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Electromyographic gait assessment, part 1: Adult EMG profiles and walking speed*

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Abstract—The profiles of the linear envelopes of surface electromyograms of seven major muscles in adults were studied as a function of walking speed. The ensemble statistical properties of average and standard deviation were used to quantitate the characteristics of the electromyograhic patterns. Analysis of variance was used to indicate periods in the gait stride that were effected by walking speed. Two general types of pattern change were observed. One was that the fundamental phasing of muscle activity never changed but that relative amplitudes of the phases were modulated as speed increased. The other was that different phases of activity existed for different walking speeds. The timing of most phases (as expressed in percentage of the stride) decreased as speed increased. This suggests that the time base should be further normalized by stance and swing phases.

INTRODUCTION

The effective evaluation of hemiparetic or other electromyographic (EMG) patterns associated with pathologic gait is dependent upon a standardized normative data base. However, there are several aspects of the state of knowledge that act as impediments to the establishment of such a data base.

Over the last three decades, investigations of normal muscular synergy patterns have been undertaken for a variety of purposes using different paradigms, and the results have been presented in different formats (14). The paradigms differ in type of footwear, walking speed requested, walking surface, and electrode type. The number of subjects studied varied between 6 and 25. The presentation format varied among raw data, on-off representation, and average envelopes. More recently, attention has been directed to variabilities in EMG patterns that exist within the population of normal individuals. Significant differences in the phasings of some major muscle groups, especially the quadriceps and hamstrings, exist (1,13,18). In addition, myoelectric phasings change with walking speed (4,5,10,18).

Recently, the use of average linear envelopes (LE) has been reemphasized as a modality for both expressing and analyzing EMG patterns (6,16), and for assessing pathologic EMG patterns (20). This is a valid representation since the stride-to-stride and trial-to-trial repeatability is very high (8). Attendant to the use of envelopes is the amount of intersubject pattern variability caused artifactually by the measurement process and filtering technique. For example, a major consideration when comparing EMG patterns across individuals is the fact that absolute magnitudes are meaningless because of varying skin resistances, electrode placements, etc. Amplitude normalization has been introduced to remove this measurement artifact and to allow concentration on the phasic variation of the patterns. Currently, different normalizing factors are being used by various investigators (2,12). Historically, normali-

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zation by the peak amplitude occurring or peak amplitude produced during a maximum voluntary contraction has been implemented. Yang and Winter (21) examined the effect of these and other factors on the variability of ensemble averages in an adult population. They demonstrated that normalizing a subject's LE by its pattern average produced the smallest coefficient of variation (CV) in the population ensemble averages in four of the five muscles tested.

The objectives of the investigation whose results are reported were to address two of the factors that are important for the development of a normative data base. These are the effects of normalization and the effects of walking speed on population variability. The results are presented in a standardized fashion, so that they can be used as the nucleus of a normative base for assessing pathologic synergies, and so that they can be upgraded easily as the size of the data base increases. The results are used in Part 2 of this paper to assist in the evaluation of synergy patterns in hemiparetic gait.

METHODOLOGY

Measurement

Thirty normal adults, (20 males and 10 females), whose ages ranged from 21 to 35 and averaged 29 years, and who did not have any history of neuro-musculo-skeletal disorders, participated in this study. All were asked to walk at self-selected speeds of free, fast, slow, and very-slow on a level 12-meter walkway. Bilateral foot-contact and electromyographic measurements were made while they traversed the middle 5 meters. The start and end of a measurement run were triggered by the subject passing through and interrupting a light beam being sensed by photoelectric cells. Two such light beams demarcated a 5-meter section in the middle of the walkway. For each speed category requested, the subjects selected their own speeds and practiced several traversals in order to acclimate themselves. The time to traverse the middle 5 meters was monitored. After acclimation, measurements were made during several traversals. Measurements made during traversals at the incorrect speed were ignored and the subject was simply requested to walk faster or slower, as needed. Depending on walking speed, two to four complete strides are made during a single traversal. A minimum of 10 strides with measurements was required at each speed.

