EMG DURING BICYCLE PEDALLING

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Propulsion of a bicycle via pedalling action of the legs is caused by contraction of the leg muscles. Understanding which muscles are active while pedalling and the forces being developed by the muscles are important for two reasons. First, in an age where fossil fuel supplies are diminishing, many people are turning to bicycles as practical transport. A thorough understanding of the power generation process could very well lead to improvements in efficiency, thereby promoting more widespread use of this form of transportation. Second, both cardiologist and physical rehabilitation therapists use stationary ergometer exercise as a form of physical therapy (Faria and Cavanagh, (1978), McLeod and Blackburn (1980), Kroon (1983)). Clearly, the maximum benefits of this activity can only be derived if the pedalling process is more fully understood.

Determining muscle forces is a complex process. Unless the physiological model is simplified, the mathematical model is indeterminate (Crowninshield (1977), Hardt (1977), Crowninshield and Brand (1981)), in which the number of unknowns (i.e. muscle and joint contact forces) exceeds the number of equilibrium equations. Thus, there are an infinite number of solutions. Two basic methods have been used to eliminate the indeterminacy of the problem (Crowninshield (1977)). The first is to simplify the model until it is mathematically determinate. The second is to use optimization techniques in which constraints are maximized or minimized. By using optimization techniques (Seireg and Arvikar (1973), Penrod et al. (1974), Cappozzo et al. (1975), Crowninshield (1977), Hardt (1977), Crowninshield and Brand (1981)) it is not necessary to simplify the model. Because of the complexities involved, however, it was decided to explore the feasibility of using a simplified model to compute muscle forces during pedalling.

The muscles, which are common to these works, are the rectus femoris, vastus lateralis and medialis, semimembranosus/semitendinosus, biceps femoris, tibialis anterior, gluteus maximus, and the gastrocnemius. Figure 1 depicts the general anatomy of these muscles. The tibialis anterior muscle, commonly known as the shin muscle, dorsi flexes the foot. The gastrocnemius or calf muscle both plantar flexes the foot and flexes the knee. The gluteus maximus extends the hip. The vastus medialis and lateralis make up half of the powerful quadriceps group. These muscles extend the knee. Another member of the quadriceps is the rectus femoris. Because it crosses both the hip and knee joints, it both flexes the hip and extends the knee. The biceps femoris long head, which is part of the hamstrings group, also crosses two joints. This muscle extends the hip and flexes the knee. Because the semimembranosus also crosses two joints, the hip and knee, its function is both to extend the hip and flex the knee.

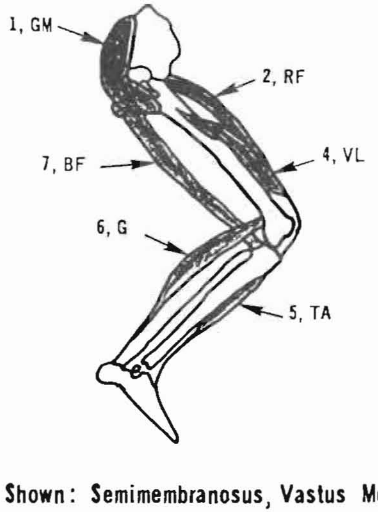

Fig. 1 General anatomy of the leg. 1 - Gluteus Maximus (GM), 2 - Rectus Femoris (RF), 3 - Vastus Medialis (VM), 4 - Vastus Lateralis (VL), 5 - Tibialis Anterior (TA), 6 - Gastrocnemius (G), 7 - Biceps Femoris (BF), 8 - Semimembranosus (S)

NOMENCLATURE

A. <u>English Symbols</u>

P - instantaneous power

T_{rw} - rear wheel torque

g_r - gear ratio

L_2 - crank arm length

F_n - component of pedal force normal to the crank arm

N - number of test subjects

i,j,k - subscripts denoting i^{th} interval, j^{th} test subject and k^{th} case

x_{ij} - IEMG value for a particular muscle

$x_{max,j}$ - max IEMG value for a particular muscle

$\bar{x}_i$ - average of the normalized IEMG values for a particular muscle in the reference case

s_i – standard deviation of the normalized IEMG values for a particular muscle.

$\bar{x}_{i,k}$ – average of the normalized IEMG values for a particular muscle in cases other than the reference case.

N_{tcw}, N_{tcog} – number of teeth on the chainring and rear cog respectively.

B. Greek Symbols

ω – cog angular velocity

Ω – chainring angular velocity

When the results of these previous EMG studies are compared, discrepancies in the regions of muscle activity are apparent. Figure 2 shows the results of the previous works in polar form. The blackened areas in Fig. 2 are regions of muscle activity. With the exception of Gregor et al. (1982), muscles of the quadriceps and hamstrings groups were combined and each group was considered as one muscle by the other researchers. In doing so, information is lost on the region of activity of the individual muscles. Inspection of Fig. 2 clearly shows that there are discrepancies in

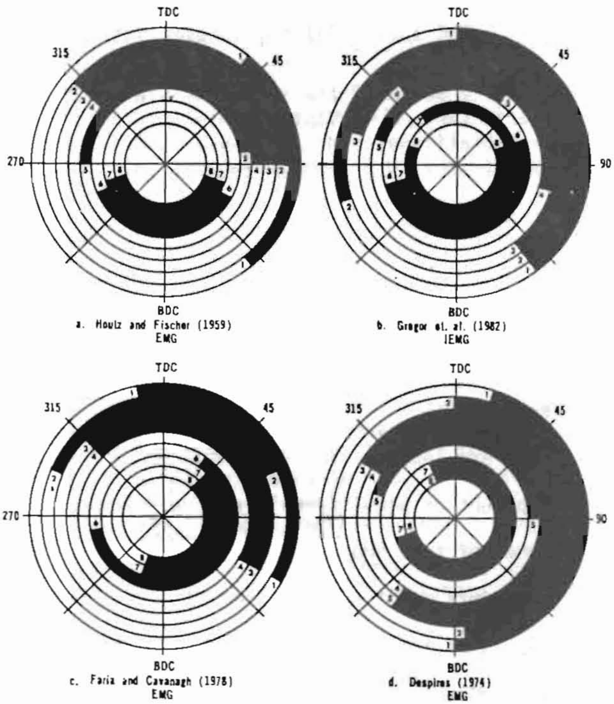

Fig. 2 Comparison of EMG activity regions reported by previous studies.

the results of these previous works. Large differences can be seen in the regions of activity of the quadriceps and hamstring muscles. Possible causes of the discrepancies are different foot-to pedal connections, different equipment (i.e. fixed stationary bicycles versus actual bicycles), and variability of test subjects.

