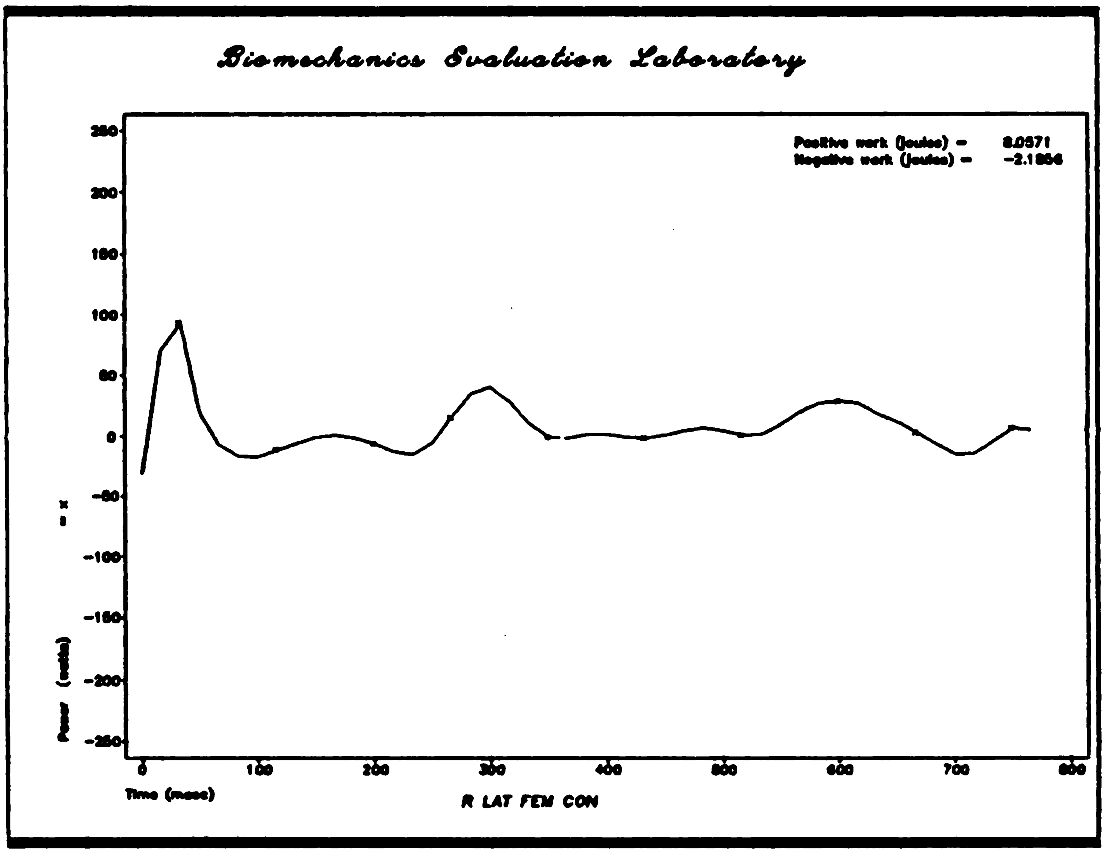

Figure 11a Subject F1

Figure 11 Rearward Walking Power Curves

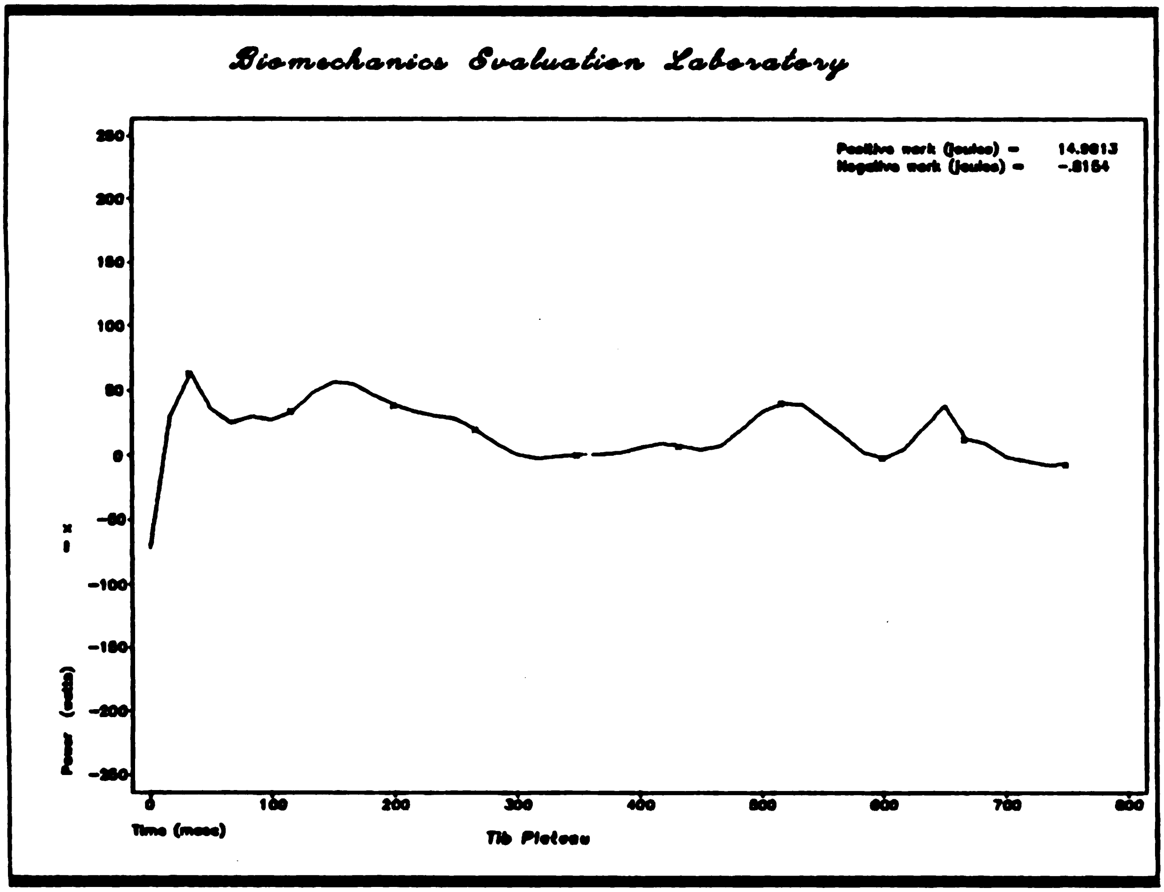

Figure 11b Subject F2

Figure 11 (cont'd.).

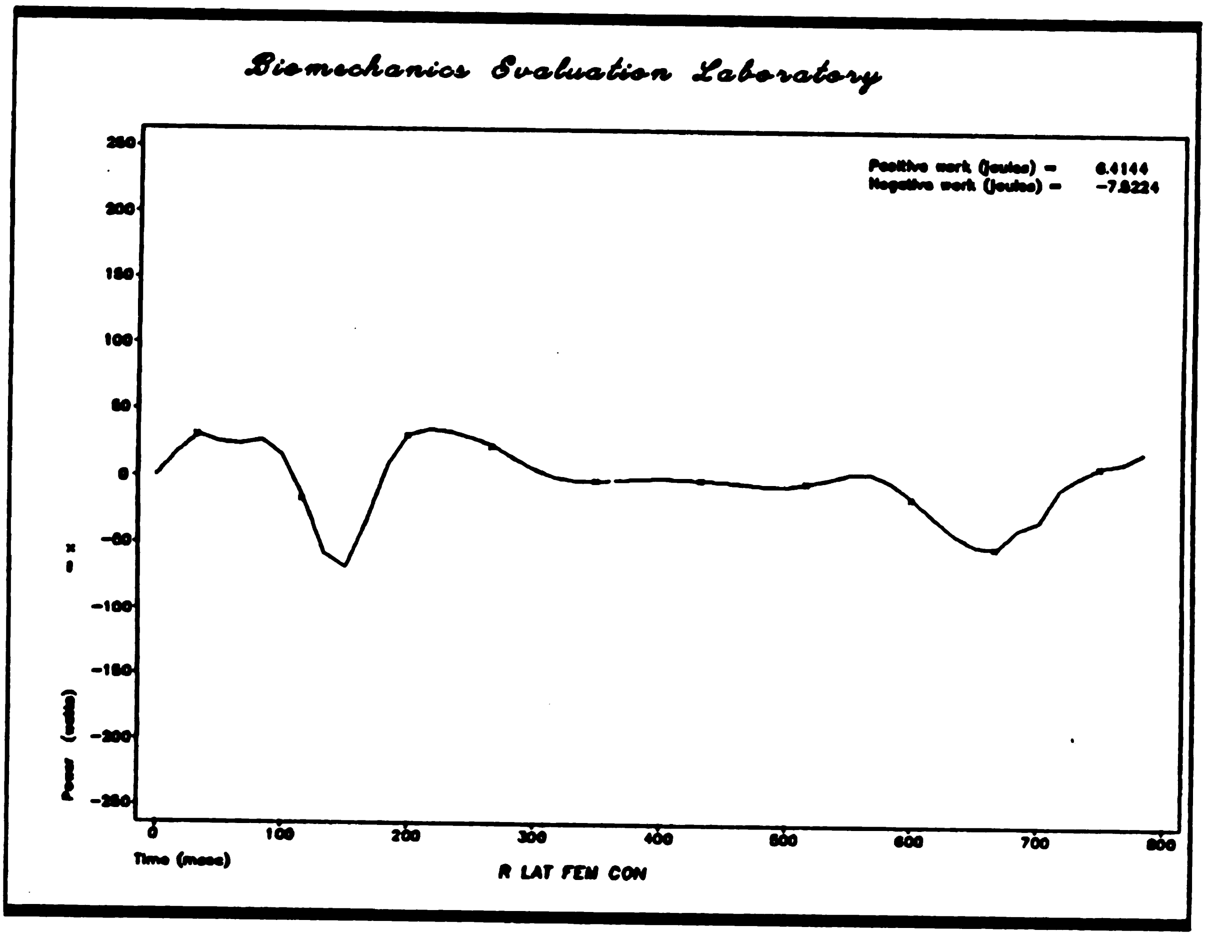

Figure 11c Subject F3

Figure 11 (cont'd.).

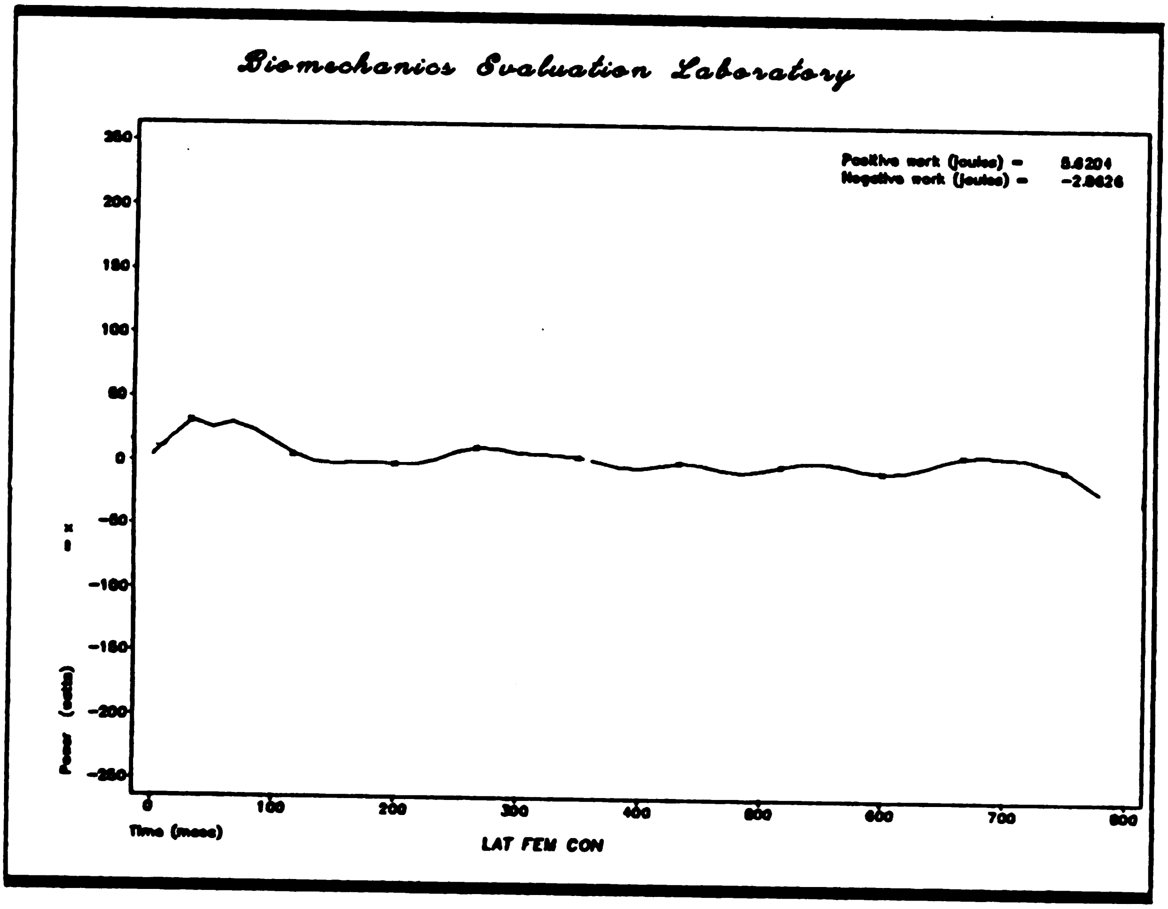

Figure 11d Subject F4

Figure 11 (cont'd.).

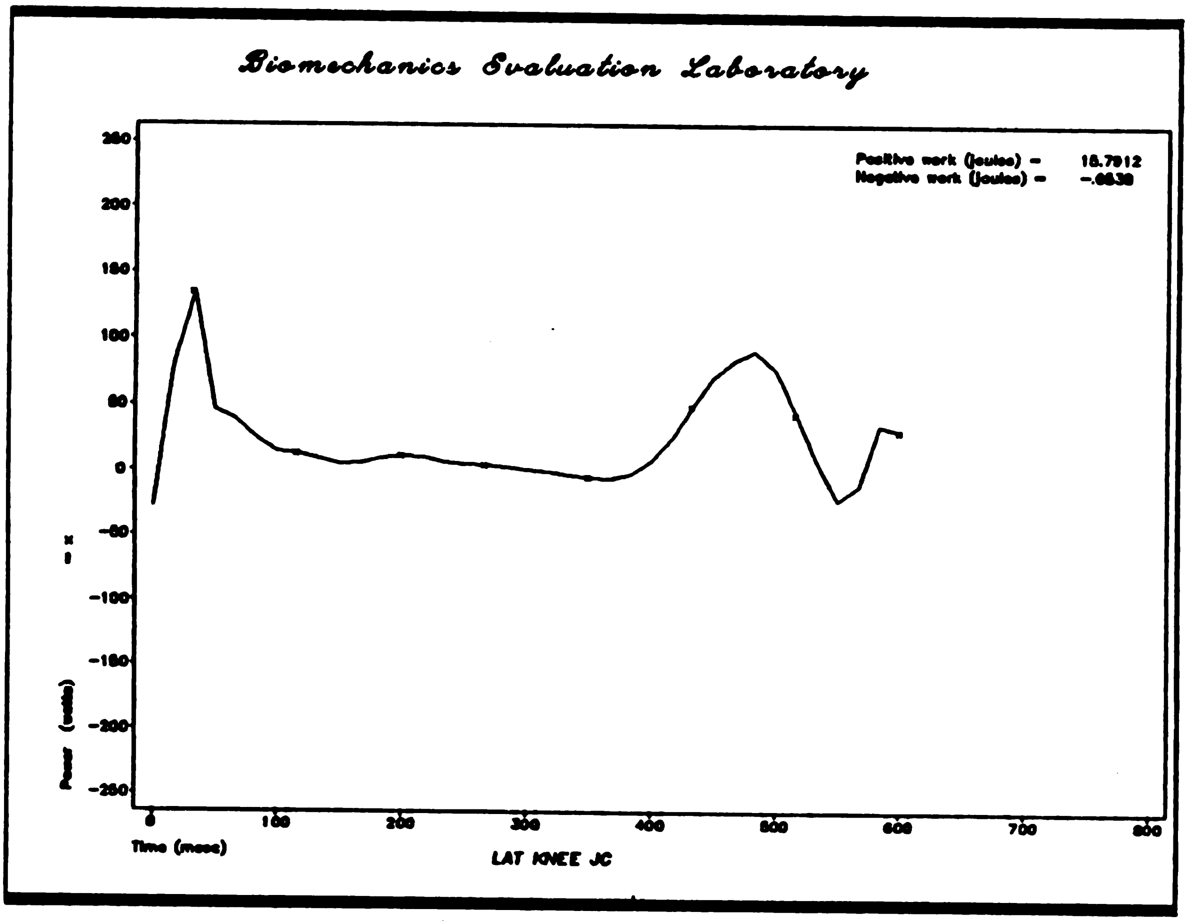

Figure 11e Subject F5

Figure 11 (cont'd.).