The onset and offset of the heel, fifth and first metatarsals, and large toe of both feet were sensed with Tapeswitch Corporation switches taped to the bottom of the subjects' daily footwear. The contact times were measured using a B&L Engineering telemetry system. EMG signals from the tibialis anterior, gastrocnemius, soleus, rectus femoris, vastus lateralis, medial hamstring, and gluteus medius muscles were sensed with miniature Beckman surface electrodes in a bipolar arrangement with a 1-centimeter spacing. The electrodes were placed on the skin over the region presenting the largest surface exposure and aligned such that a line connecting the pair was parallel to the direction of the muscle fibers. Proper attachment and crosstalk checks were done by having the subjects voluntarily contract their muscles against limb resistance and perform test walks while the raw EMG recordings were monitored. For each leg muscle the criterion for "no crosstalk" was that baseline activity must be observed during a period of time while other muscles are active. With these precautions, crosstalk is not a problem, since it has been shown that a maximum of 10 percent crosstalk is incurred between adjacent muscles (9). The amplitude was measured using the standard differential amplifier configuration in the Biosentry telemetry system. The EMGs are bandpass-filtered to eliminate noise and motion artifacts. The filters have a fourth-order, one-half dB ripple, Chebyshev high-pass section with a cutoff frequency of 40 Hz, and a second-order Butterworth lowpass section with a cutoff frequency of 400 Hz. Linear envelopes of the EMGs are formed with a detector consisting of an active full-wave rectifier and a third-order Paynter integrator. The integrator is designed to have a bandwidth of 80 Hz and an equivalent integration time of 12.5 milliseconds. This linear detector has been shown to be a good estimator of the envelope of the nonstationary EMG.

Acquisition and Averaging

All measured signals were acquired in real time using a PDP 11/03 minicomputer. The LE were acquired from the envelope detectors described above with an ADAC A/D converter. The foot-contact sequences were acquired with a parallel I/O port through a specially built 8-bit parallel line interface. The data acquisition program sampled at 250 Hz during the measurement session and stored the data on a disk file. In a subsequent operation, signal processing software implementing a time base interpolation scheme transformed the LE for each stride into a 256 point representation. The interpolated LE from strides of similar tasks were averaged to produce the average LE. The fiducial mark for the start and end of each stride was the time of foot contact following swing. The time base is expressed as a proportion of the actual stride time, making the duration of the normalized time base equal to 1.0 or 100 percent. Refer to Shiavi and Green (16) for details.

Envelope Characteristics

Several analyses of the average linear envelopes were performed to study their characteristics and variabilities and to determine significant effects of walking speed. These analytic techniques are standard procedures in signal processing and statistics and are briefly explained first (7).

Ensemble Properties. The ensemble averages and standard deviations across the subject population describe the general properties of the LE. For any grouping of LE the ensemble average is defined as

$$M(i) = \frac{1}{N}\sum_{j=1}^{N} NLE(j, i); 0 \leq i \leq 255$$
$$NT = 1/256, \text{stride}\,\% = NT \times 100xi \quad [1]$$

where the subscript i is the time index for the stride and NT represents the normalized time sampling interval. NLE(*j,i*) represents the amplitude of the normalized linear envelope at time point i for subject j. N represents the number of subject envelopes in the ensemble and M(i), the ensemble average for time point i. The ensemble variance is defined as

$$S(i)^2 = \frac{1}{N}\sum_{j=1}^{N} (NLE(j, i) - M(i))^2; 0 \leq i \leq 255 \quad [2]$$

where S(i) represents the population standard deviation for a specific normalized time.

Variability Measures: Two statistics are used to measure the overall variability of ensembles of LE. One is the coefficient of variation (CV), since it has been presented and used in the literature. The second statistic is analogous to a common one found in signal processing applications. It is an adaptation of the noise to signal ratio and is the variation to signal ratio. They are similar measures, and after a comparison the variation to signal ratio will be the measure of variability since it is a theoretically derivable measure. The coefficient of variation for a particular time in the stride, COV(i), is defined as

$$COV(i) = \frac{S(i)}{M(i)}; 0 \leq i \leq 255 \quad [3]$$

The coefficient of variation over a full stride, CV, is defined as

$$CV = \sum_{i=0}^{255} S(i) \Big/ \sum_{i=0}^{255} M(i) \quad [4]$$

The variation to signal ratio, V/S, is the ratio of the energy in the variation over the stride to the energy in the average over the stride. It is defined as

$$V/S = \sum_{i=0}^{255} S(i)^2 \Big/ \sum_{i=0}^{255} M(i)^2 \quad [5]$$

Normalization: For each of the seven muscles studied, the population ensemble averages and standard deviations and the variability measures were calculated using two amplitude-normalizing parameters. These are the average value of the within-subject ensemble average (hereafter called the "pattern average") and the peak of the within-subject ensemble average (hereafter called the "pattern peak". The parameter producing the smaller variability will be used in the analyses.