Because of the large differences in the regions of muscle activity from previous work, not enough confidence was gained to construct a simplified model from available data. Thus, it was decided to repeat the electromyogram experiments. Hence, one objective of this study is to determine the regions of activity for eight muscles of the leg.

In addition to determining regions of muscle activity, additional objectives were formulated. In order to appreciate the objectives, it is useful to understand the relation between power output, pedalling rate, gear ratio, and pedal force. Consider that the instantaneous power P is given by

$$P = T_{rw}\,\omega = T_{rw}\,g_r\Omega \tag{1}$$

where T_{rw} is the rear wheel torque, ω is the angular velocity of the cog driving the rear wheel, and Ω is the angular velocity of the chainring (i.e. pedalling rate). The gear ratio is given by $g_r = N_{tcw}/N_{tcog}$ where N_{tcw} and N_{tcog} are the number of teeth on the chainwheel and rear cog respectively. The power can also be expressed by

$$P = F_n L_2 \Omega \tag{2}$$

where F_n is the component of the pedal force normal to the crank and L_2 is the crank arm length. Combining Eqns (1) and (2) yields

$$F_n = \frac{T_{rw}}{L_2}\,g_r = \text{constant} \times g_r \tag{3}$$

so that the pedal force depends directly on the gear ratio.

A second objective is to determine the effects on muscle activity using different foot-to-pedal connections, soft sole shoes without toeclips versus toeclips and cleats. The rationale behind this objective can be traced to the work of Davis and Hull (1981b) who experimented with pedal loading using different foot-pedal connections and found that for constant average power, peak crank torque at 100° was significantly higher when soft sole shoes were used instead of cleats. Inasmuchas the crank torque $T_c = F_n L_2$, the pedal force would increase as well and there is an intuitive dependence of muscle activity levels on pedal force.

A third objective is to relate muscle activity levels to average power output at constant pedalling rate. Both the works of Gregor (1976) and Davis and Hull (1981b) have explored the effects of power output on pedal forces and found (not surprisingly) that the peak value of the pedal force is directly related to the power. This result is clear from Eq (2). Similar to the second objective, the third objective further examines the dependence of muscle activity levels on pedal force.

The final objective is to explore the relationship between muscle activity and seat height. This objective stems from several previous studies which have addressed the issue of optimal seat height. In a study of oxygen consumption with respect to seat height, Hamley and Thomas (1967) concluded that the most efficient seat height based on minimum oxygen uptake is at 109 percent of the symphysis pubis height. Nordeen-Snyder (1977) states that a seat height of 100 percent of the trochanteric height is the most efficient. Because 100 percent of the trochanter height corresponds to 107.1 percent of the symphysis pubis height, Nordeen-Snyder (1977) concluded that both studies produce similar results. Taking an analytical approach, Hull and Butler (1981) concluded that higher quadriceps loads are developed when the seat height is lowered. Based on both the experimental and analytical results, muscle activity levels are expected to bear an inverse relationship to seat height.

METHODS

The subjects in the experiment were six experienced male cyclists whose pertinent anthropometric data and experience are summarized in Table 1. In the experiments, subjects rode a bicycle of the same size frame (23 inch) as the bicycle they normally used. The bicycle was ridden on rollers to simulate actual riding. Rollers simulate actual riding conditions because no lateral support is provided for the bicycle. Also, both wheels revolve giving the feeling of road conditions; thus, the rider must balance himself as in actual riding. Because tire pressure affects rolling friction and thus the work needed to turn the pedals, the tires were kept at a constant pressure of 90 PSIG during the experiment.

TABLE 1. TEST SUBJECT DATA

SUBJECT	HEIGHT	WEIGHT	TROCHANTER LENGTH	EXPERIENCE
A	1.78 m	75.75 kg	0.89 m	tourist
B	1.75 m	74.75 kg	0.82 m	recreational rider
C	1.75 m	70.31 kg	0.90 m	former racer
D	1.85 m	74.84 kg	0.91 m	former racer
E	1.74 m	64.40 kg	0.90 m	racer
F	1.80 m	70.31 kg	0.89 m	racer

Despires (1974) reported changes in muscle activity when the seat angle was changed. Accordingly, the same seat (Cool Gear patent no. 3807793) was used throughout the experiments, and it was kept horizontal. The seat height was adjusted to 100 percent and 95 percent of the trochanter length of each subject. Trochanter length is defined as the distance measured from the greater trochanter to the floor with the subject standing straight-legged on bare feet. Seat height is defined as the distance from the top of the seat to the top surface of the pedal platform, measured along the seat tube with the crank arm in the down position but parallel to the seat tube.

The muscles (see Table 2) that were considered were the gluteus maximus, vastus medialis and lateralis, biceps femoris long head, rectus femoris, tibialis anterior, the lateral head of the gastrocnemius and the semimembranosus.