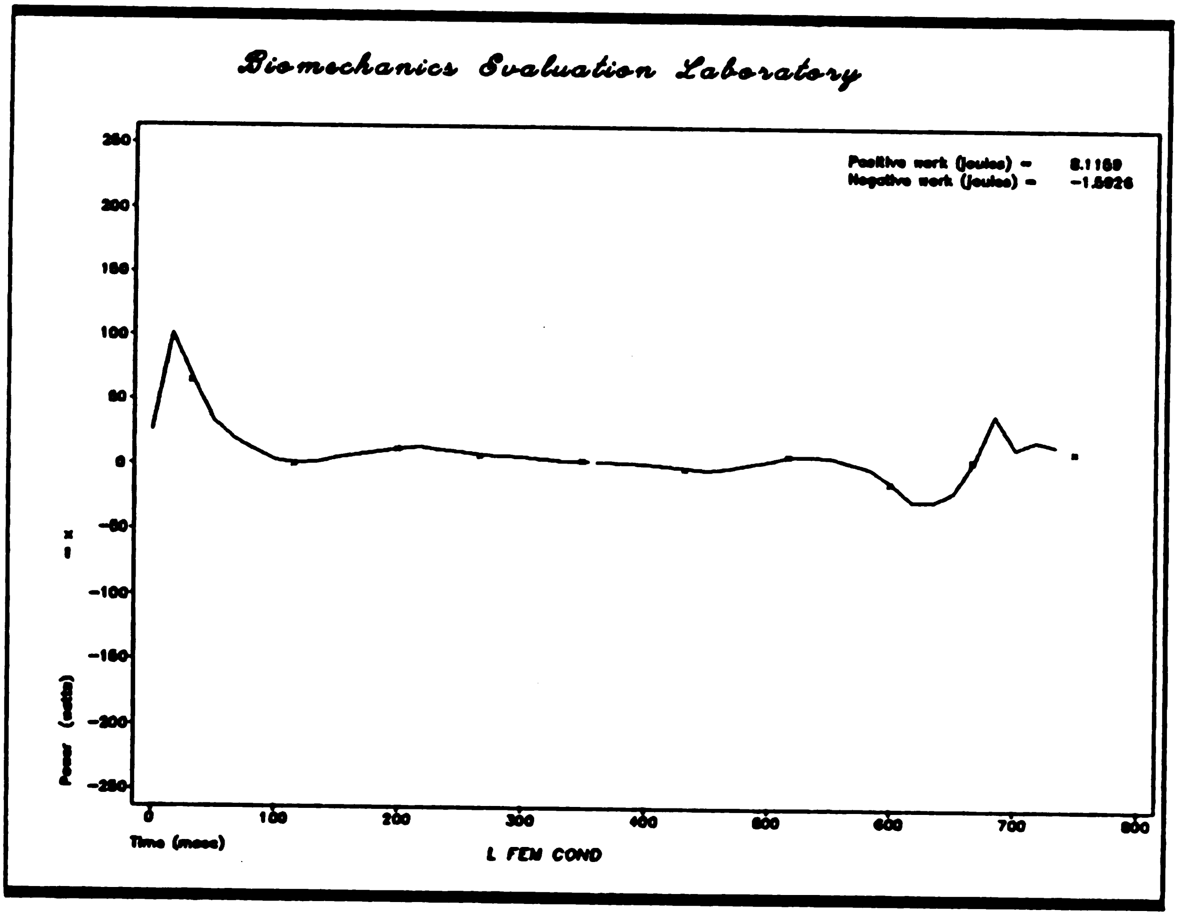

Figure 11f Subject F6

Figure 11 (cont'd.).

eccentric quadriceps activity of braking present during forward walking.

Table 3 presents the average peak positive power of three trials for each subject during the forward walking and rearward walking conditions. A paired students t- test demonstrated no significant differences in peak positive powers between conditions.

Table 3
Average peak positive power - walking
(Watts)

Subject #	Forward	Rearward	Difference
F1	118	86	32
F2	81	135	-54
F3	129	67	62
F4	53	36	17
F5	104	124	-20
F6	57	90	-33

$X_D = 0.7$ $S_D = 44$ $t = .039$

Table 4 presents the average peak negative power of three trials for each subject during the forward and rearward walking conditions. A paired students t- test demonstrated significantly ($p<.05$) greater peak negative values during the forward walking condition. The peak negative value in forward walking occurred during late stance and was primarily due to concentric activity of the hamstrings and gastrocnemius muscles.

Table 4
Average peak negative power - walking
(Watts)

Subject #	Forward	Rearward	Difference
F1	-209	-21	-188
F2	-204	-59	-145
F3	-461	-60	-401
F4	-162	-25	-137
F5	-181	-28	-153
F6	-151	-23	-128
	X_D = -192	S_D = 104	t = -4.522 p<.05

Integration of the power curve will yield the joint muscle work. Table 5 presents the average positive work of three trials from each subject during the forward and rearward walking conditions. A paired students t- test demonstrated no significant differences between conditions.

Table 5
Average positive work - walking
(Joules)

Subject #	Forward	Rearward	Difference
F1	9	10	-1
F2	11	10	1
F3	15	7	8
F4	6	6	0
F5	9	16	-7
F6	.3	13	-10
	X_D = -1.5	S_D = 6	t = -.612

Table 6 presents the average peak negative work of three trials from each subject during the forward and rearward walking conditions. A paired students t- test demonstrated significantly (p<.05) greater negative work during the forward walking condition than during the rearward walking condition.

Table 6
Average negative work - walking
(Joules)

Subject #	Forward	Rearward	Differences
F1	-30	-2	-28
F2	-27	-1	-26
F3	-43	-7	-36
F4	-21	-2	-19
F5	-24	-1	-23
F6	-14	-1	-13

$X_D = -24$ $S_D = 8$ $t = -7.348$
$p < .05$

The EMG activity from the muscles surrounding the knee joint were analyzed. The active muscles should be responsible for the knee power generation and absorption requirements. Figure 12 shows the firing pattern for each of the six muscles studied during the stance phase of forward walking from heel strike (HS) to toe off (TO). Figure 13 shows the firing pattern for each of the six muscles studied during the stance phase of rearward walking from toe strike (TS) to heel off (HO). Figures 12 and 13 show the average pattern from all six subjects. For ease of

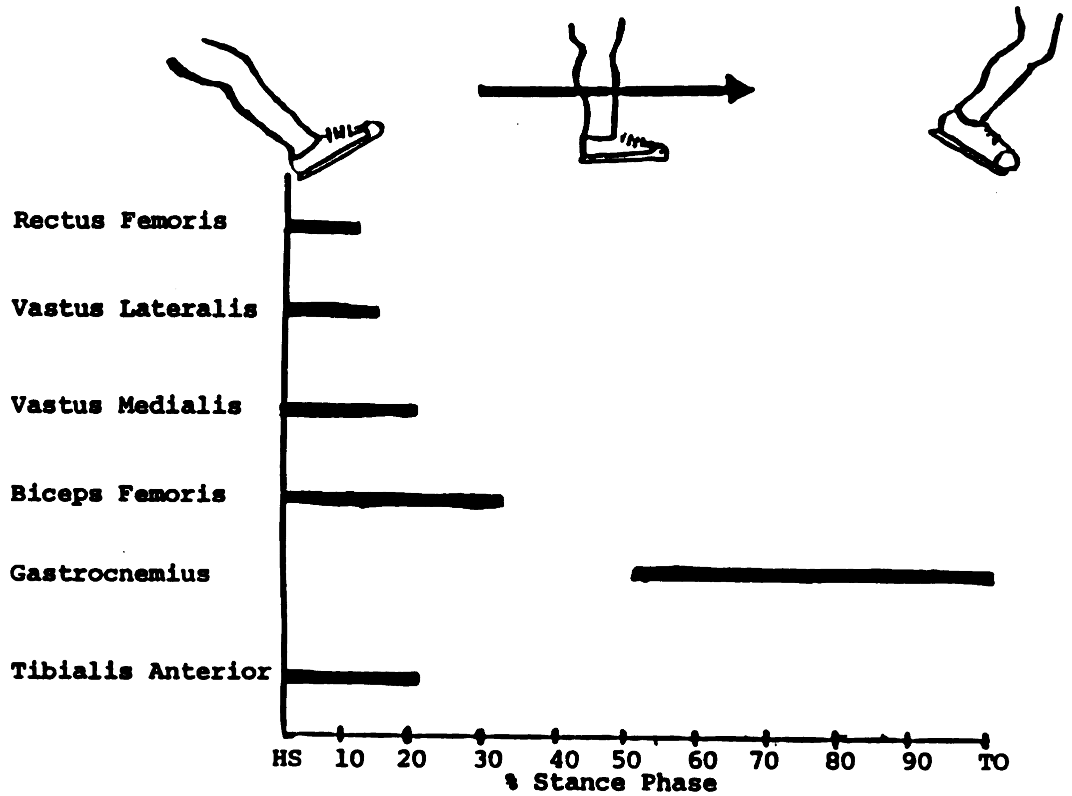

Figure 12 Forward walking EMG pattern

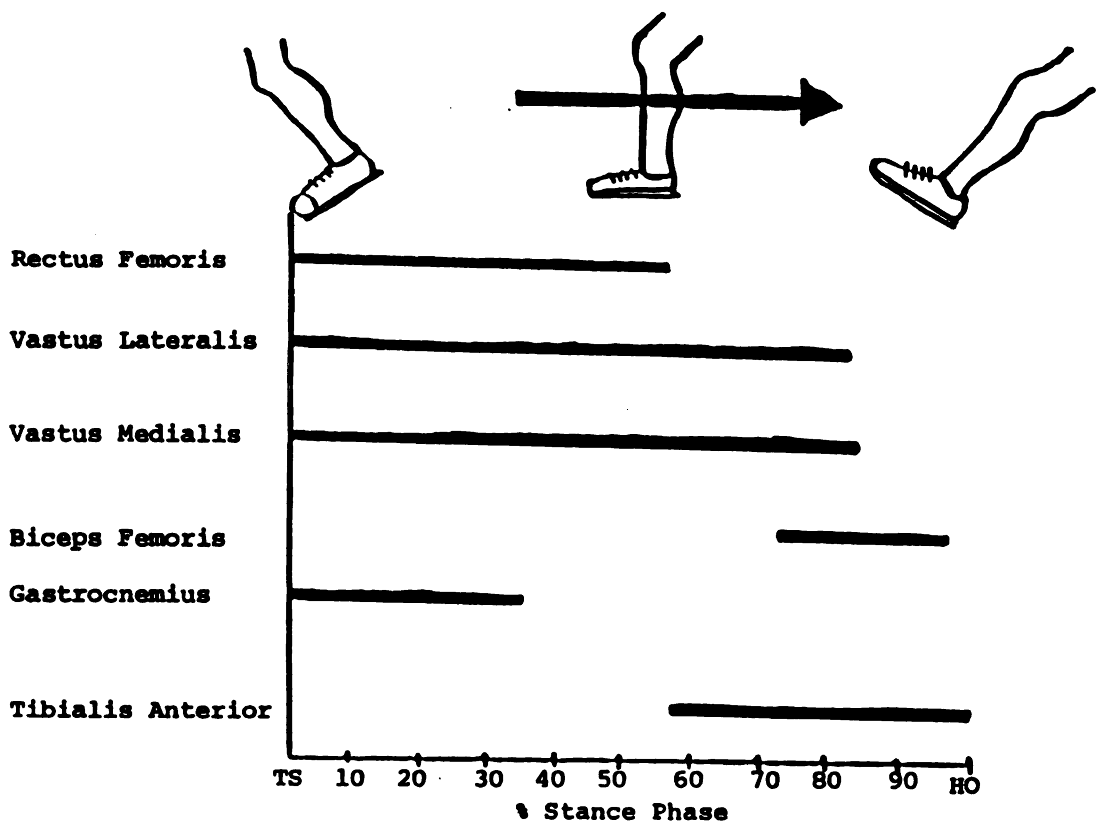

Figure 13 Rearward walking EMG pattern

viewing the direction of travel is from left to right in both figures.

Table 7 presents the mean muscle total on time and standard deviation (sd) across subjects in % of stance phase for each muscle during the forward and rearward walking conditions. A within subject paired students t-test with the appropiate level of significance is presented for each muscle.

Table 7
Total on time for individual muscles - walking
(%)

Muscle	Forward	Rearward	t-level	$p<$
Rectus femoris	16 $\pm$ 4	58 $\pm$ 23	-4.719	.01
Vastus lateralis	19 $\pm$ 12	82 $\pm$ 6	-13.498	.001
Vastus medialis	23 $\pm$ 5	83 $\pm$ 5	-25.46	.001
Biceps femoris	37 $\pm$ 10	21 $\pm$ 7	3.989	.02
Gastrocnemius	48 $\pm$ 18	35 $\pm$ 18	1.027	-
Tibialis anterior	22 $\pm$ 15	43 $\pm$ 25	-2.784	.05

The EMG muscle firing sequence presented in Figure 12 is consistent with the literature (5). A comparison of the forward walking EMG signal with the rearward walking signal demonstrated a marked increase in total on time of the three knee extensor muscles (rectus femoris, vastus lateralis, and vastus medialis) during the rearward walking condition compared to the forward walking condition. Comparing the EMG activity to the muscle moment and power curves suggested primarily concentric propulsion activity of the knee extensors in rearward walking with periods of isometric stabilizing activity through the first 80% of stance phase. This contrasts sharply with the forward walking conditions where the knee extensors are acting primarily as eccentric shock absorbers during early stance phase.

The biceps femoris showed greater inconsistent activity both within and across subjects. The biceps femoris muscle activity demonstrated a statistically significant increase in total on time during forward walking as compared to rearward walking. In forward walking the biceps femoris was active during early stance cocontracting with the knee extensors. In rearward walking the phase of biceps femoris activity that was consistent across subjects was during late stance phase.

No statistically significant differences were noted in gastrocnemius total on time between conditions, but a marked change in the type of muscular contraction and function was noted. In forward walking, the gastrocnemius acted primarily concentrically from mid to late stance assisting with propulsion of the limb forward. This contrasts with rearward walking where the gastrocnemius appeared to be acting eccentrically during the first portion of stance. This would allow the gastrocnemius to function primarily as a shock absorber via the ankle joint during weight acceptance in rearward walking.

The tibialis anterior demonstrated a statistically significant increase in total on time during rearward walking. In forward walking the tibialis anterior functioned eccentrically in early stance to allow a controlled lowering of the foot to the floor. In rearward walking the tibialis anterior appears to act concentrically from mid to late stance functioning to raise the foot from the floor.

The resultant ground reaction torque is presented in Figure 14. This is a representative sample from subject F2 comparing forward walking (red) to rearward walking (green). A positive value coincides with the limb externally (laterally) rotating relative to the force plate, conversely negative values coincides with the limb internally (medial) rotating relative to the force plate. The forward walking torque was consistent with the literature (30,38), demonstrating a sinusoidal like curve begining as an internally directed torque at heel strike until approximately 50% of stance then becoming externally directed for the remainder of stance. The rearward walking torque is externally directed until approximately 70% of stance then a short small amplitude internally rotated torque which returns to an externally directed torque in the final 10% of stance.

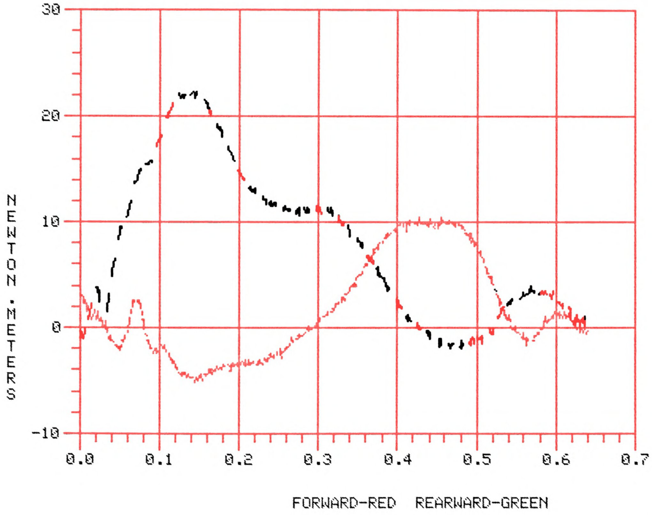

Figure 14

Ground reaction torque forward vs rearward walking

RUNNING

The knee muscle moment curves from the six subjects running forward are presented in Figure 15. Each graph shows the three trials from each subject for the forward running condition. An extension muscle moment of the knee is positive (+) and a flexion muscle moment is negative (-).

The shape of the forward running knee muscle moment curves compare favorably with Winter's(54). The knee muscle moment was entirely extensor through the stance phase of running. The shape of the moment curve was consistent across subjects, but the magnitudes varied. The large magnitude of the knee extensor moment in subject F3 possibly represented a mechanical inefficiency in this subject, since he was the only subject that had not previously jogged with any regularity.

The knee muscle moment curves for the six subjects running rearward are presented in Figure 16. Each graph shows three trials from each subject in the rearward running condition. An extension muscle moment is positive (+) and a flexion muscle moment is negative (-).