Speed Effects

To examine the effect of walking speed on the linear envelope (LE), a one way analysis of variance (ANOVA) using the components of variance model was performed. Each LE was associated with one of the four speed categories of very-slow, slow, free, or fast in the following manner:

1. The average self-selected speed in each category was calculated;
2. A histogram of all the speeds produced was calculated;
3. The histogram was studied for naturally occurring minima in order to select boundaries for the speed catagories; and
4. Patterns were assigned to the speed category according to the subject's actual speed.

In this application, the LE values at specific times in the stride are being compared across subjects. Hence, the ANOVA is performed for each time period. Consider any quantity that is hypothesized to have influences modeled linearly. Mathematically, the model of the data is

$$Y(l, n) = U + P(l) + E(l, n);\ 1 \leq l \leq 4 \quad [6]$$

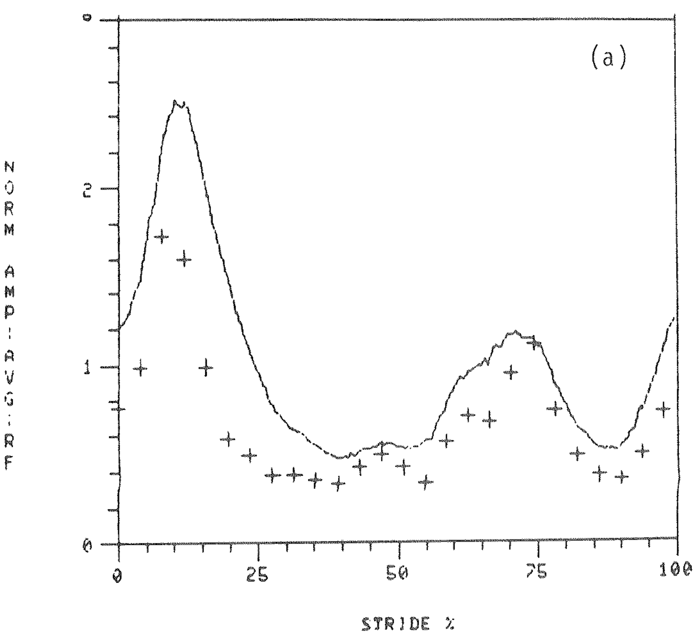
where Y(l,n) is the value of a LE for subject n in category l, U the grand mean value, P(l) the influence of speed in category l, and E(l,n) the noise or unmodeled residual. The ANOVA method calculates the within-group and between-group mean square values of the LE, and uses the F ratio to test for any significant influence of speed. Specifically, the hypothesis tested is whether or not the variance of the P(l) is zero. A 5.0 percent significance level was chosen. Directly applying the ANOVA would require performing the analysis at all 256 time points. However, since 95 percent of the energy is within a 6-Hz bandwidth, reducing the representation to 20 time periods is acceptable and valid (16). As a consequence, the analyses will be executed much faster and the results will be easier to present. The stride was divided into 5 percent periods and the average LE value over each period for each envelope was calculated. Thus the ANOVA was performed 20 times in order to determine within which stride periods there are significant differences in the values of the linear envelopes.

RESULTS

Normalization

For each muscle, the population ensemble averages produced by each of the two normalizing parameters are remarkably similar in shape; however, the standard deviations are different. **Figures 1 a and b** show the population ensemble properties for the rectus femoris muscle. Notice the trends in the standard deviation across the stride. The variability produced by pattern maximum is almost uniform whereas the variability produced by the pattern average fluctuates throughout the stride. The fluctuations correlate with amplitude and are very low during periods of quiescence and higher during periods of activity. This general observation is the same for all muscles.

The CV and V/S produced by each of the two normalization methods for all the muscles are listed in **Table 1**. Either measures yield the same relative results. The variability produced by the pattern average method is less than that produced by the pattern maximum method in the three calf muscles,