TABLE 2. MUSCLES CONSIDERED

1. Gluteus maximus (GM)
2. Rectus femoris (RF)
3. Vastus medialis (VM)
4. Vastus lateralis (VL)
5. Tibialis anterior (TA)
6. Gastrocnemius (G)
7. Biceps femoris (BF)
8. Semimembranosus (S)

Silver disc surface electrodes with a diameter of 1.5 cm were used to pick up the electromyogram potentials. Electrode placement was determined as described by Basmajian et al. (1980). Before attaching the electrodes, the skin was shaved at the appropriate areas, rubbed with 320 grit emery cloth to remove the dead skin layer, and cleaned with alcohol. To reduce electrical resistance between electrode and skin, the cavity in the electrode was filled with Hewlett-Packard Redux Creme. The electrodes were attached to the skin with Hewlett-Packard 14029A adhesive discs. The electrode wires were taped to the skin to reduce wire movement and prevent accidental removal of the electrodes. A velcro strap was worn around the thigh to further secure the electrode cables. Data were taken from the right leg.

A four channel Beckman R612 Dynograph Recorder was used to process the electromyogram signals. The amplifier outputs were digitized and stored in an LSI-11/23 computer. For data acquisition, the LSI-11/23 has a 16 channel multiplexer with a 12 bit analog-to-digital converter. The computer has powerful graphics capability so that the data could be viewed a short time after they were taken. Data were digitized at 2000 samples per second in the EMG data acquisition routine.

In addition to the EMG signals, both the crank angle and cadence were recorded. The crank angle was monitored by a continuous rotation potentiometer which was clamped to the seat tube near the bottom bracket. A gear is attached to the wiper and is driven by an identical gear which is clamped to the chainwheel. With a 1:1 gear ratio, the voltage from the potentiometer is proportional to the crank angle. Data acquisition was triggered when the crank angle potentiometer indicated that the crank was at top dead center. Details of the computer processing of the potentiometer output can be found in Hull and Davis (1981a). Cadence was determined by using a PACER 2000H, which is a bicycle computer capable of monitoring cadence, elapsed time, instantaneous speed, average speed, distance traveled, and heart rate.

For each of the pedalling cases studied (see Table 3), data were recorded for two trials of one revolution each and then averaged. Preliminary experiments indicated that the EMG data for a test subject were highly repeatable not only over a number of revolutions in a given trial but also over seperate trials. Accordingly, two trials were deemed adequate to give reliable information.

TABLE 3. EMG EXPERIMENT CASES

CASE	RPM	GEAR[1]	SINGLE PEDAL POWER LEVEL	SHOES	SEAT HEIGHT[2]
1	80	52 X 19	100 W	CLEATS	100 percent
2	80	52 X 19	100 W	SOFT SOLE	100 percent
3	80	52 X 23	83 W	CLEATS	100 percent
4	80	52 X 15	125 W	CLEATS	100 percent
5	80	52 X 19	100 W	CLEATS	95 percent

[1] GEAR RATIO (NUMBER OF TEETH: CHAINWHEEL X COG)

[2] PERCENT OF TROCHANTER LENGTH

The experimental procedure had to accomodate the fact that of the eight muscles considered, only four could be sampled at one time. After acquiring data from four muscles using the five conditions shown in Table 3, the electrodes were detached and reapplied to the next four muscles. The procedure was then repeated. Because of the intrasubject reproducability of the data, superimposing the muscle activity histories does not limit the accuracy of results.

Table 3 shows the different cases in the study. Case 1 is the reference case to which all cases are compared. The single pedal power level for Case 1 was about 100 W. Case 2, which uses soft sole shoes, is directed towards satisfying the second objective. Cases 3 and 4, where the power levels were adjusted by varying the gear ratio, focus on the third objective. The gear ratios chosen are such that the reference power level (100 W) is 20 percent greater than Case 3 (83 W) and 20 percent less than Case 4 (125 W). Case 5, which differs from the reference only in the seat height, is clearly directed towards the final objective.

Digitized EMG signals were processed to indicate both on-off activity regions and activity levels. On-off activity regions were indicated by simply generating polar plots from visual inspection of activity histories. The validity of this rather subjective technique can be traced to the work of Basmajian (1978). Examples of such plots may be found in Figs. 3 and 4. Activity levels were determined via specialized computer software. First digitized EMG signals were full-wave rectified. Then, using a window of 75 ms according to Norman et. al. (1978), the rectified signal was integrated over adjacent 75 ms data segments. Inasmuchas the pedalling rate was 80 RPM, the integration produced ten IEMG values for each muscle at crank angle intervals seperated by 36°.

Because of the relative nature of EMG measurements both between subjects and between muscles, further processing was warranted. First a procedure was developed for establishing a baseline from the reference case. For each muscle IEMG data for each interval were normalized to each subject's maximum and then averaged over the six subjects. The sample average $\overline{x}_i$ for a particular muscle is given by

$$\bar{x}_i = \frac{1}{N} \sum_{j=1}^{N} \left(\frac{x_{ij}}{x_{max\,j}}\right) \tag{4}$$

where i is the interval of interest, j is the subject, N = 6 for the six subjects, and $x_{max\ j}$ is an individual subject's maximum. Also computed for each interval was the sample standard deviation s_i given by

$$s_i = \left[\frac{\sum_{j=1}^{N} \left(\frac{x_{ij}}{x_{max\,j}} - \bar{x}_i\right)^2}{N-1}\right]^{1/2} \tag{5}$$

The results of these computations are plotted in Fig. 5. The values of $\bar{x}_i$ are indicated by the elevations of individual bars while the corresponding values of s_i are indicated by the vertical lines crossing the bars. Vertical lines are scaled to show $\pm$ 1/2 standard deviation about the average. To facilitate understanding the information presented in Fig. 5 several points are worth mentioning. First, the vertical scale is dimensionless. This follows from Eq. (4). Second, note that although the maximum possible value of $\bar{x}_i$ is 1.0 for a particular muscle, none of the maximum values of $\bar{x}_i$ equal this value. The reason for this is that the interval containing the individual subject's maximum $x_{max,j}$ was not the same for all subjects. The variation s_i at each interval is due, therefore, not only to relative differences in IEMG levels when all subjects are considered, but also to differences in timing.