The rearward running knee muscle moment patterns vary considerably with their forward running counterparts. The rearward running knee muscle moment consistently displayed a two stage pattern beginning with a brief extensor period from toe strike until 25% of stance, followed by a second larger extensor moment from 50-100% of stance. The

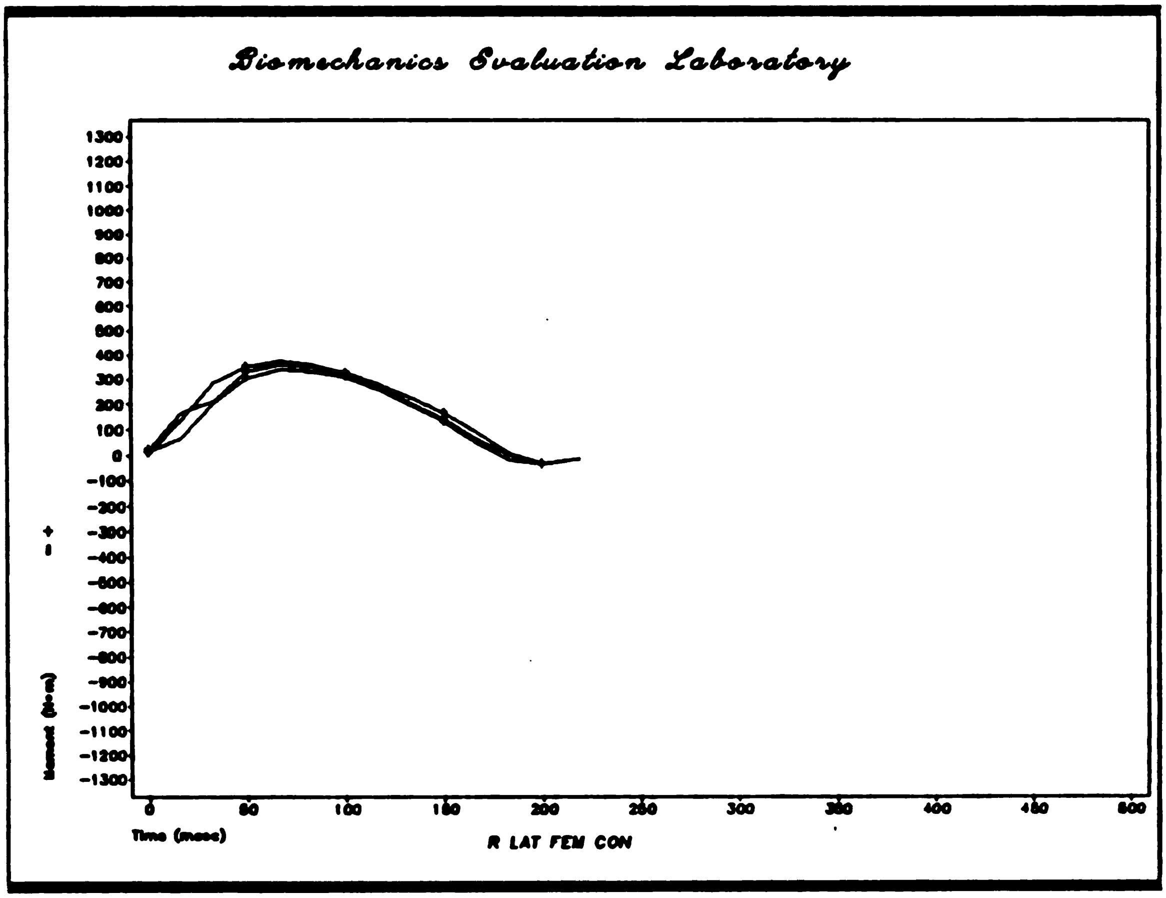

Figure 15a Subject F1

Figure 15 Forward Running Knee Muscle Moment Curves

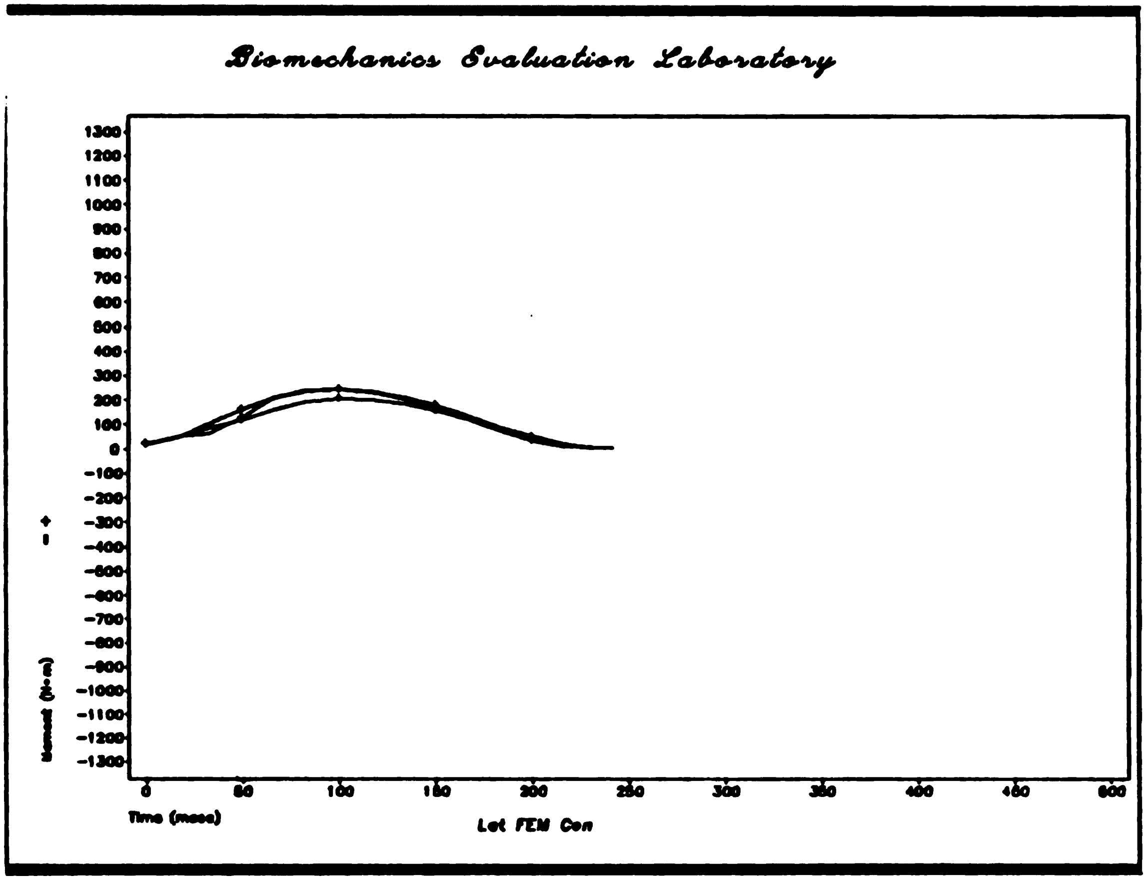

Figure 15b Subject F2

Figure 15 (cont'd.).

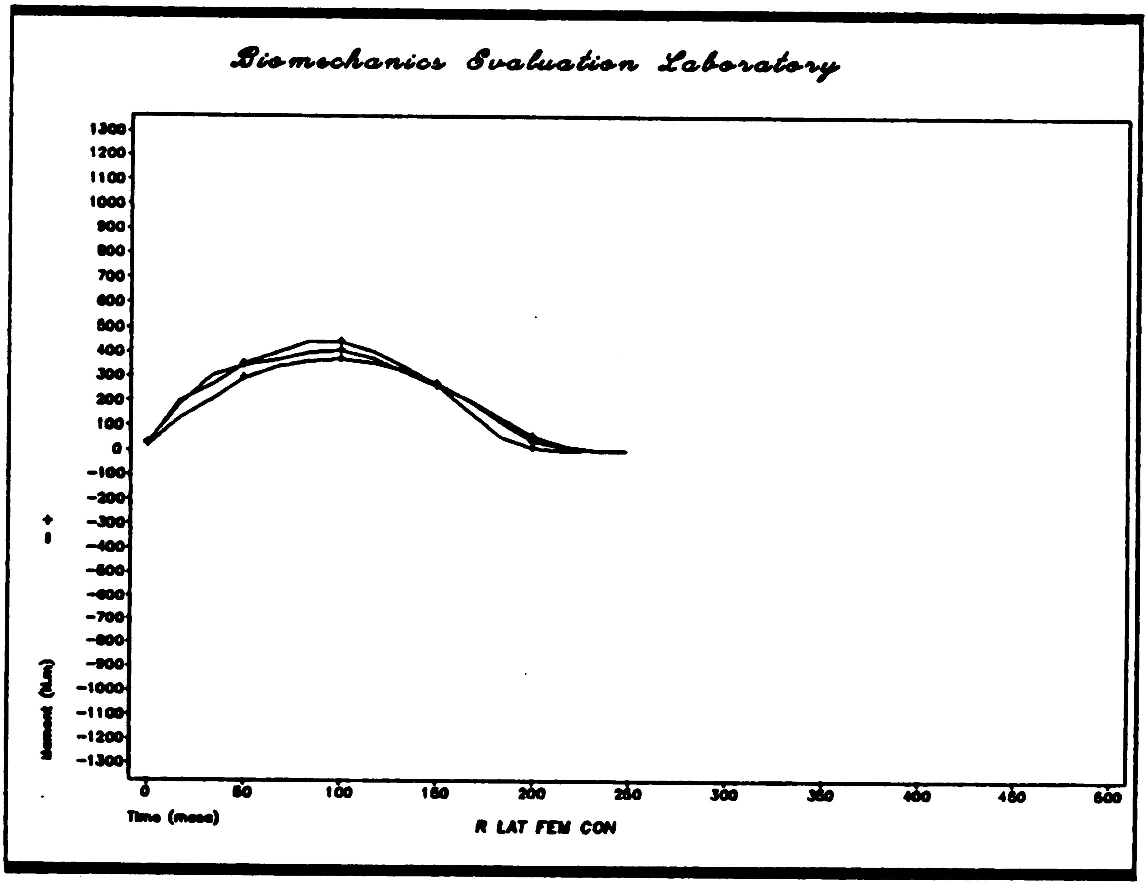

Figure 15c Subject F3

Figure 15 (cont'd.).

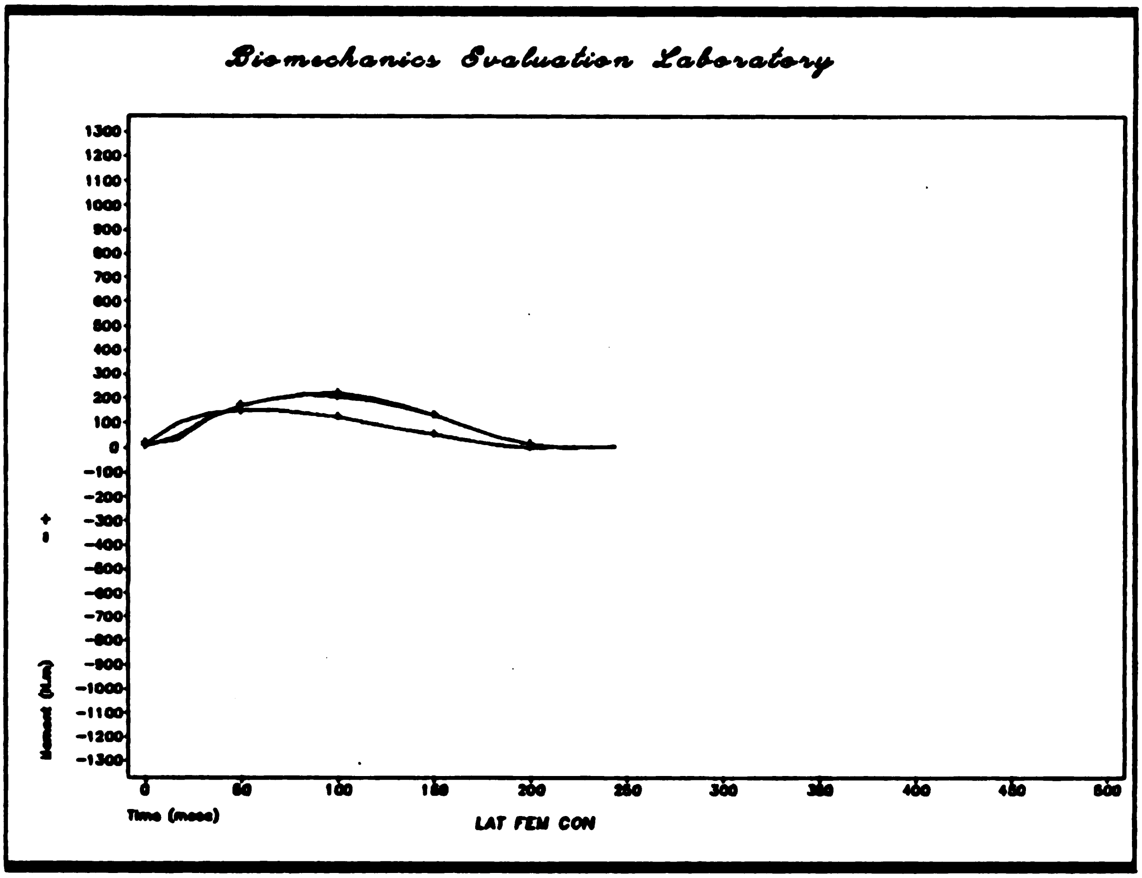

Figure 15d Subject F4

Figure 15 (cont'd.).

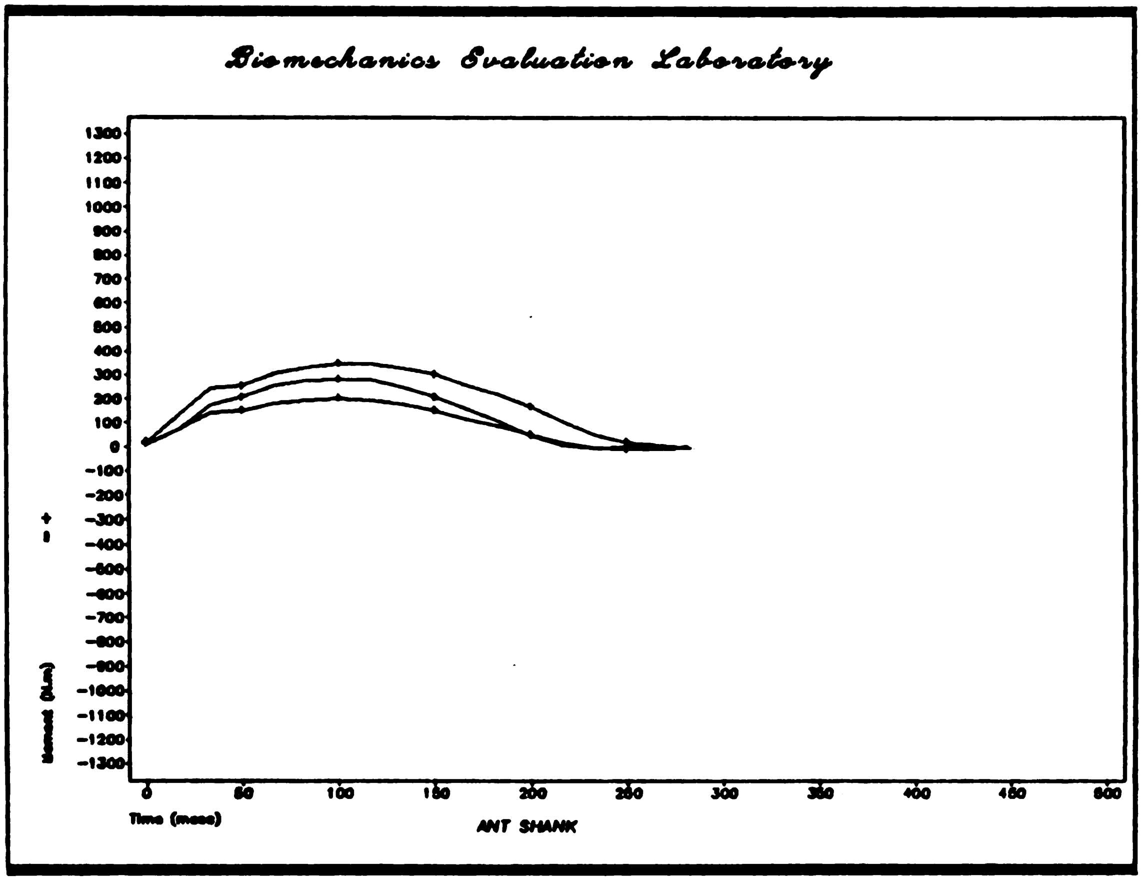

Figure 15e Subject F5

Figure 15 (cont'd.).

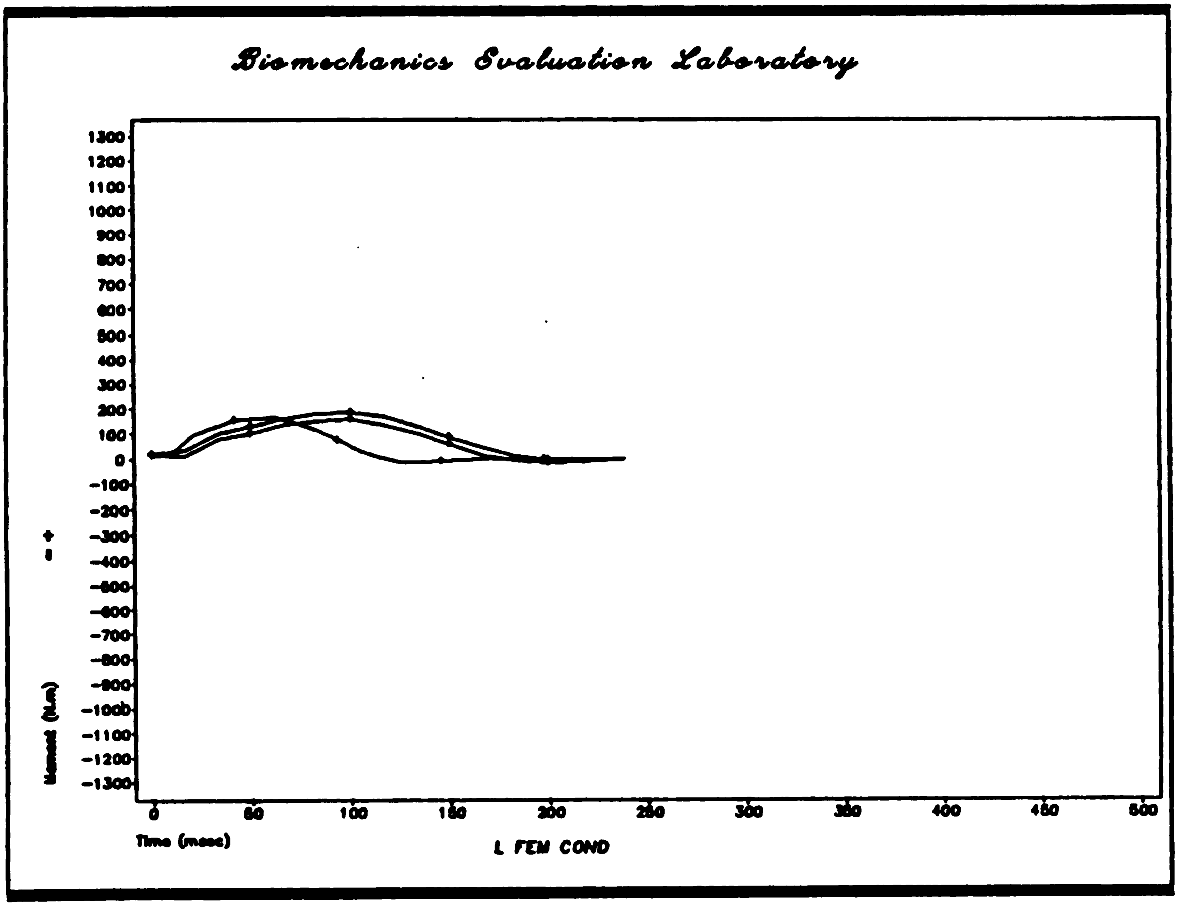

Figure 15f Subject F6

Figure 15 (cont'd.).

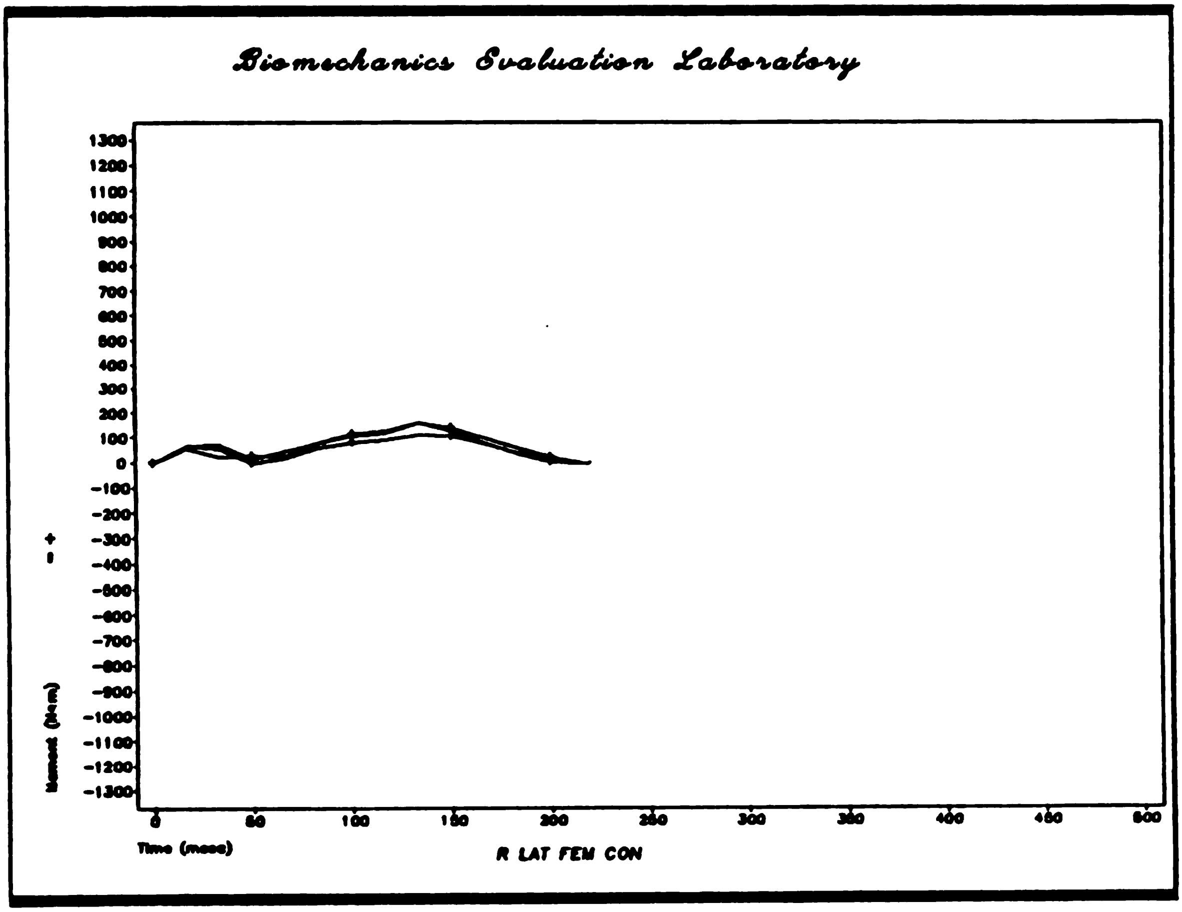

Figure 16a Subject F1

Figure 16 Rearward Running Knee Muscle Moment Curves

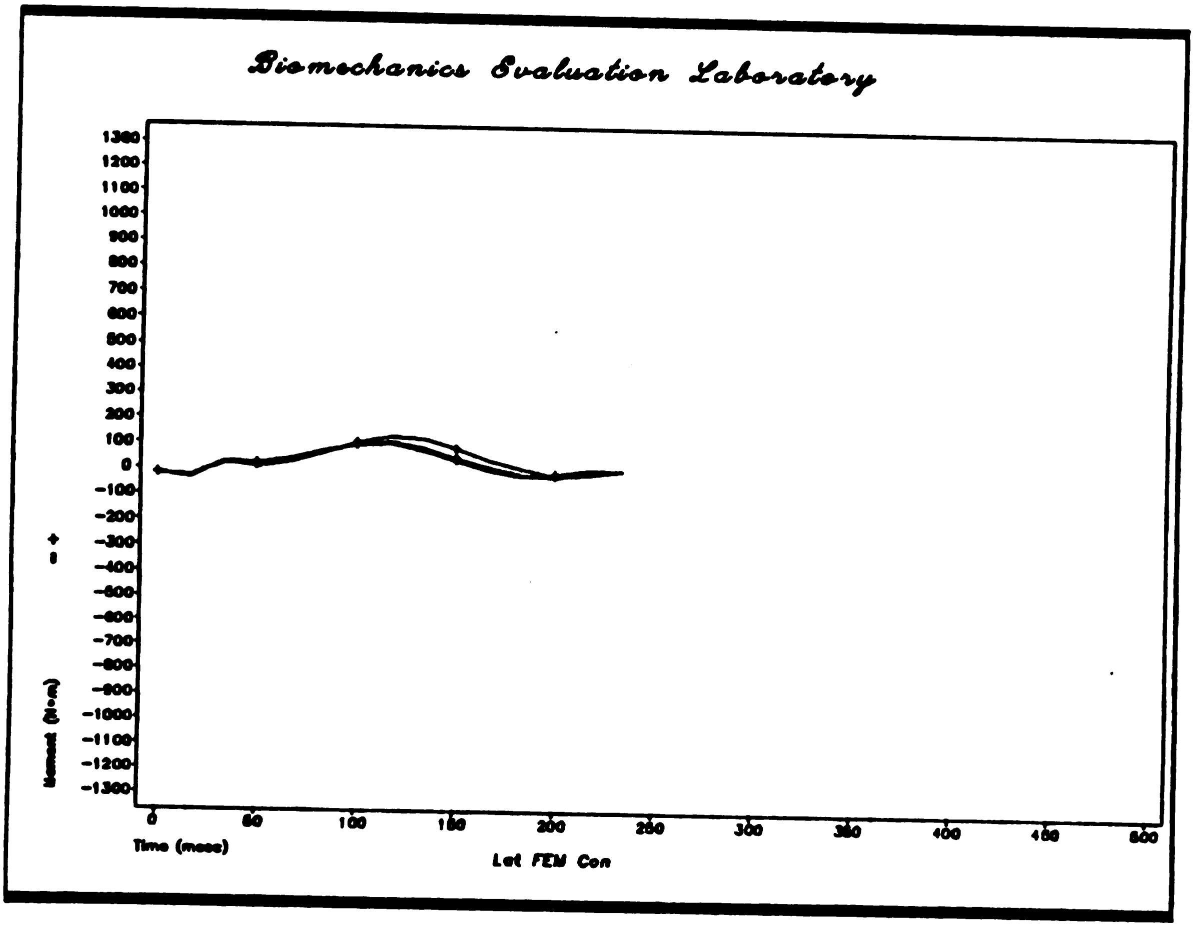

Figure 16b Subject F2

Figure 16 (cont'd.).

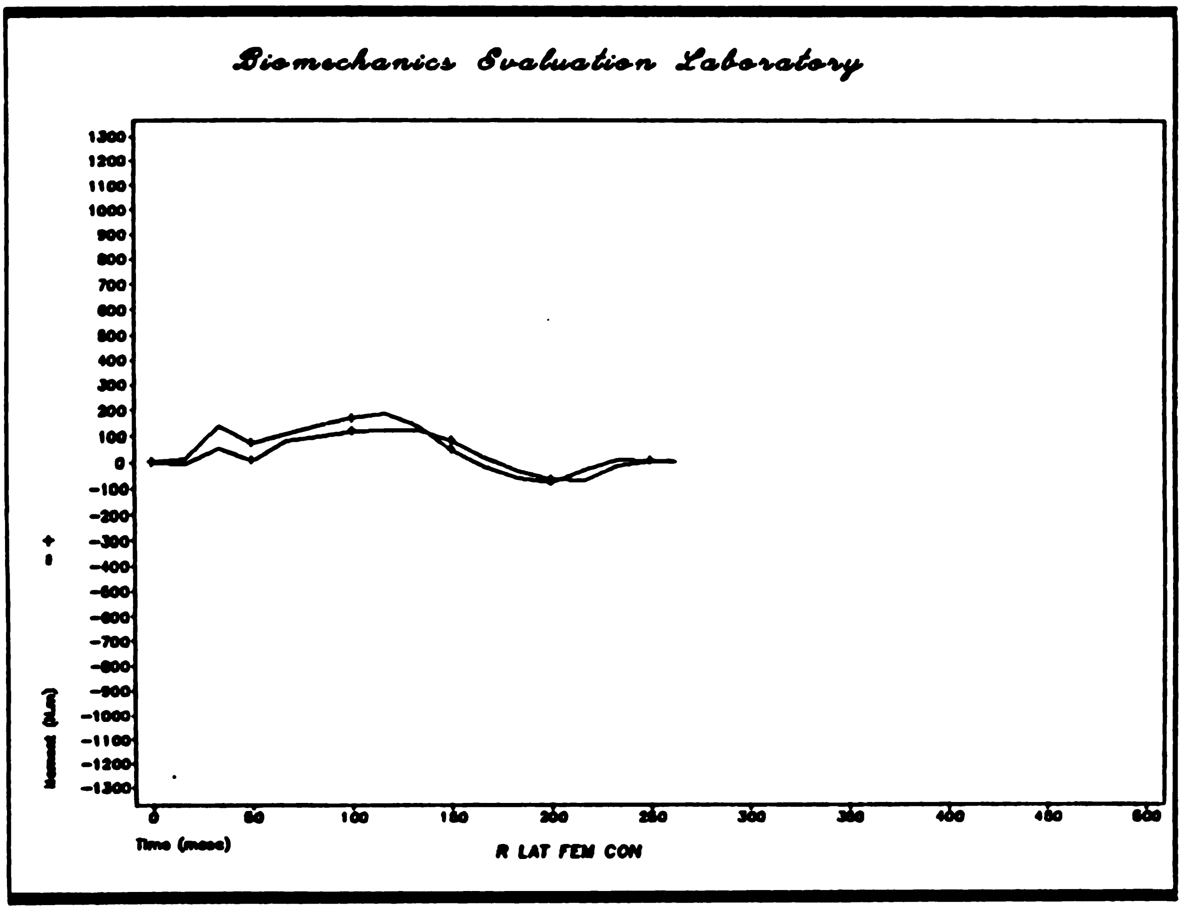

Figure 16c Subject F3

Figure 16 (cont'd.).

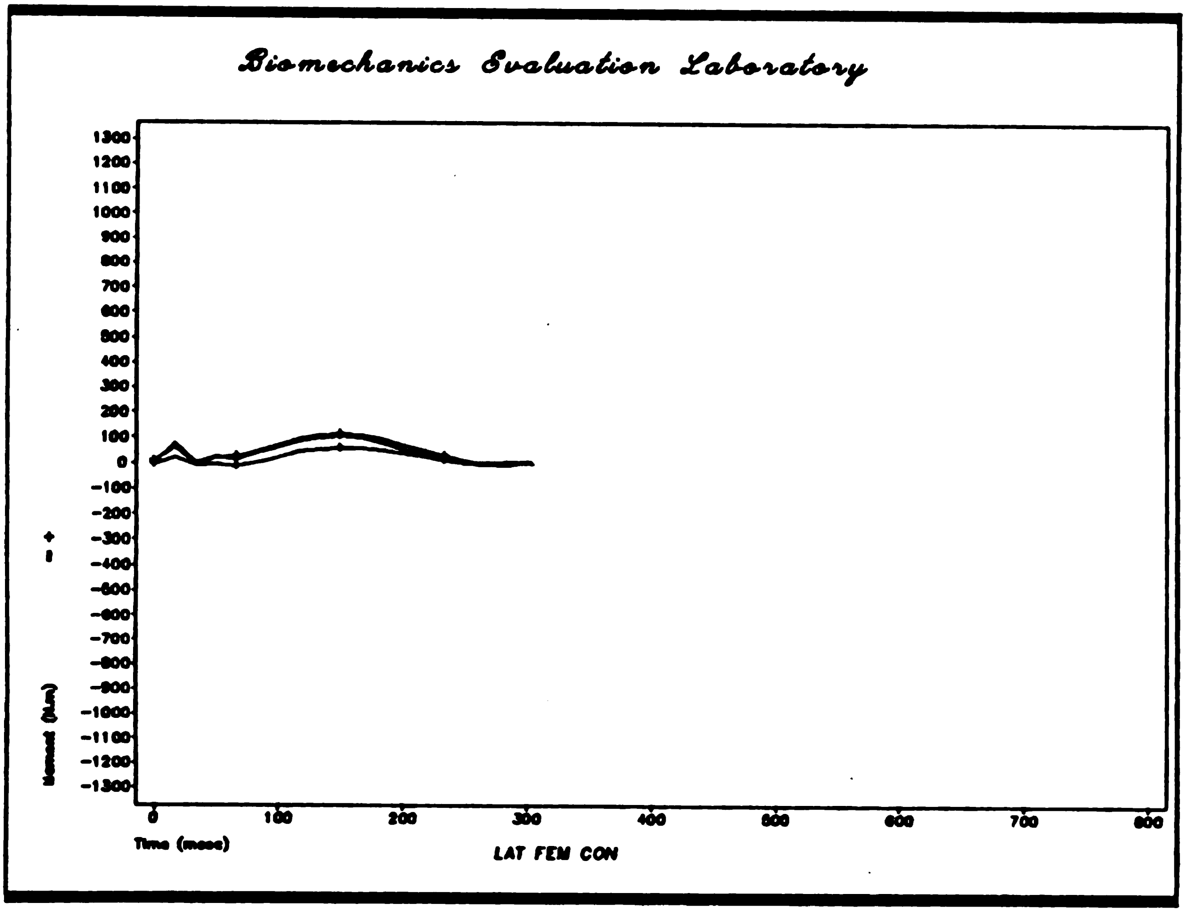

Figure 16d Subject F4

Figure 16 (cont'd.).