In order to compare the results of Cases 2-5 to the reference, for each muscle each IEMG level of the case of interest was normalized to the maximum of the reference for each subject. Similar to the reference case the normalized data were then averaged according to

$$\bar{x}_{ik} = \frac{1}{N} \sum_{j=1}^{N} \left(\frac{x_{ij}^{k}}{x_{max\,j}}\right) \tag{6}$$

where i is the interval, j is the subject, and k is the case of interest. The $\bar{x}_{ik}$ for Cases 2-5 are plotted in Figs. 6-9 respectively. Note that in order to facilitate comparing the results in these figures to the reference, $\bar{x}_i$ given by Eq. (4) are also plotted and appear as dashed lines.

RESULTS

In examining the on-off muscle activity regions for the test subjects it was found that for each subject the regions for the five cases are similar. Among subjects, however, differences in the regions of muscle activity are apparent. Figures 3a-f illustrate the on-off activity regions for the individual subjects over a full revolution for the reference case. The shaded portions are regions of muscle activity with the muscles numbered according to Table 2. The most pronounced differences are in regions of activity for the rectus femoris and tibialis anterior. Considering the variability in both experience and anatomical characteristics of the six subjects, the comparison of EMG polar plots is surprisingly good.

Referring now to Fig. 4, which illustrates the average of the on-off muscle activity regions of the six subjects over a full revolution of the crank arm for the reference case, the onset of activity for all muscles in the quadriceps group occurs well before 0° or top dead center (TDC). Onset

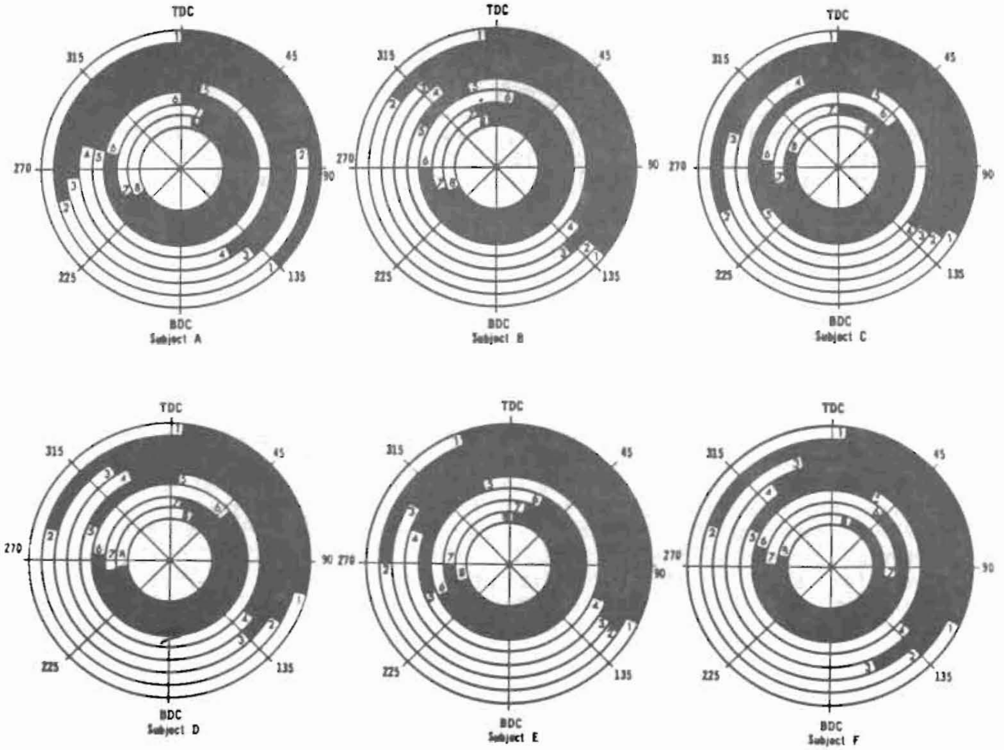

Fig. 3 Regions of on-off muscle activity for individual test subjects (reference Case 1)

of activity for the rectus femoris is close to the middle of the recovery phase (200° to TDC) and terminates at about 120° to 130°. Even though the vastus medialis and lateralis are part of the same muscle group as the rectus femoris, it is interesting to note that onset of activity for these muscles is generally later than that of the rectus femoris. Termination of activity for the muscles of the quadriceps group is at about the same crank angle.

Figure 4 also illustrates that activity of all hip extensors commences at the same angle. The gluteus maximus is active from TDC to about 130°, which is inside the region of the power stroke (25° to 160°). The muscle of the hamstrings group, biceps femoris, and semimembranosus have the largest regions of activity, from just after TDC to about the middle part of the recovery phase. Their activity in the power stroke is an interesting phenomenon because these muscles are antagonistic to the quadriceps. This point will be discussed in more detail shortly.

The muscles of the lower leg offer an interesting contrast to the muscles of the upper leg. The region of activity of the tibialis anterior muscle is in the second half of the recovery phase from about 280° to slightly after TDC. Onset of activity of the gastrocnemius muscle is at about 30° and termination is at about 270°. Note that activity of the gastrocnemius starts a few degrees after termination of the tibialis

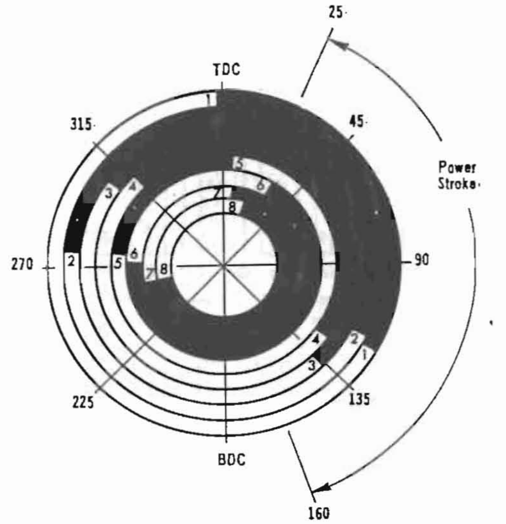

Fig. 4 Average regions of muscle activity for the reference case.

anterior. Also, termination of activity of the gastroc occurs a few degrees before onset of the tibialis anterior. The activity regions for these two muscles are on opposite sides of the pedal cycle.