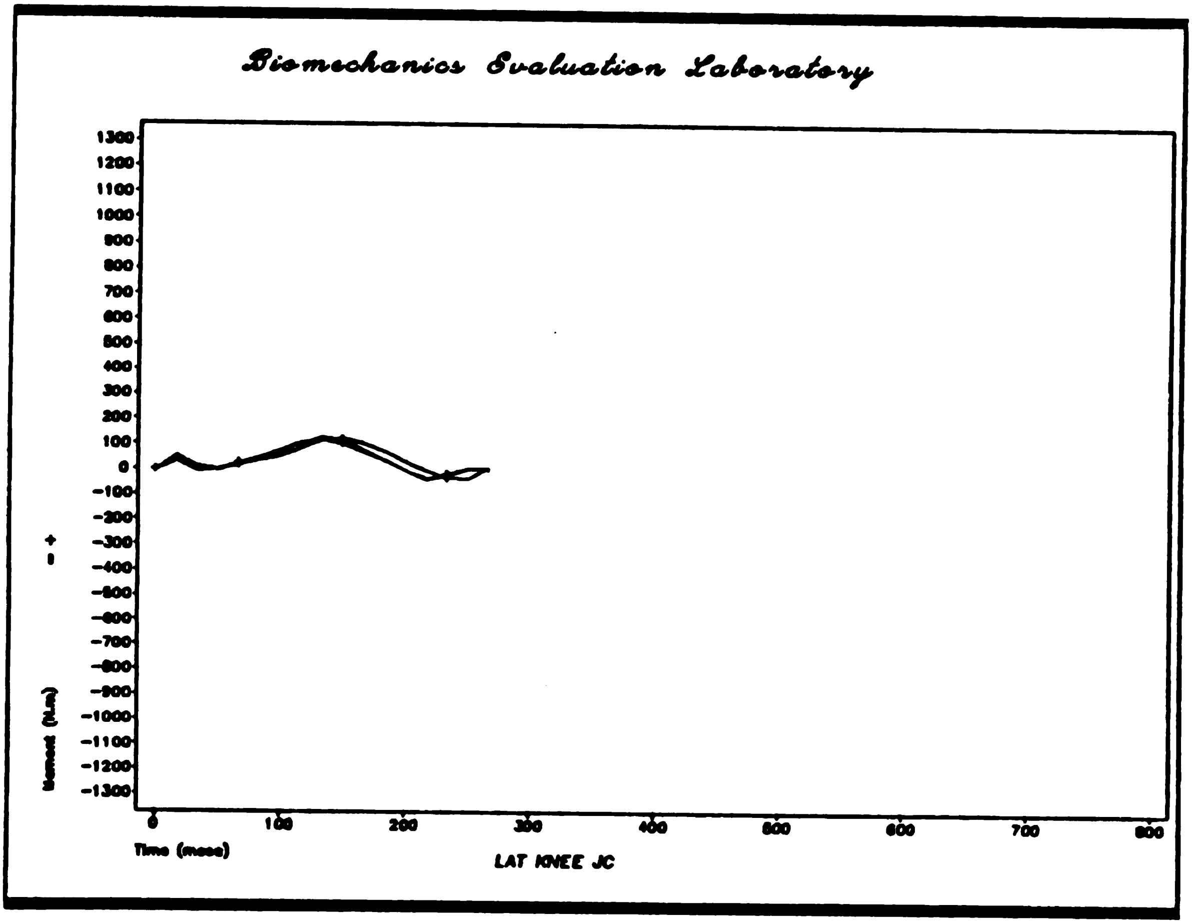

Figure 16e Subject F5

Figure 16 (cont'd.).

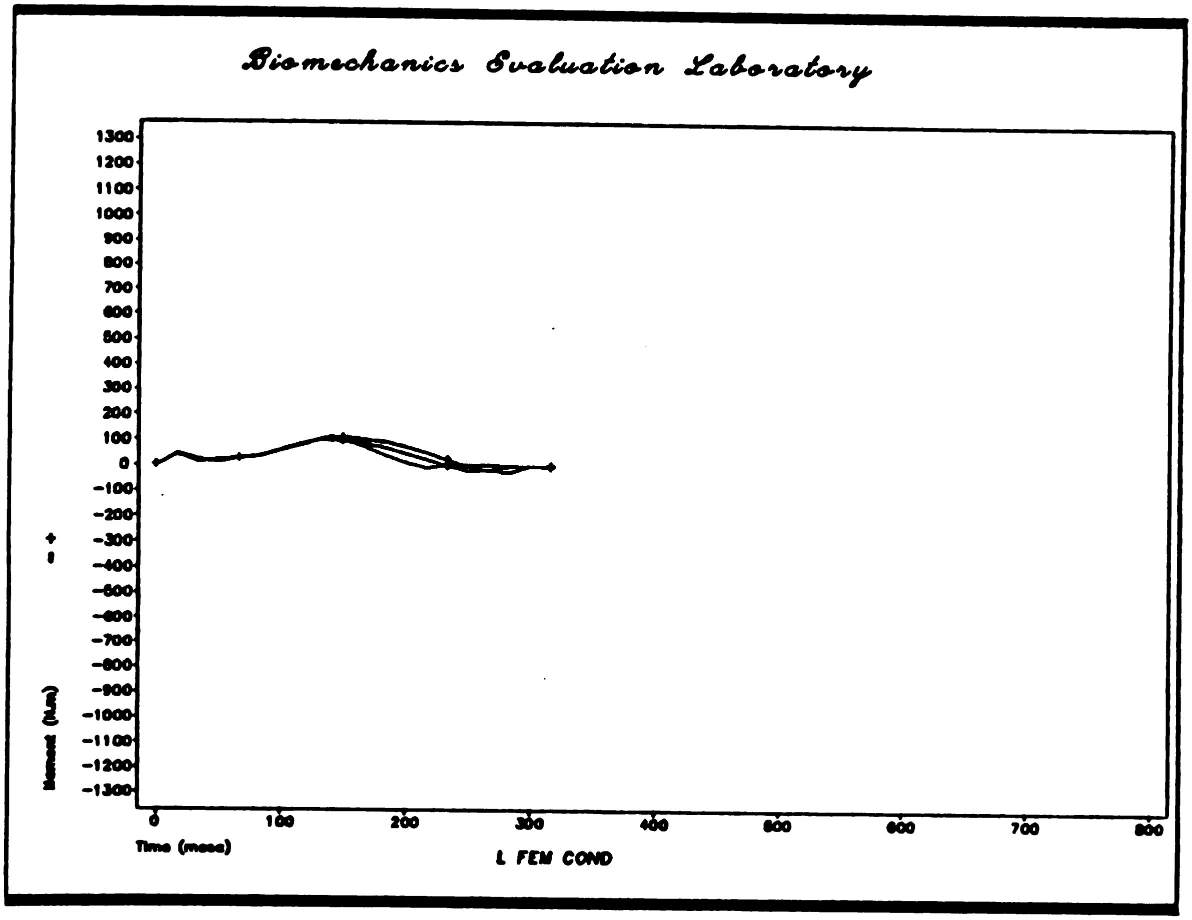

Figure 16f Subject F6

Figure 16 (cont'd.).

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magnitude of the knee muscle moments in rearward running were consistently lower than those noted in forward running.

The representative sample of the knee muscle power curves from each of the six subjects running forward are presented in Figure 17. A positive value indicates concentric knee extensor muscle activity or eccentric knee flexor muscle activity. Conversely, a negative value indicates eccentric knee extensor muscle activity or concentric knee flexor muscle activity.

The shape of the forward running power curves compare favorably with Winter's(54) results. During stance phase there were two distinct phases. The first power phase began immediately after heel strike and continues until mid stance. This phase was a shock absorbing phase controlled by eccentric activity of the knee extensors. The second power phase, acted to propel the limb forward from mid stance until push off, and was a function of the concentric activity of the knee extensors.

The representative sample of the knee muscle power curves from each of the six subjects running rearward are presented in Figure 18. A positive value indicates concentric knee extensor activity or eccentric knee flexor muscle activity. Conversely, a negative value indicates eccentric knee extensor muscle activity or concentric knee flexor muscle activity.

The rearward running power curves demonstrated a four phase pattern. The first phase was a small amplitude

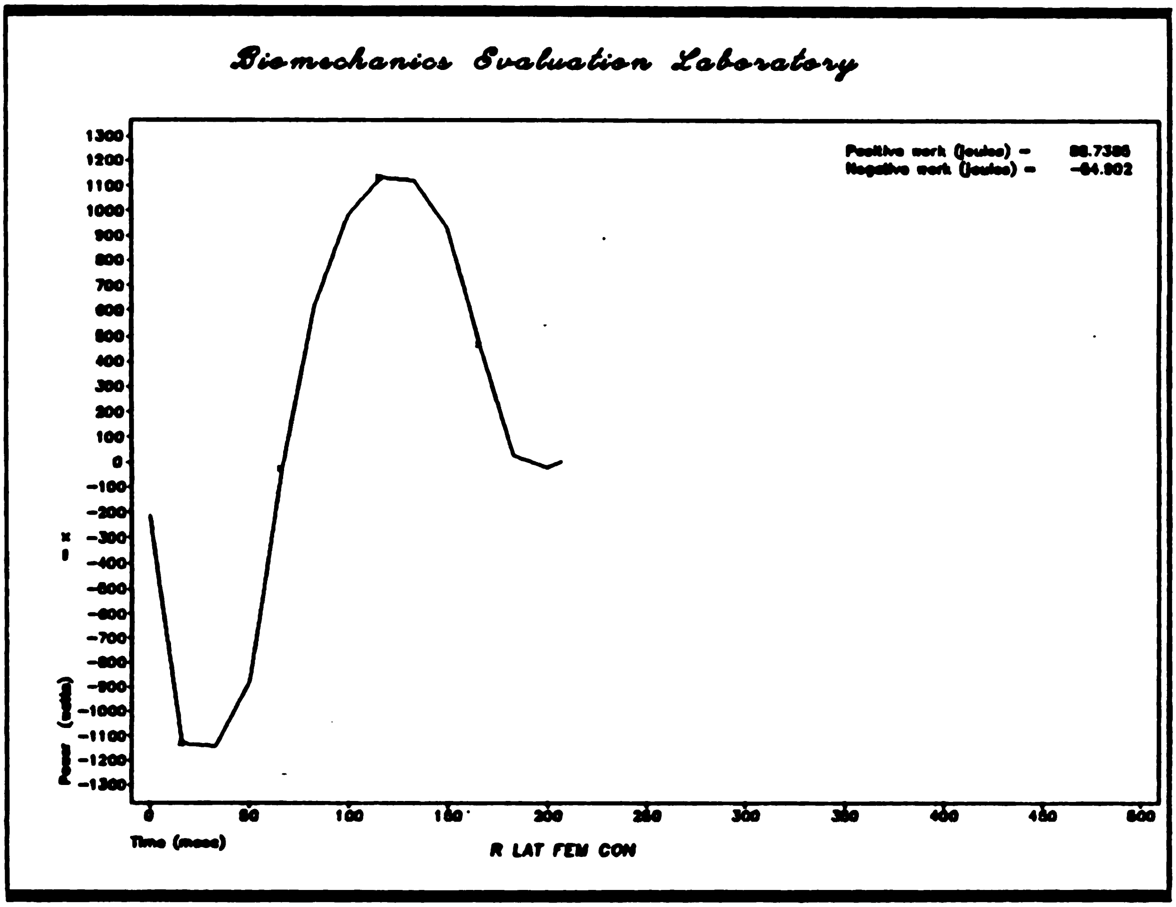

Figure 17a Subject F1

Figure 17 Forward Running Muscle Power Curves

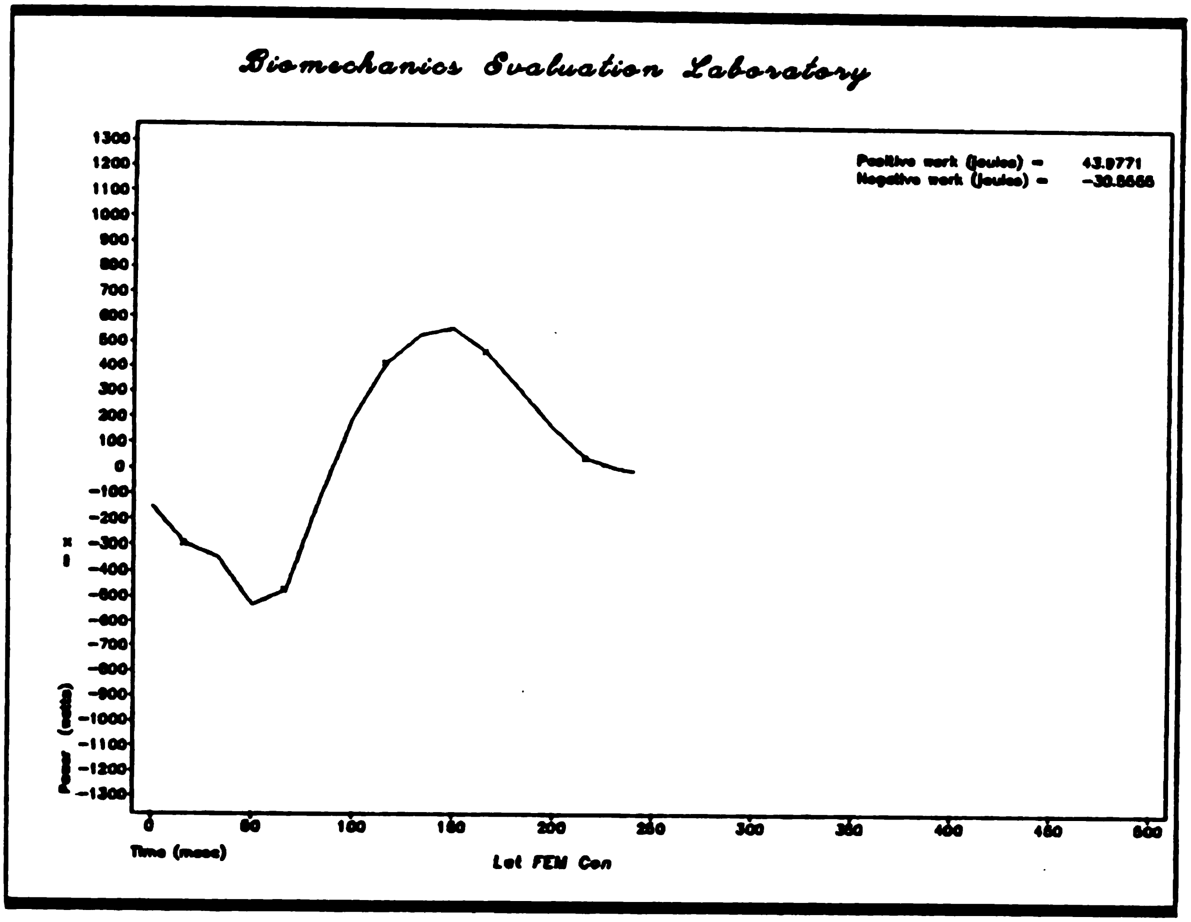

Figure 17b Subject F2

Figure 17 (cont'd.).

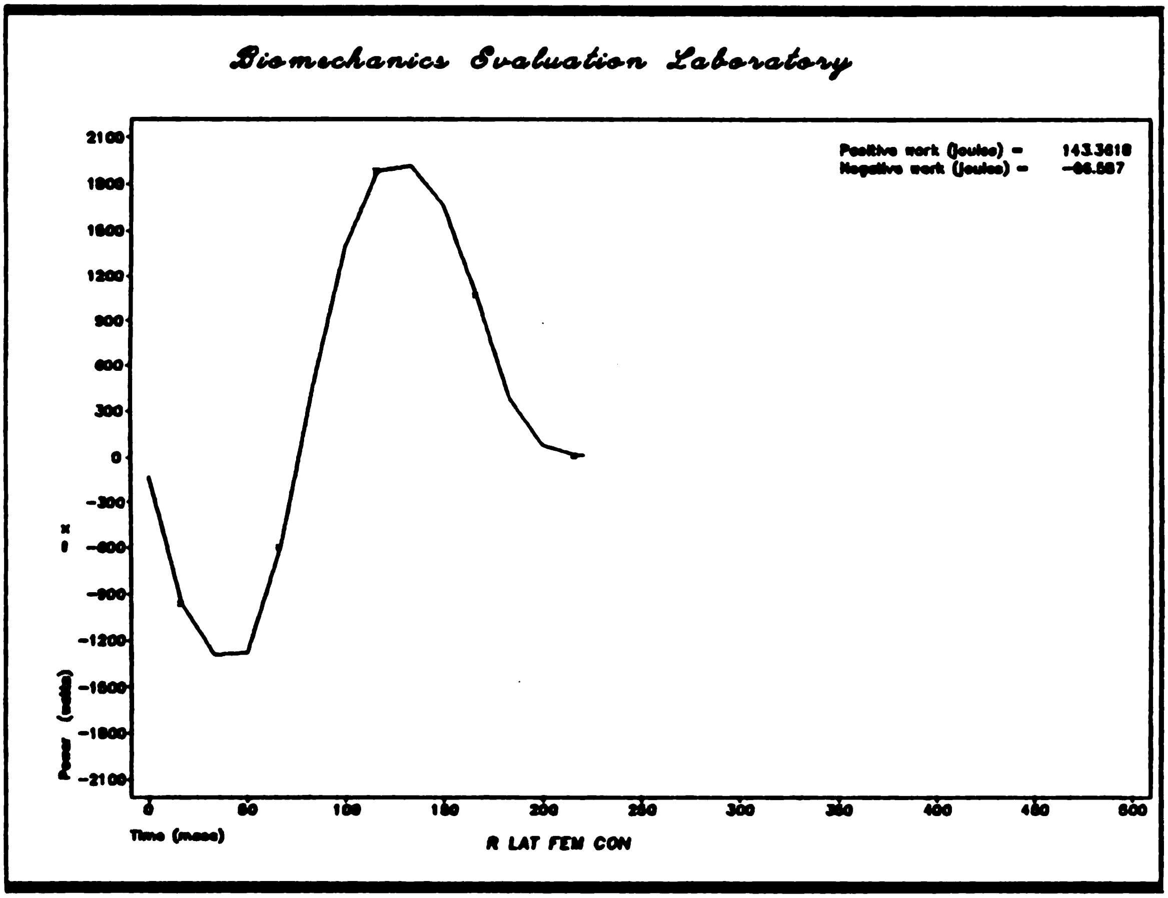

Figure 17c Subject F3

Figure 17 (cont'd.).

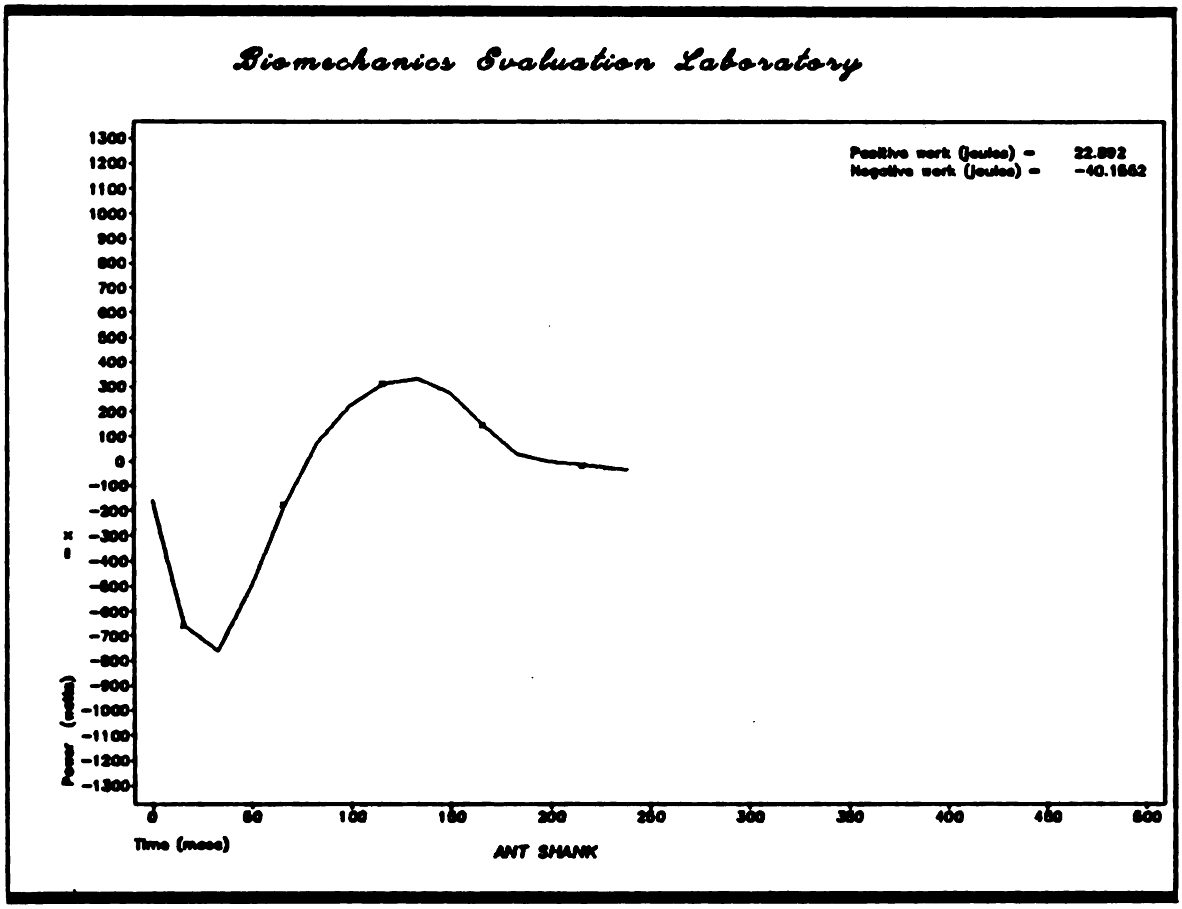

Figure 17d Subject F4

Figure 17 (cont'd.).

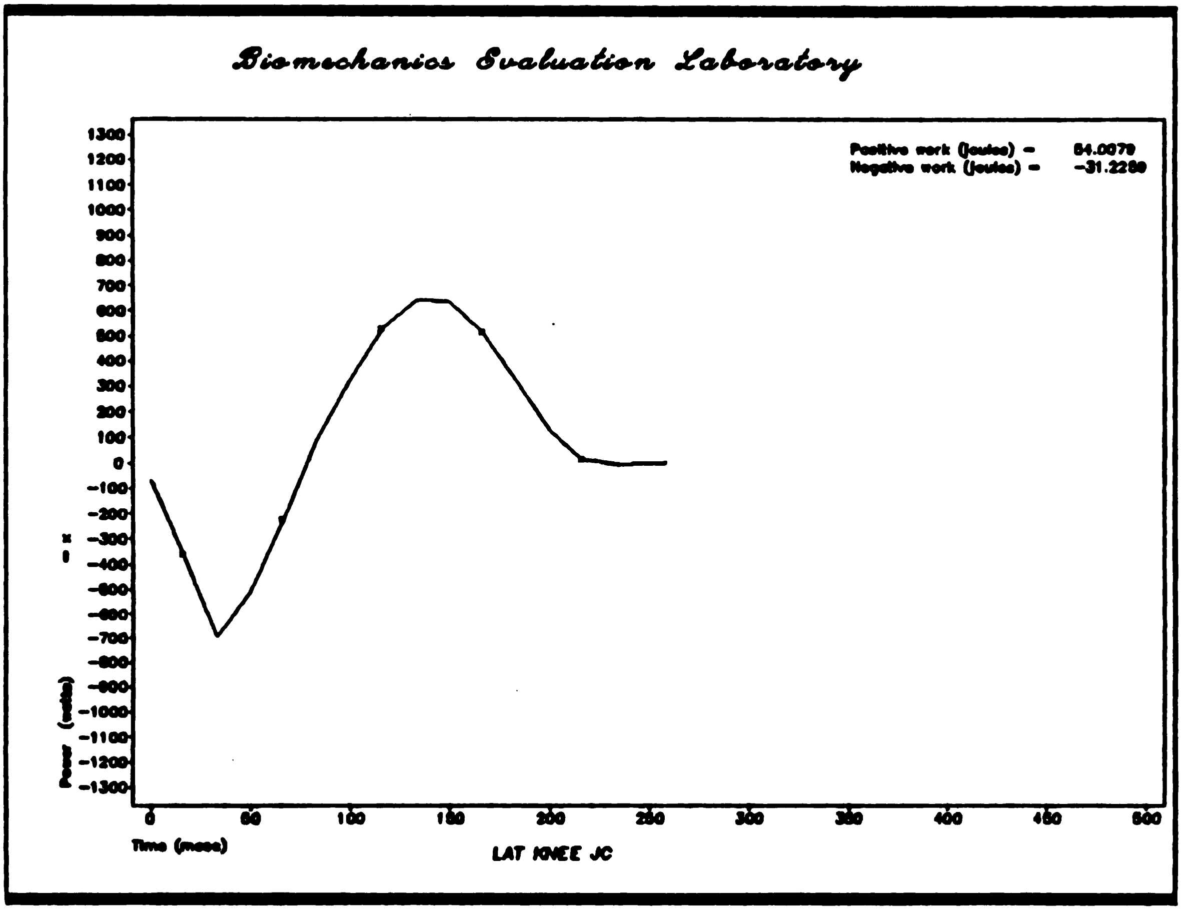

Figure 17e Subject F5

Figure 17 (cont'd.).

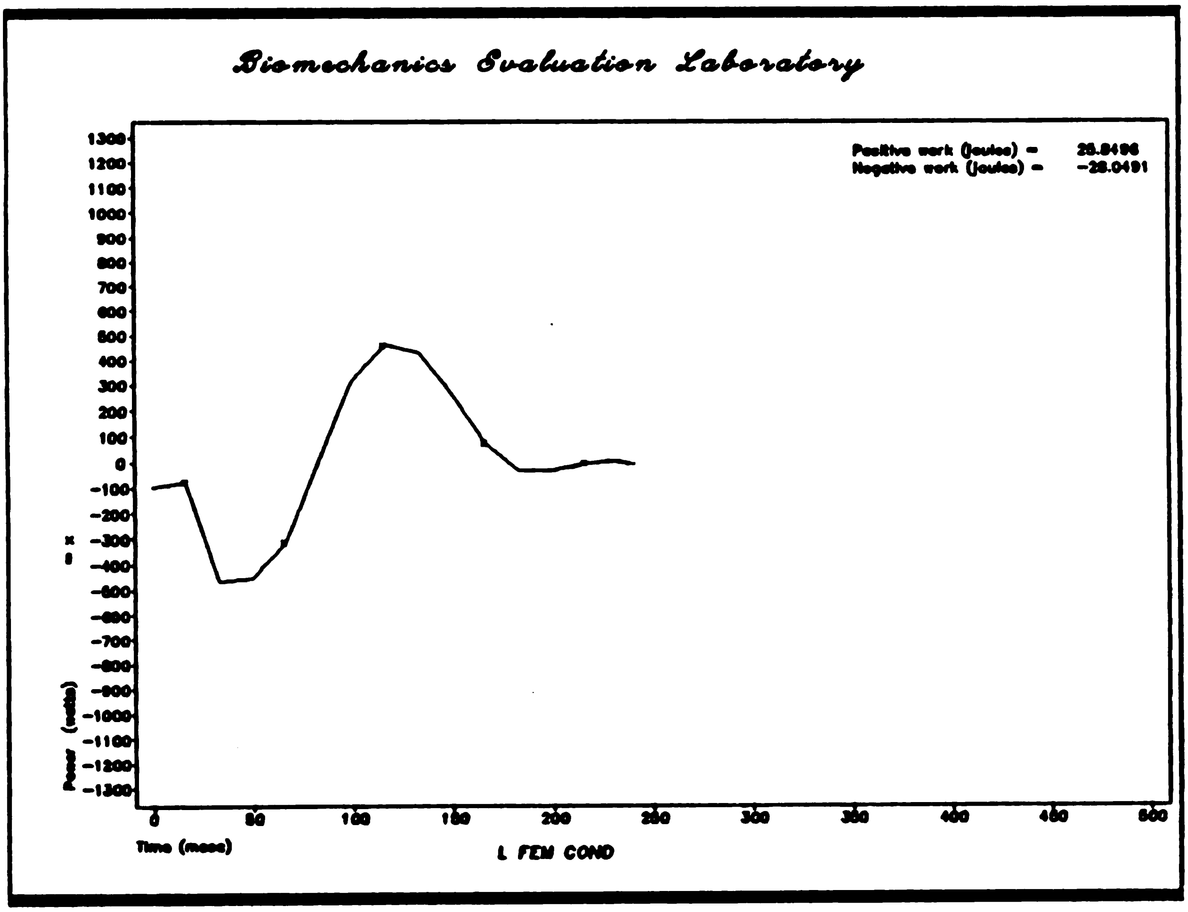

Figure 17f Subject F6

Figure 17 (cont'd.).

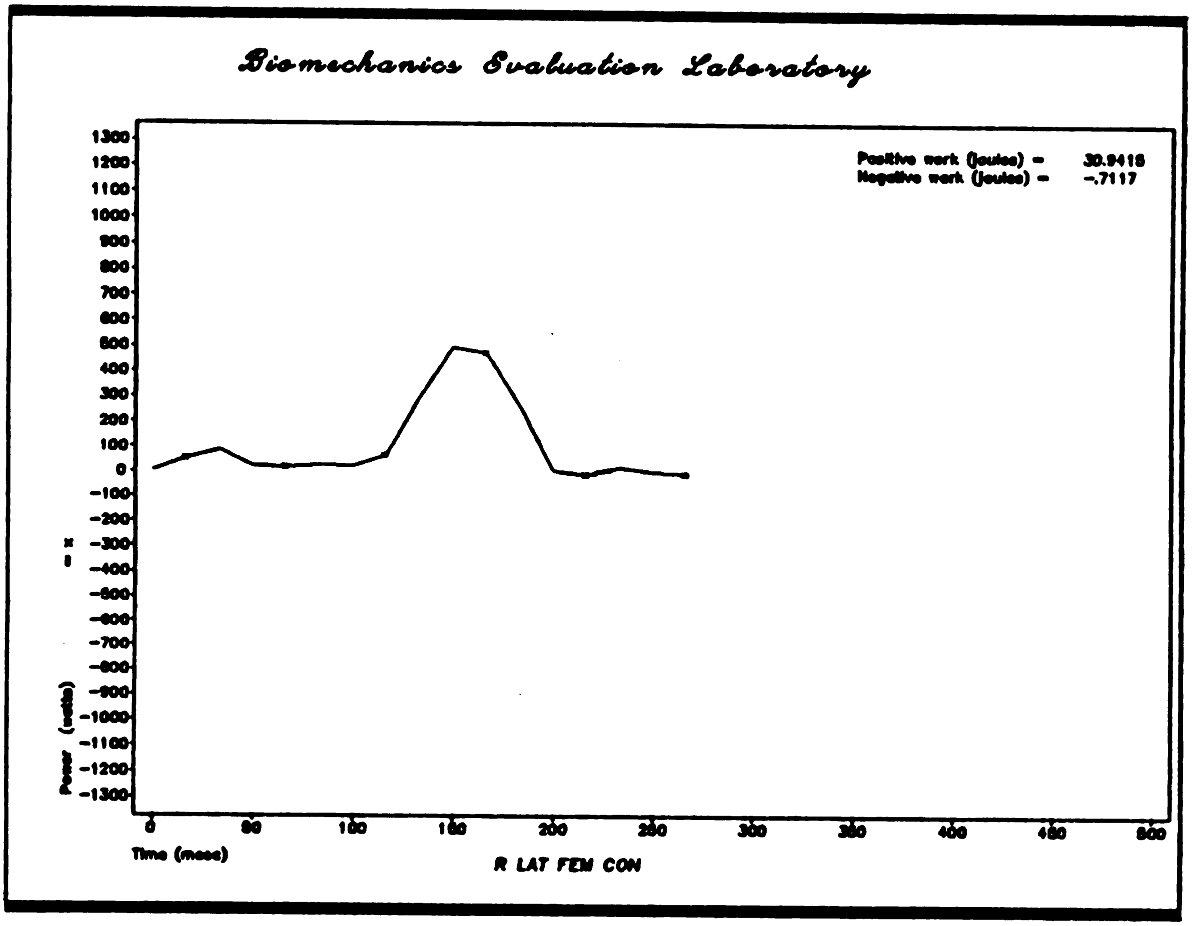

Figure 18a Subject F1

Figure 18 Rearward Running Muscle Power Curves

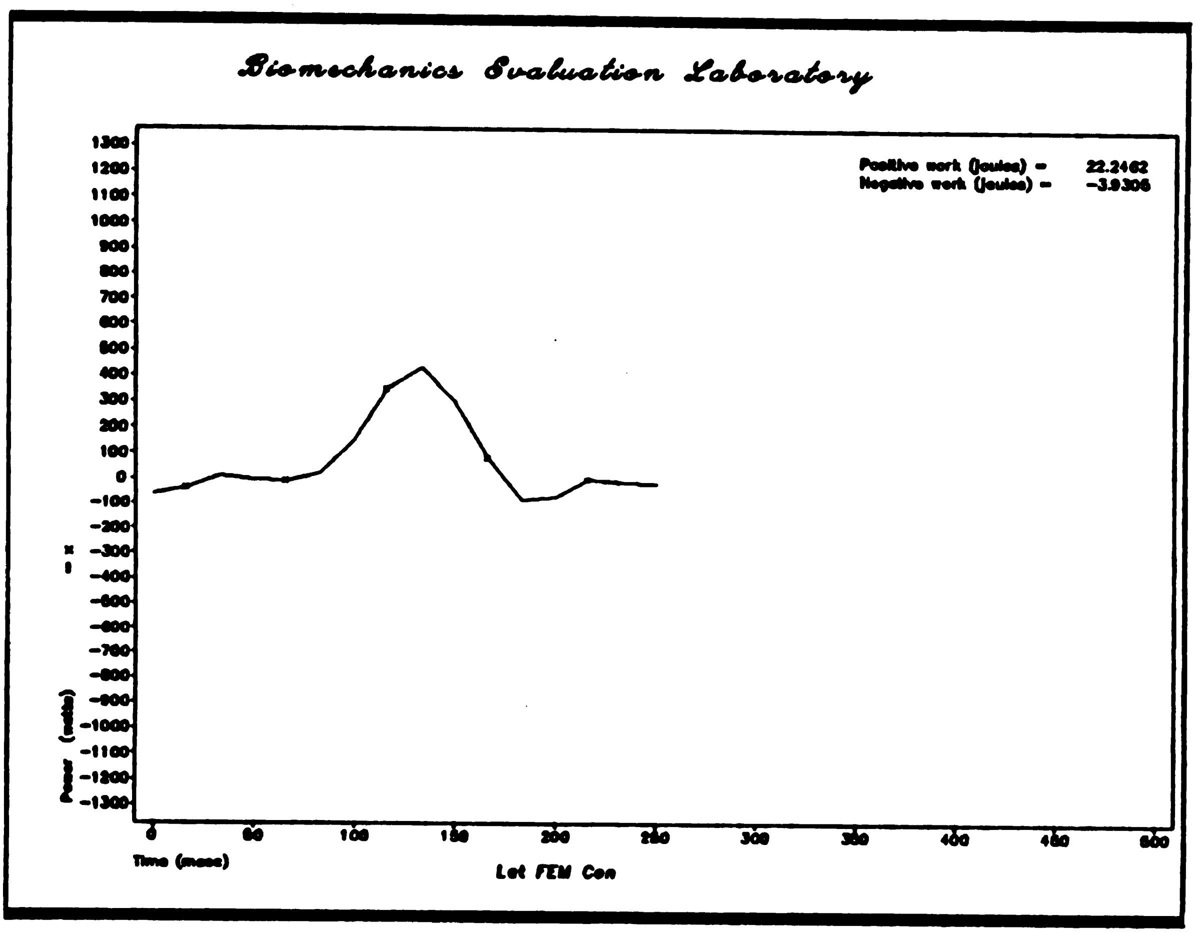

Figure 18b Subject F2

Figure 18 (cont'd.).