It is interesting to note that only slight variations in the on-off muscle activity regions resulted with different pedalling conditions. This suggests that muscle activity patterns (i.e. timing) are not significantly affected by changes in pedalling conditions, and one can conclude that the

Figure 5 illustrates the average plots of the IEMG for the six test subjects using the first case in Table 3. On the average, the region of greatest activity (greater than 50 percent) for the gluteus maximus for the six subjects is between 10° and 110°. This is in the power stroke where the hip is being extended. The average region of greatest activity of the rectus femoris is from about 300° to 70°. Both vastii muscles have regions of greatest activity between 340° and about 100°. Onset of greatest activity for the vastii muscles is about 40° to 50° later than that of the rectus femoris. The tibialis anterior, on the average, has its greatest activity at about 300° to TDC. The gastrocnemius, which is antagonistic to the tibialis anterior, exhibits the greatest activity on the opposite side of the crank cycle. Note that the gastrocnemius, a knee flexor, is active when the quadriceps are extending the knee (45° to 110°). One of the hamstring muscles considered, the biceps femoris, exhibits a region of greatest activity between 80° and bottom dead center (BDC). The other hamstring muscle, semimembranosus, shows more prolonged activity with the greatest activity region extending from 60° to 240°.

A similar pattern to the reference is seen when the IEMG of the other experiments are examined (Figs. 6-9). The values, however, are different, indicating that a change in muscle activity levels occurred when the pedalling conditions were changed. The different pedalling conditions will now be compared with the reference.

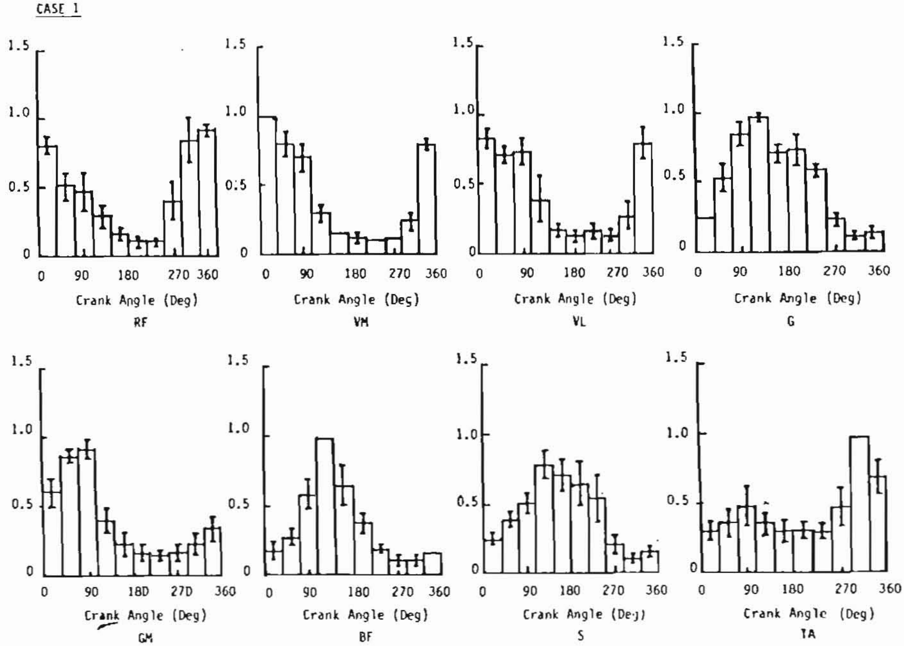

Fig. 5 Average IEMG results for the eight muscles (reference Case 1) Vertical lines indicate ± 1/2 standard deviation.

Case 2: Examination of the data in Fig. 6 reveals that four muscles exhibit marked increases in activity levels in greatest activity regions. Both vastus medialis and lateralis show increased muscle activity of up to 53 percent and 67 percent respectively. Large increases in muscle activity are seen for the single joint muscles which act to extend the knee. Large increases in muscle activity levels are seen not only by the vastii but also by the muscles of the hamstring group. Both the biceps femoris and semimembranosus show increases of up to 133 percent and 122 percent respectively in their regions of greatest activity. Two muscles, gastrocnemius and rectus femoris, exhibit small decreases in at least one interval in the maximum activity regions.

Cases 3 and 4: Comparing the results in Fig. 7 (low power) and Fig. 8 (high power) on a muscle by muscle basis, first, note that the gluteus maximus, a single joint hip extensor, exhibits decreased muscle activity levels in Fig. 7 and increased activity levels in Fig. 8. This trend is not consistent for the biceps femoris and semimembranosus which also function as hip extensors. Both Figs. 7 and 8 show generally increased activity for these muscles. The activity increase for both muscles in Fig. 8, however, is approximately 45 percent greater than Fig. 7 at the maximum activity interval. It can be concluded that all of the hip extensor muscles including the hamstrings exhibit greater activity in Case 4 than in Case 3.