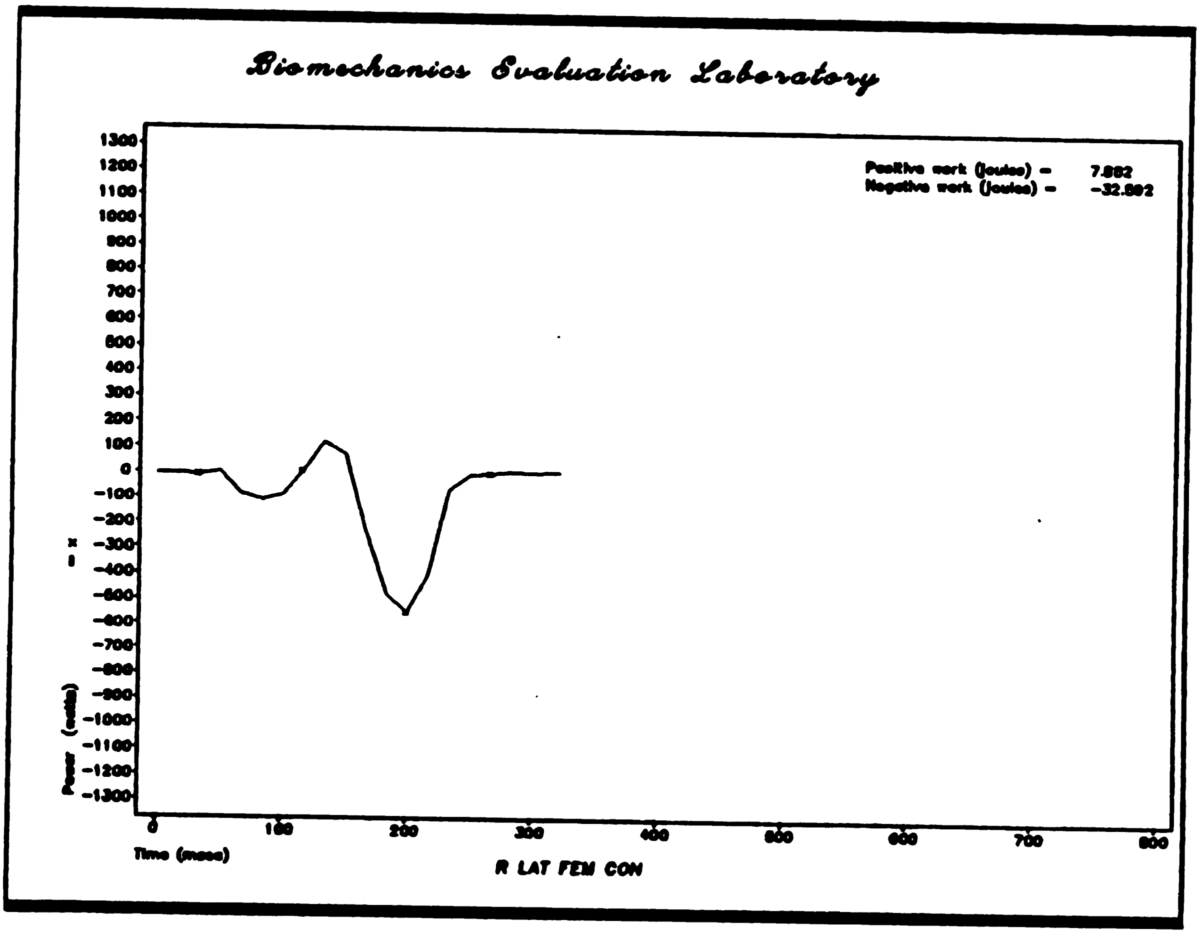

Figure 18c Subject F3

Figure 18 (cont'd.).

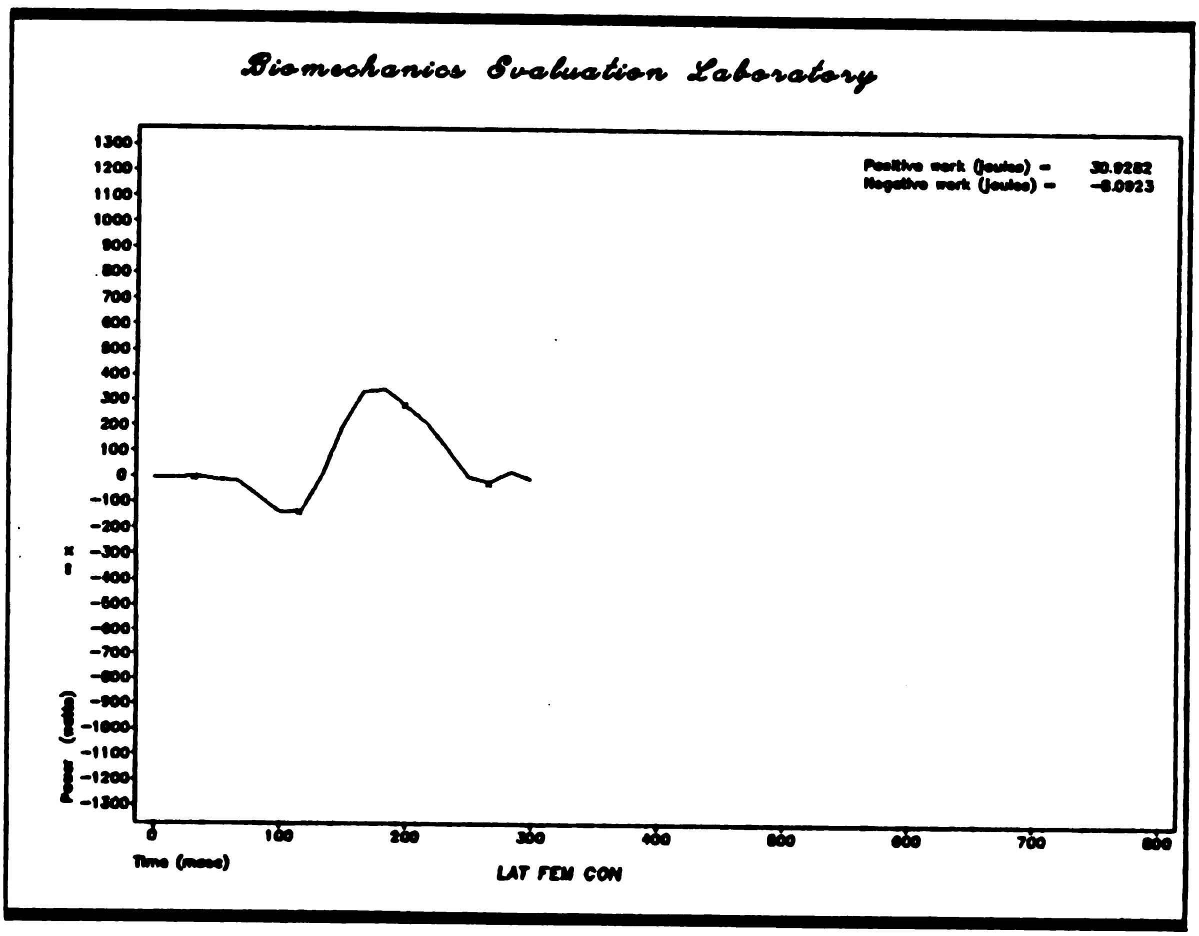

Figure 18d Subject F4

Figure 18 (cont'd.).

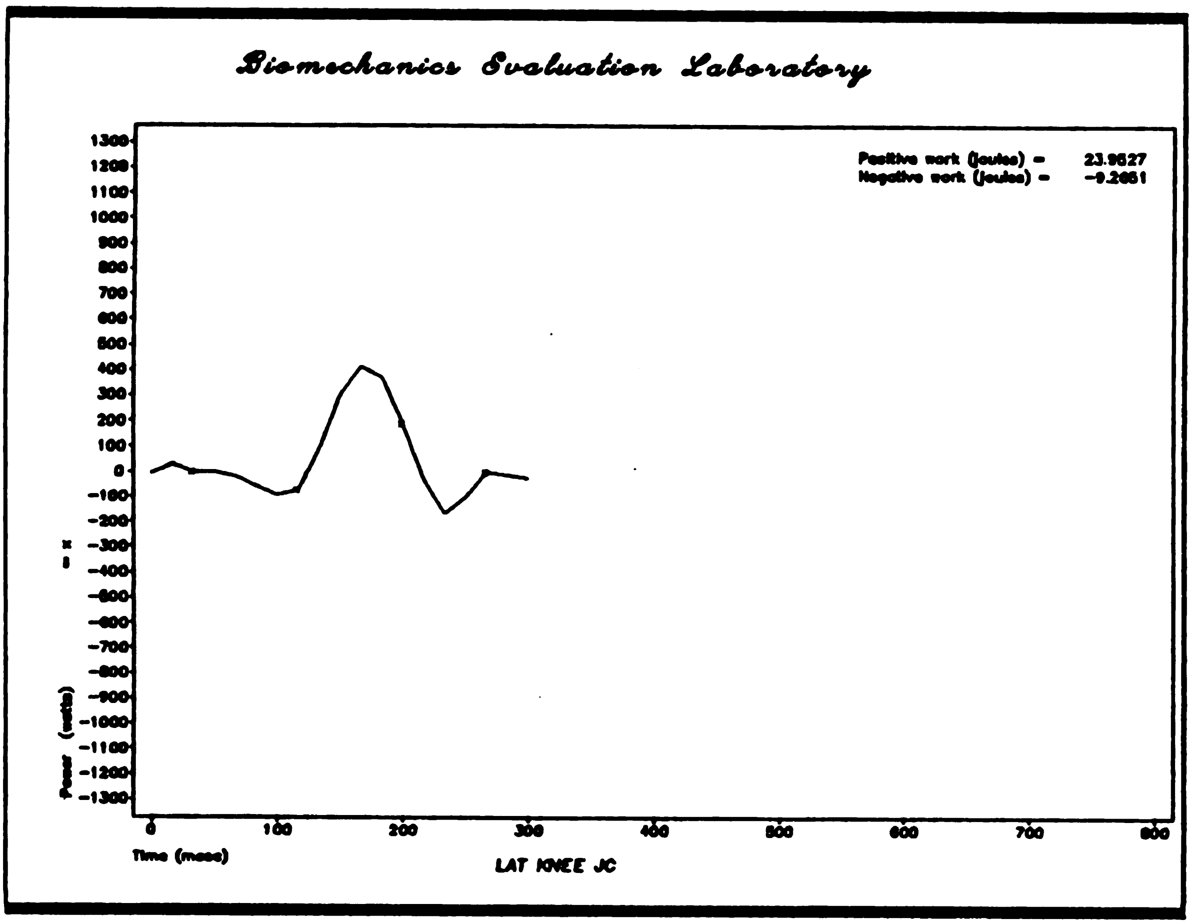

Figure 18e Subject F5

Figure 18 (cont'd.).

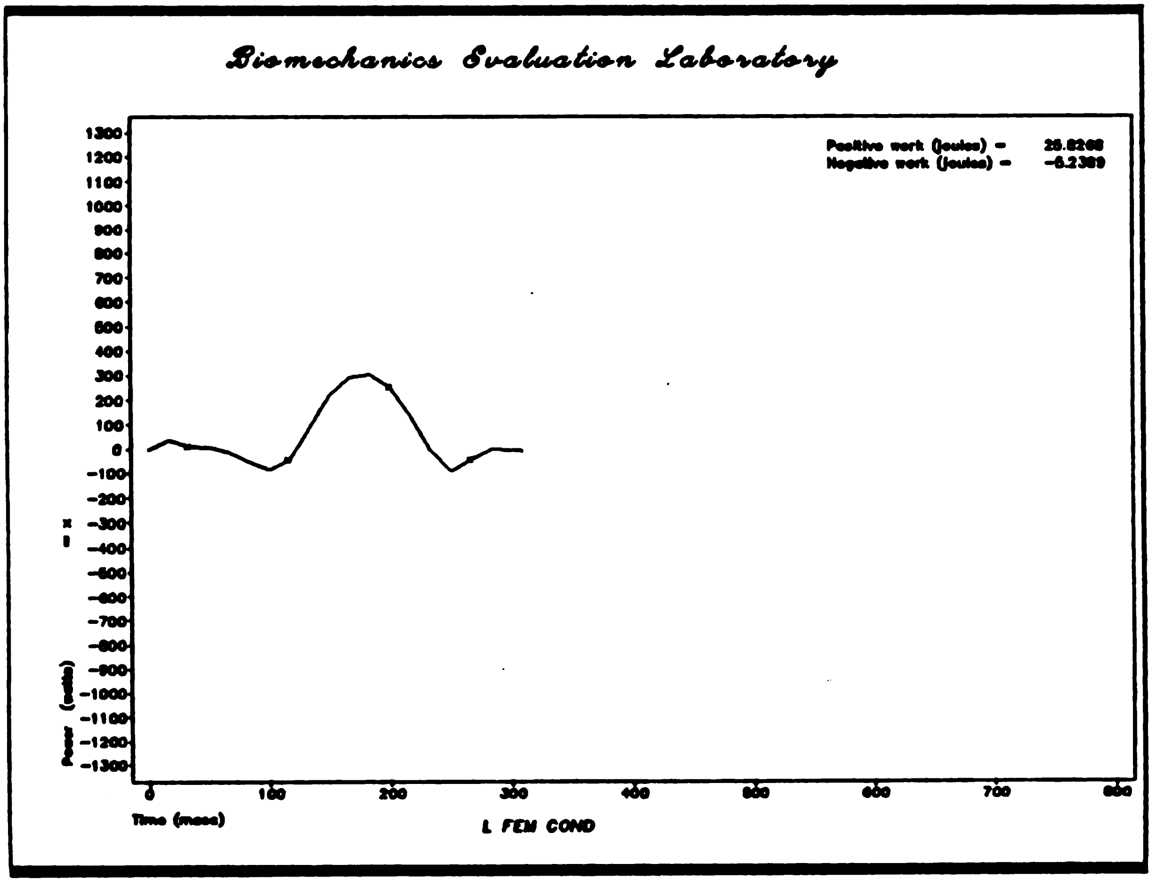

Figure 18f Subject F6

Figure 18 (cont'd.).

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positive burst from toe strike to approximately 15% of stance. This was followed by a small amplitude negative burst until approximately 45% of stance. The third power phase was a large amplitude positive burst from mid stance until just prior to heel off when a final small amplitude negative burst occurred. Following presentation of the EMG data the power curves will be explained relative to the muscle action.

Table 8 displays the average peak positive power of three trials from each subject during the forward running and rearward running conditions. A paired student t-test demonstrated no significant differences in peak positive power between conditions.

Table 8
Average peak positive power - running
(Watts)

Subject #	Forward	Rearward	Difference
F1	1031	636	395
F2	528	439	89
F3	2050	251	1799
F4	538	354	184
F5	634	439	195
F6	467	303	164

$X_D = 471$ $S_D = 658$ $t = 1.753$

Table 9 displays the average peak negative power of three trials from each subject during the forward and rearward running conditions. A paired student t-test demonstrated significantly ($p<.05$) greater peak negative values during the forward running conditions.

Table 9
Average peak negative power - running
(Watts)

Subject #	Forward	Rearward	Difference
F1	-1225	-38	-1187
F2	-550	-89	-461
F3	-1099	-513	-586
F4	-805	-132	-673
F5	-635	-165	-470
F6	-588	-119	-469

$X_D = -641$ $S_D = 280$ $t = -5.608$
$p<.05$

Table 10 presents the average positive work of three trials from each subject during the forward and rearward running conditions. A paired student t-test demonstrated no significant differences between conditions.

Table 10
Average positive work - running
(Joules)

Subject #	Forward	Rearward	Difference
F1	71	40	31
F2	43	23	20
F3	182	9	173
F4	33	31	2
F5	49	26	23
F6	29	29	0

$X_D = 42$ $S_D = 66$ $t = 1.559$

Table 11 presents the average peak negative work of three trials during forward and rearward running conditions. A paired student t-test demonstrated significantly ($p<.05$) greater negative work during the forward running condition.

Table 11
Average negative work - running
(Joules)

Subject #	Forward	Rearward	Difference
F1	-56	-1	-55
F2	-33	-4	-29
F3	-47	-33	-14
F4	-41	-9	-32
F5	-32	-8	-24
F6	-32	-7	-25
	X_D = -30	S_D = 14	t = 5.249 $p<.05$

The EMG activity from the muscles surrounding the knee joint are now analyzed for the two running conditions. The active muscles should be responsible for the knee power generation and absorption requirements. Figure 19 presents the firing pattern of each of the six knee muscles studied during the stance phase of forward running from heel strike (HS) to toe off (TO). Figure 20 presents the firing pattern of each of the six knee muscles studied during the stance phase of rearward running from toe strike (TS) to heel off (HO). Figures 19 and 20 present the average timing pattern from all six subjects. For ease of viewing the direction of travel is from left to right in each of the figures.

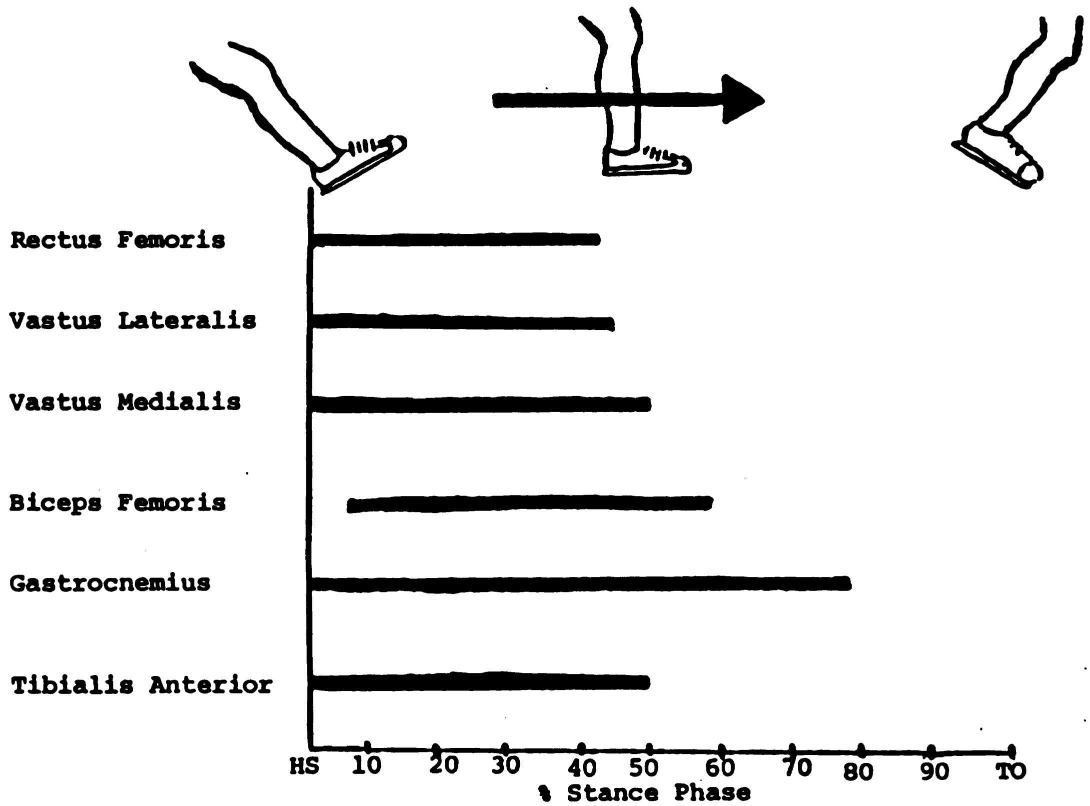

Figure 19 Forward running EMG pattern

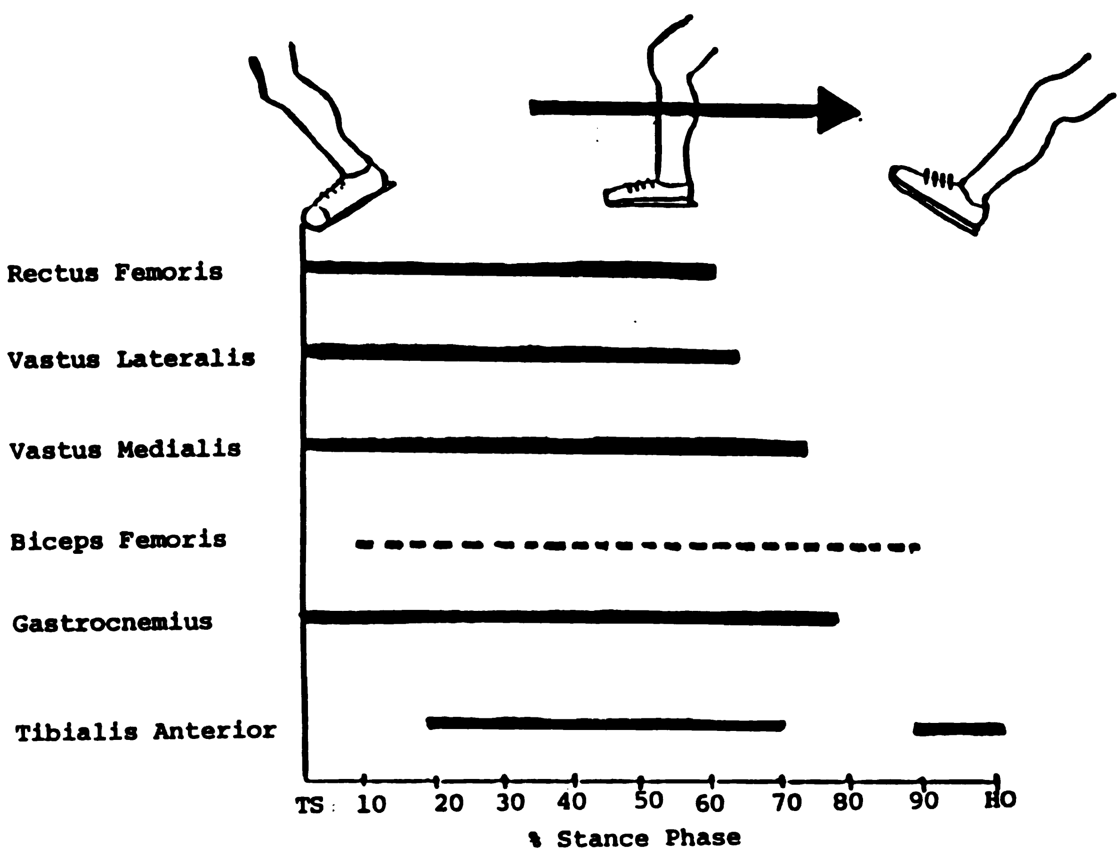

Figure 20 Rearward running EMG pattern

Table 12 gives the mean muscle on time and standard deviation (s.d.) across subjects in percentage (%) of stance phase for each muscle during the forward and rearward running conditions. A within subject design using a paired t test with the level of significance is presented for each muscle.

Table 12
Total on time for individual muscles - running
(%)

Muscle	Forward	Rearward	t-level	p<
Rectus femoris	45 $\pm$ 16	60 $\pm$ 3	-2.070	
Vastus lateralis	48 $\pm$ 7	65 $\pm$ 3	-7.765	.001
Vastus medialis	53 $\pm$ 9	71 $\pm$ 5	-6.379	.01
Biceps femoris	55 $\pm$ 14	77 $\pm$ 4	-4.234	.01
Gastrocnemius	83 $\pm$ 11	78 $\pm$ 7	0	
Tibialis Anterior	50 $\pm$ 37	58 $\pm$ 7	.52	

EMG data during forward running has considerable variability in the reported literature (36). The results presented in Figure 19 are generally consistent with Nillson's(35) and Komi's(26) data. Comparison of the three knee extensor muscles (rectus femoris, vastus lateralis, and vastus medialis) between the forward running and rearward running conditions yield several consistent results. The vastus lateralis, vastus medialis, and biceps femoris had statistically greater (p<.05) total "on time" in the rearward running condition than in the forward running condition. The rectus femoris demonstrated a trend toward increased on time in rearward running but the higher variability in the forward running condition decreased the power of the statistic. The increased variability may have been a function of the dual role of the rectus femoris as both a hip flexor and a knee extensor. Comparing the EMG data to the moment and power curves suggests that in forward running the knee extensors acted primarily eccentrically to absorb the shock of heel strike and then provide a smaller concentric generation of power at mid stance for propulsion. This contrasts sharply with rearward running where the knee extensors acted predominately concentrically as power generators, with only small eccentric phases.

Though the biceps femoris showed statistically greater on time in rearward running, the muscle's activity was of a low level and exemplified by small burst like activity. The biceps femoris also showed greater periods of inconsistent activity across subjects.

No statistically significant differences were noted in gastrocnemius total on time between conditions, but a marked change in type of muscular contraction and function was noted. In forward running the gastrocnemius acted concentrically to propel the limb forward, but in rearward running it acted eccentrically to absorb the shock of toe strike.

No statistically significant differences were noted in tibialis anterior total on time between conditions. The high variability across subjects in the forward running condition was primarily the result of two subjects whose firing patterns were only on from the initial 0-15 % of stance. Functionally, the tibialis anterior acted eccentrically to lower the foot to the floor in forward running. In contrast to forward running, the tibialis anterior demonstrated a two stage pattern in rearward running. The first phase from 20-70 % of stance appeared to be eccentric in nature which allowed a controlled plantar flexion of the ankle. The second phase involved a concentric contraction which functioned to raise the foot from the floor in the later stance phase.

The resultant ground reaction torque is presented in Figure 21. This was a representative trial from subject F5 comparing forward running (red) to rearward running (green). The forward running curve was in general agreement with Holdan and Cavanaugh's(21) finding that the torque was externally directed during the first 50-60% of stance and then internally directed during the remainder. The forward running ground reaction torque in subject F5 suggested pronation from heel contact to approximately 45% of stance followed by supination. The rearward running ground reaction torque was markedly different. In the rearward running condition the torque began in an inward direction followed by an outward directed torque. The rearward running ground reaction torque suggested supination at the subtalar joint from heel contact to approximately 45% stance followed by pronation through the remainder of stance. If Holden and Cavanaugh's assumptions of the ground reaction

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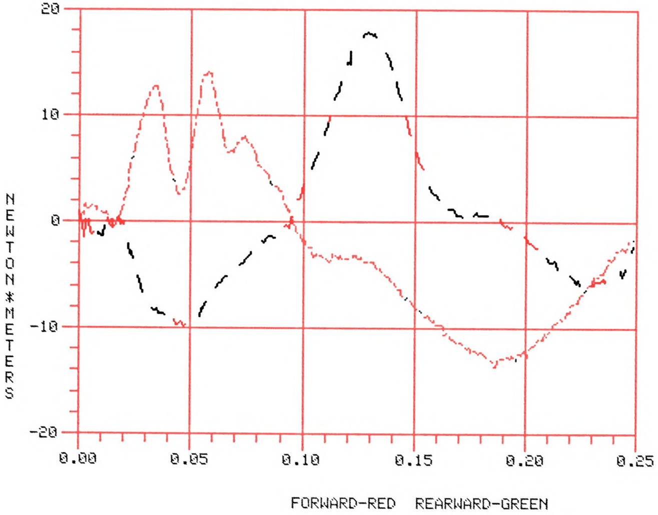

Figure 21

Ground reaction torque forward vs rearward running torques relationship to pronation are valid for rearward running, then it appeared that rearward running prevented pronation from occurring during the first 45% of stance.

VI. CLINICAL IMPLICATIONS

The following chapter presents the application of the results of this research to three common musculoskeletal conditions. The three conditions that will be addressed are patello-femoral dysfunction (PFD), patellar tendinitis, and anterior cruciate ligament reconstruction. The purpose of this chapter is to provide a basis for utilizing retropropulsion in the rehabilitation of patello-femoral dysfunction and patellar tendinitis and to question the use of retropropulsion in anterior cruciate rehabilitation.

Patello-femoral dysfunction (PFD)

The most common area of pain in runners is the patello-femoral joint(24). Typically, a runner suffering from PFD is unable to resume full forward running status for an extended period. It has been the author's experience that a runner suffering from patello-femoral dysfunction can participate in painfree retro-walking and retro-running, when forward walking and running are symptom producing. When forms of retropropulsion are incorporated into the rehabilitaion process, the injured runner returns to forward running sooner.

The results of this research suggest several possible contributions to the above observations. The first benefit of retropropulsion is the longer period of sustained quadriceps concentric activity and the decrease in quadriceps eccentric activity when compared to forward propulsion. Concentric quadriceps exercises are a standard tool in patello-femoral rehabilitation. Though no statistical differences in positive work was found between conditions, the subjects speed of progression was slower in the backward conditions, suggesting that at equal speeds there may have been increased work in the rearward conditions. Negative work was statistically lower in backward versus forward propulsion. Negative work can be performed with less use of a muscle's contractile components and increased use of the non-contractile components of a muscle. This suggests from the standpoint of the knee extensors, retropropulsion is primarily a concentric (contractile component) force generating activity. This finding substantiates Threlkeld's(45) and Mackie's(29) findings of increased concentric torque production after a training program of retro-running.

The clinical observation of painfree retropulsion is also supported by the following example comparing the patello-femoral joint reaction force (PFJRF) in forward running versus rearward running. If the quadricep force (Fq) and the patellar mechanism angle (β) are known, the PFJRF can be calculated if we assume the force in the patellar tendon and the quadriceps to be equal. The moment arm for the quadriceps mechanism is 4.9 cm based on

Smidt's(42) results. Quadriceps force (Fq) can be calculated using equation (4):

$$(4) \quad Fq = \text{Knee Moment} \cdot 100 \ / \ 4.9 \text{ cm}$$

The patellar mechanism angle (β) is calculated using equation (5) from Mathews et al.(31), where (α) is the knee joint angle.

$$(5) \quad \beta = 30.46 + 0.53\ \alpha$$

Equation (6) will yield the patello-femoral joint reaction force.

$$(6) \quad PFJRF = 2Fq \cdot \sin \beta/2$$

Using the peak knee moment during running for subject F5 the PFJRF is 4808 Newtons in forward running and 2038 Newtons in rearward running. This represents a 58% reduction in the PFJRF when running backwards in this subject.

A final possible benefit of retropulsion in the rehabilitation of patello-femoral dysfunction is the apparent reduction of tibial internal rotation during rearward walking, and subtalar joint pronation during rearward running. If symptoms at the patello-femoral joint are the result of excessive or poorly timed pronation, then retropropulsion may reduce the symptoms. The faulty lower extremity mechanics may be the result of weak or poorly firing muscles. If this is the case, then the marked change in muscle demands during retropropulsion may serve to reeducate and strengthen these muscles. This muscle reeducation could possibly be carried over when ambulating forwards.

Patellar tendinitis

Patellar tendinitis or "jumper's knee" is common in athletes involved in running or jumping sports. Curwin and Stanish(13) report that the eccentric loading phase of running and jumping is the major etiological factor contributing to patellar tendinitis. During the rehabilitative process atrophy and weakening of the muscle-tendon unit must be avoided while the inflammatory process subsides. The results of this thesis would support the use of retropropulsion in the rehabilitation of this condition. Retropropulsion would decrease the eccentric work which is involved in forward walking and forward running allowing the inflammatory process adequate time to subside. The concentric quadriceps portion of retropropulsion would decrease atrophy and weakening of the muscle-tendon unit.

Anterior cruciate ligament (ACL)

The literature has reported the use of rearward running in the rehabilitaion of the ACL reconstructed knee(29). Based on the EMG findings of this thesis, retro-running may in fact be detrimental to healing during the early to middle phase of rehabilitation. This research found an increase in quadriceps firing time in rearward running, while the hamstring firing time was inconsistent. This coupled with the gastrocnemius also firing consistently, may cause a force couple to cause a rotation which results in an increased anterior tibial shear in the stance phase of rearward running. Further research is indicated to support or refute the use of retropropulsion in ACL rehabilitation.

VII. CONCLUSIONS

The purpose of this study was to compare the stance phase of rearward walking and running with the stance phase of forward walking and running. Based on the findings of this research the following conclusions are made:

1. Statistically greater peak negative (-) power occurs at the knee during forward walking and forward running conditions when compared to their rearward ambulation counterparts.

2. Statistically greater negative (-) work occurs at the knee during forward walking and forward running when compared to their rearward ambulation counterparts.

3. The pattern of EMG activity during walking is significantly different between forward and rearward conditions. The rectus femoris, vastus lateralis, vastus medialis, and tibialis anterior have greater total on time in rearward walking when compared to forward walking. The biceps femoris has significantly greater on time in forward walking when compared to rearward walking.

4. The pattern of EMG activity in running is significantly different between forward and rearward conditions. The vastus lateralis, vastus medialis, and biceps femoris have significantly greater total on time in rearward running when compared to forward running. The

biceps femoris activity is inconsistent and at a lower level in backward running when compared to forward running.

5. The ground reaction torque is markedly different between the forward and rearward conditions of walking and running. The direction of the torque suggests a decreased tibial internal rotation during rearward walking and a decreased pronation in early-mid stance during rearward running when compared to their forward ambulation counterparts.

The stated conclusions have applications in the training and rehabilitation community. It appears that retropropulsion may be of benefit in overuse injuries such as patello-femoral dysfunction and patellar tendinitis. The use of rearward running may not be benefical during the early to mid healing phase of anterior cruciate ligament reconstruction.

Further studies should focus on the hip and the ankle. Research should also investigate the possiblity of increased tibial anterior shear during rearward running.

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