Considering the muscles of the quadriceps group, Fig. 7 indicates that the activity levels of these muscles in the maximum activity region do not

CASE 2

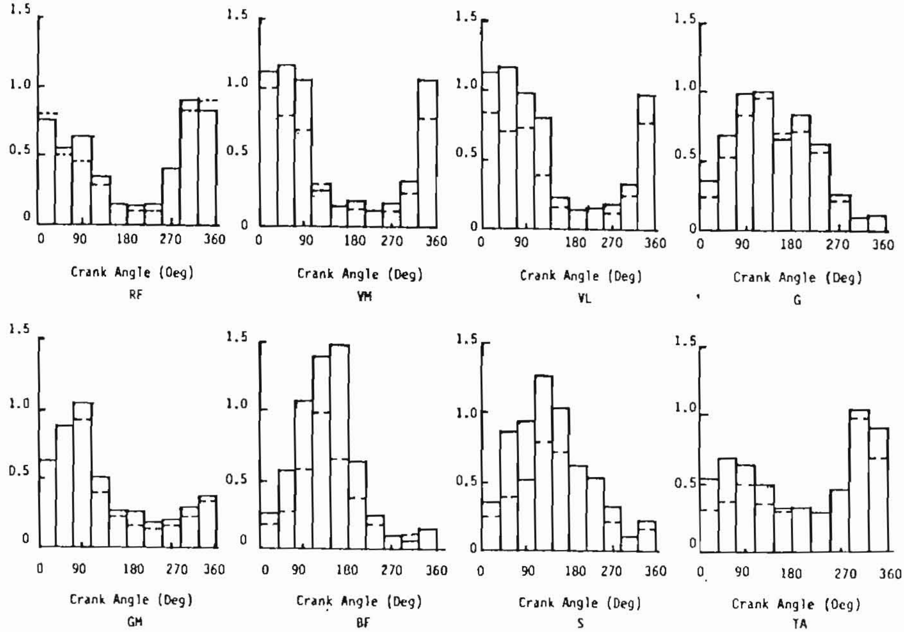

Fig. 6 Average IEMG results for the eight muscles (soft sole shoes, Case 2). Dashed lines indicate average results for Case 1.

differ markedly from the reference. Fig. 8, on the other hand, indicates approximately a 70 percent increase for both the rectus femoris and vastus lateralis with a smaller increase of 14 percent for the vastus medialis at the maximum activity interval. Similar to the hamstrings, the quadriceps group exhibits greater activity in Case 4 than in Case 3. In contrast to the hamstrings group, the activity levels of the quadriceps in Case 3 are not increased above the reference.

Examining the results for the muscles of the lower leg, Fig. 7 indicates that the gastroc activity is consistently below the reference in the maximum activity region. Fig. 8, however, indicates no marked difference. Similar to the hamstrings, the tibialis anterior exhibits large increases over the reference in both Figs. 7 and 8. Also similar to the hamstrings, the tibialis anterior activity level is greater for Case 4 than for Case 3.

Case 5: Figure 9 illustrates that all muscles of the quadriceps group exhibit increases in muscle activity levels throughout the maximum activity regions. Largest increases for both the rectus femoris and vastus lateralis exceed 70 percent while the largest increase in vastus medialis activity exceeds 30 percent. Similar to the muscles of the quadriceps group, the activity levels of the muscles of the hamstring group increase significantly. Both biceps femoris and semimembranosus indicate maximum increases of 120 percent and 130 percent respectively. Contrasted to the two joint hip extensors, the single joint hip extensor gluteus maximus shows no marked change from the reference.

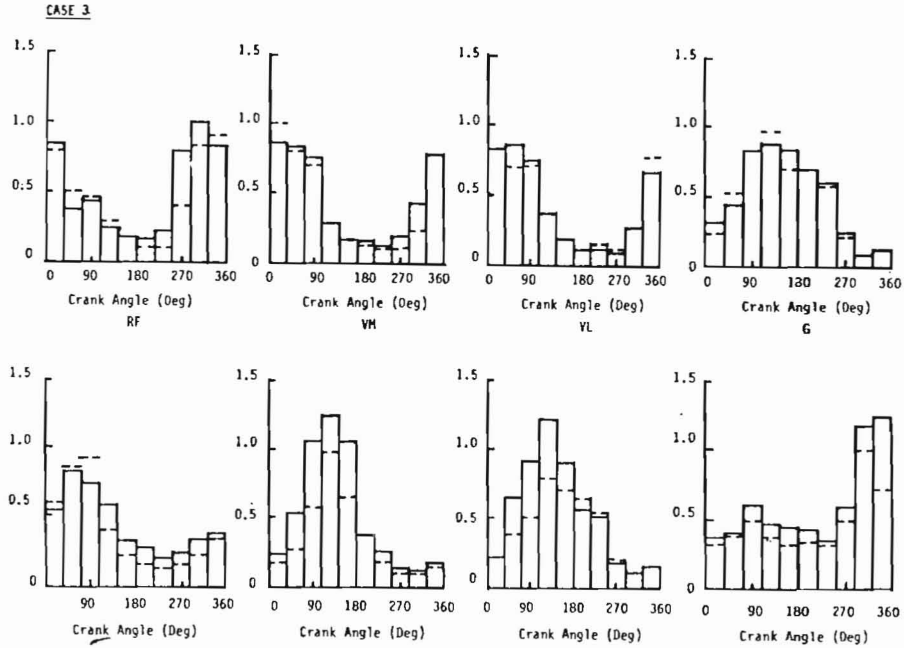

Fig. 7 Average IEMG results for the eight muscles (lower gear ratio, Case 3). Dashed lines indicate average results for Case 1.

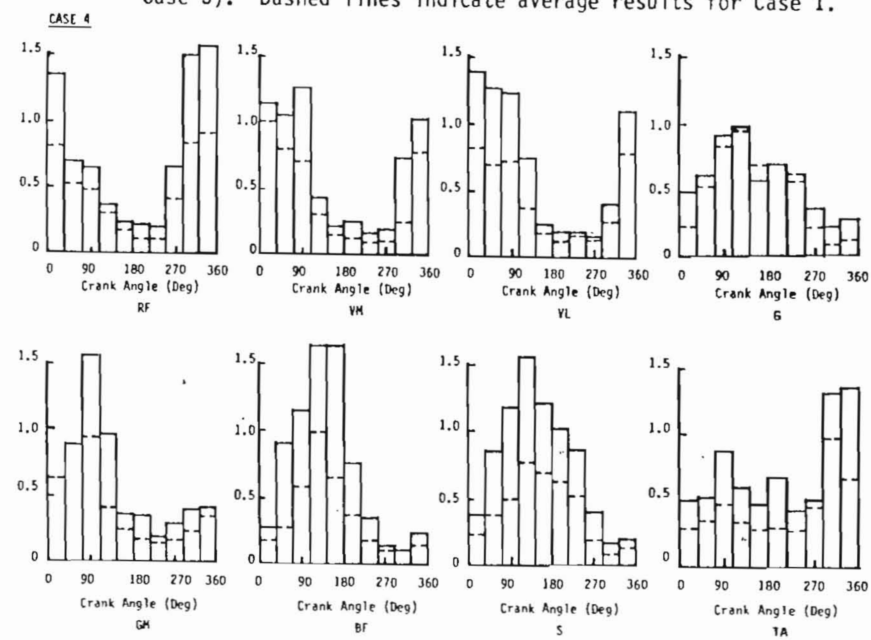

Fig. 8 Average IEMG results for the eight muscles (higher gear ratio, Case 4). Dashed lines indicate average results for Case 1.

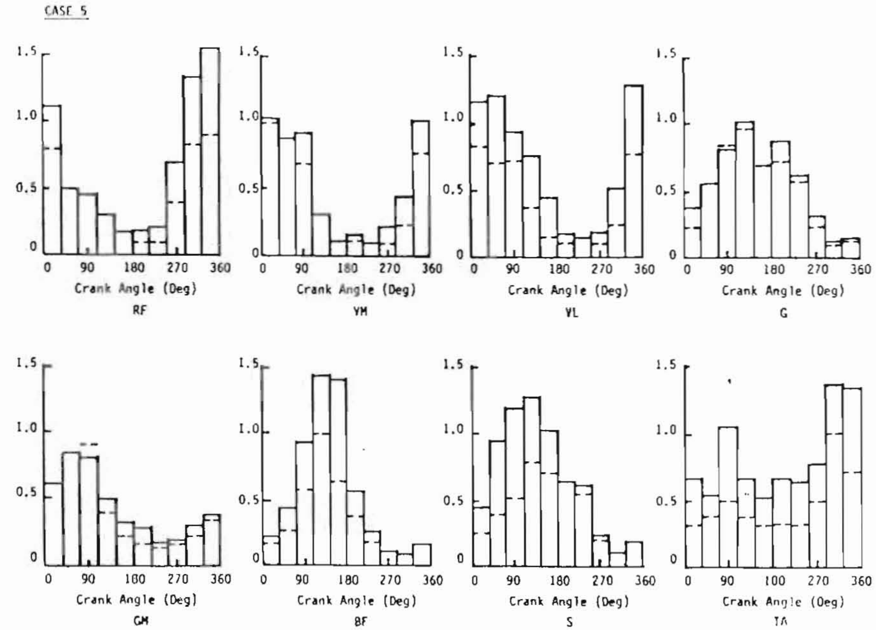

Fig. 9 Average IEMG results for the eight muscles (lower seat height, Case 5). Dashed lines indicate average results for Case 1.

DISCUSSION

Contrary to the polar plot of on-off muscle activity in Fig. 4, the IEMG results in Fig. 5 show that there is no significant overlap in regions of greatest activity of the quadriceps and hamstrings. Accordingly, there is not much antagonism between these muscles. Thus, looking at only on-off EMG data may not show the whole picture of muscle activity and also may result in incorrect conclusions. To avoid this pitfall, a quantitative EMG measure must supplement the EMG data.

The lack of significant co-contraction of agonist/antagonist muscles in both the leg crossing the ankle and the thigh crossing the knee establishes the feasibility of using a simplified model to compute muscle forces. Because different muscles are active during different regions of the crank cycle, however, the model must be region dependent. Referring to Fig. 5, three different regions can be identified. One is the region from TDC to 90°. During this region the active muscles include the gluteus maximus, the muscles of the quadriceps group, and the gastrocnemius. A second region extends from 90° to 270° where the active muscles are limited to the gastrocnemius and the hamstrings. Muscles active in the third region, which ranges from 270° to TDC, are the tibialis anterior and the rectus femoris.

Once the active muscles are specified, the muscle forces can be computed. The procedure calls for first specifying both the external loads (i.e. pedal forces) and the kinematic variables. Then, the three joint moments could be computed. With the joint moments known, the individual muscle forces can be computed by solving the equilibrium problem at first the ankle, then the knee, and finally the hip. For each of the three regions, unique solutions to the muscle forces are possible. Consequently, the computational complexity of the muscle force solution problem is diminished greatly. Although this modelling procedure might not provide the detailed picture of muscle forces that solution to the redundant problem might provide, it should prove useful for modelling problems directed towards gaining insight into the biomechanics of pedalling.

Before leaving the subject of modelling, two points regarding the above procedure are worth mentioning. The first is that although the modelling procedure described relies on the lack of co-contraction on the part of both the gastroc/tibialis and hamstrings/quadriceps, it does allow for co-contraction of agonist/antagonist muscles crossing the hip. This allowance is important for computing muscle forces in the first region (TDC-90°) because of the simultaneous activity of the gluteus maximus and rectus femoris. The second is that the modelling procedure must include some means of allocating force to individual members of the quadriceps group. In the first region, even though all three quadriceps muscles are active, only the two-joint muscle rectus femoris crosses the hip. In order to compute the gluteus force at the hip, the percentage of the total quadriceps force (determined from equilibrium at the knee) carried by the rectus femoris must be estimated. Two methods for estimating this percentage are possible. One would include defining the percentages based on the assumption that force carried by each muscle is proportional to its cross-sectional area. More elaborate, the second would entail calibrating the IEMG of the rectus femoris.

While the regions of muscle activity found in this study compare favorably with the regions reported by two of the previous studies, large discrepancies for at least one muscle are apparent when comparing the results herein with the other two previous studies depicted in Fig. 2. In comparing Fig. 3 to Figs. 2a-d, the closest correspondence is with the results of Gregor et al. (1982) shown in Fig. 2b. The onset and termination of activity for all muscles compare within 20°. Considering the variability of the results from subject to subject (see Fig. 3), such differences are quite reasonable. Inasmuch as Gregor et al. (1982) used experienced test subjects wearing toeclips and cleats, it is not surprising that the results from the two studies agree. With the exception of the rectus femoris in which activity terminates approximately 45° before the average angle in Fig. 4a, the results of Faria and Cavanagh (1978) are also within expected limits. Noting that test subject A in Fig. 3a terminated activity of the rectus femoris at about 80°, the 45° descrepancy may be at the extreme of normal limits.

Discrepancies between the results herein and those of Houtz and Fisher (1959) and Despires (1974) are more severe. Despires, for example, reports that the vastii muscles remain active past BDC (see Fig. 2d). This result is surprising because the knee is clearly being flexed beyond this point. Activity of the vastii during knee flexion implies these muscles are actively resisting this motion. Although this action is certainly possible on the part of a test subject, it would, in general, be unexpected. Houtz and Fisher (see Fig. 2a), on the other hand, report no activity in the vastii muscles beyond 90°. This result is much more palatable than that of Despires. Despite the fact that extension of the knee occurs beyond this point, vastii activity would be expected to cease when the joint moment required for equilibrium changes direction. Inasmuchas the direction is determined by the orientation of the resultant pedal force vector, it is certainly possible that the vector calls for this change before the crank arm reaches horizontal.

In discussing the results related to the various experiment cases in Table 3 recall that te objective of Case 2 was to explore muscle activity when using soft sole shoes vs toeclips and cleats. As anticipated by the results of Davis and Hull (1981b), the IEMG levels especially of the muscles active during the power stroke (25°-160°) increase when soft soled shoes are used. Bearing in mind that the average power over a crank revolution was the same for Cases 1 and 2, it might also be anticipated that the increase in IEMG activity in the power stroke region might be accompanied by a decrease in IEMG activity for other muscles elsewhere in the crank cycle. A small decrease, in fact, is seen in one and two intervals of the maximum activity regions for the gastrocnemius and rectus femoris respectively. Accordingly, the load sharing of muscles in generating power with the use of toeclips and cleats (see Davis and Hull (1981b)) appears to be valid to some degree.

To explore the relation between muscle activity levels and the loading developed at the pedal, the pedal loading variable was isolated by maintaining the reference cadence of 80 rpm and using lower (Case 3) and higher (Case 4) gear ratios. Note that according to Eq (3), the pedal force is directly related to the gear ratio. In comparing Fig. 7 (lower power) with Fig. 8 (higher power) the most striking dissimilarity is the number of intervals where the IEMG is lower than the reference. The greater number of intervals with IEMG lower than te reference in Fig. 7 suggests generally lower muscular activity than the reference while the virtual absense of intervals with IEMG lower than the reference in Fig. 8 indicates greater muscle activity than the reference. Accordingly, it can be concluded that increased muscle activity accompanies increased pedal load.

While the results of Cases 3 and 4 generally confirm that muscle activity increases with increasing pedal load, several anomalies surfaced. The increased levels relative to the reference of the hamstrings and the tibialis anterior at decreased pedal load are two. A third is the lack of an increase in the gastroc activity at higher pedal loads. A fourth is the lack of a marked decrease throughout the maximum activity regions in quadriceps activity at lower pedal loads.

At least two factors may account for these apparent inconsistencies. First, it is possible that test subjects change their pedalling technique depending on the load demand. Pedal loading profiles presented by Davis and Hull (1981b) lend some credence to this argument. Second, the IEMG data exhibits variability between the subjects. This variability is especially large for both hamstrings and the tibialis anterior in the maximum activity regions. This variability is due primarily to pedalling technique where slight differences in timing of maximum IEMG intervals occur between the subjects.

Case 5 was performed to determine the muscle activity when the seat is lowered to 95 percent of the trochanter length. The increases in muscle activity levels for both the quadriceps and hamstrings at the lower seat height are consistent with the results of Nordeen-Snyder (1977) and Hamley and Thomas (1967) who concluded that a seat height of 100 percent of the trochanter length and 109 percent of the symphysis pubis height, respectively, is the most efficient when oxygen consumption is taken as the criterion. An increase in oxygen consumption is the result of the increase in muscle activity. The increase in muscle activity levels at a lower seat height is also consistent with the biomechanical analysis of muscle forces in the leg by Hull and Butler (1981), who concluded that higher quadriceps force will result when the seat is lowered. Higher muscle force would be the result of higher muscle activity.

CONCLUSION

The primary motivation for the study reported herein was to identify timing patterns of the leg muscles which play an important role in generating human power by means of pedalling action. Timing patterns have been identified by means of surface EMG electrodes where EMG data have served for computation of IEMG. One significant observation is that the general activity picture at all three joints is characterized by little cocontraction of agonist/antagonist muscles. This is especially true for the gastrocnemius/tibialis anterior at the ankle and the hamstrings/quadriceps at the knee. One result of this observation is that muscle forces may be computed uniquely by solving a series of equilibrium problems each valid over a different region of the crank cycle. Regions are differentiated by the combinations of muscles which are simultaneously active.

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