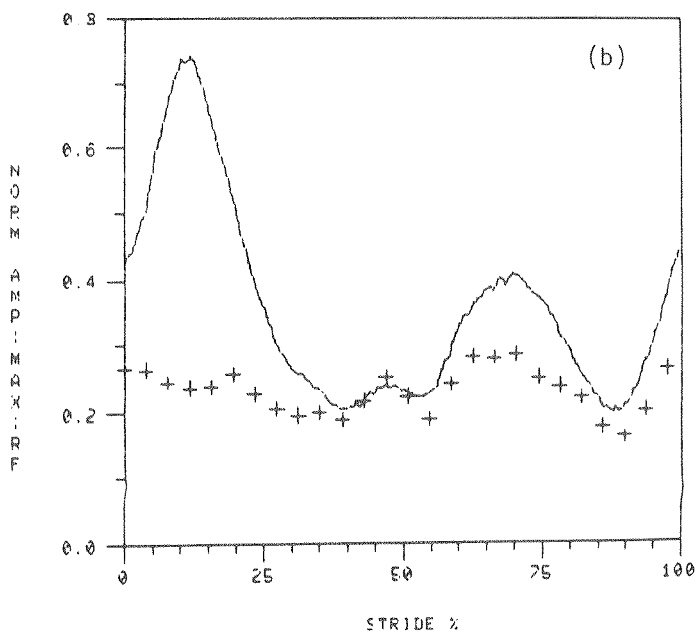


Figure 1.
The population ensemble averages (−) and standard deviations (+) for the rectus femoris muscle are plotted using normalization by the pattern average (a) and pattern maximum (b).

greater in the three thigh muscles, and equal in the gluteus medius muscle. Neither is generally better using the overall variability criteria. However, the standard deviation across the stride produced by the pattern average method is less during periods of quiescence and, therefore, better emphasizes periods of activity. For this reason, the pattern average will be the normalizing parameter. An important consequence is that the average of all normalized LE is one.

Table 1
Variability Measures for Pattern Average (AVG) and Pattern Maximum (Max) Normalization

	V/S		CV	
Muscle	AVG	MAX	AVG	MAX
tibialis anterior	.19	.22	.50	.55
gastrocnemius	.28	.35	.64	.72
soleus	.21	.23	.56	.58
rectus femoris	.44	.36	.66	.64
medial hamstring	.52	.47	.78	.75
vastus lateralis	.70	.52	.88	.79
gluteus medius	.33	.33	.58	.58

Pattern Changes with Speed

Speed Range Division: The histogram of speeds produced is shown in **Figure 2**. In **Table 2** are listed the average self-selected speeds. The only naturally occurring minimum is around 1.5 meters per second (m/s) and divides the free and fast speed categories. The remainder of the speed range is divided equally into three divisions. The resultant speed category boundaries and the average speeds within these categories are also listed in **Table 2**.

Variability Properties: The V/S ratios plotted versus average speed in each speed category for all the muscles are shown in **Figure 3**. Notice that the average speeds in a given category are not equal. This is because some LE had to be ignored because of poor measurement quality or conditions occurring during the course of the measurement session. Also, only five EMG were measured simultaneously, with the result that speeds during subsequent trials differed slightly. For all of the muscles except the soleus, the ratios have decreasing trends with respect to walking speed. In four of the muscles, such as the tibialis anterior, the trends are monotonic; whereas, in the gastrocnemius and gluteus medius, the LE in the free-speed category have greater variability than LE in the slow-speed category. The

Table 2
Speed (m/s) Categories, Averages, and Ranges

	Self-Selection	**Division**	
	Average	Range	Average
Very-slow	.59	s < 0.74	0.50
Slow	1.02	0.74 < s < 1.14	0.86
Free	1.44	1.14 < s < 1.53	1.23
Fast	1.81	s > 1.53	1.73

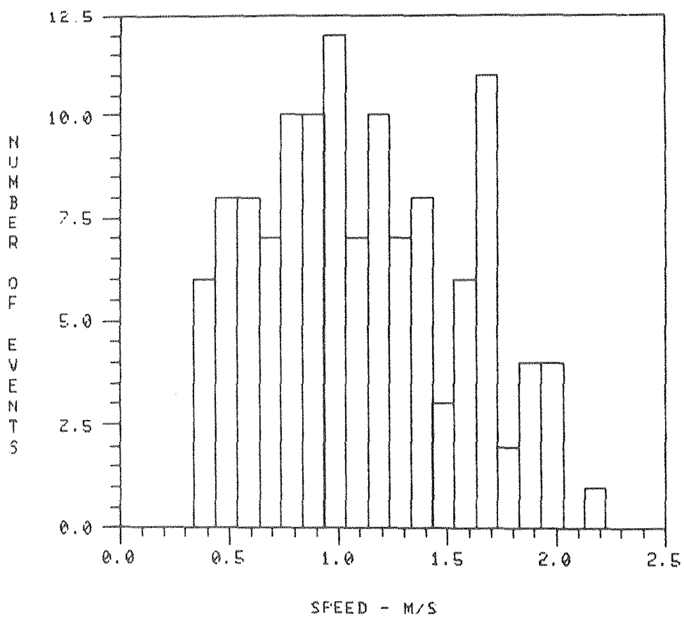


Figure 2.
The histogram of all walking speeds produced is plotted.

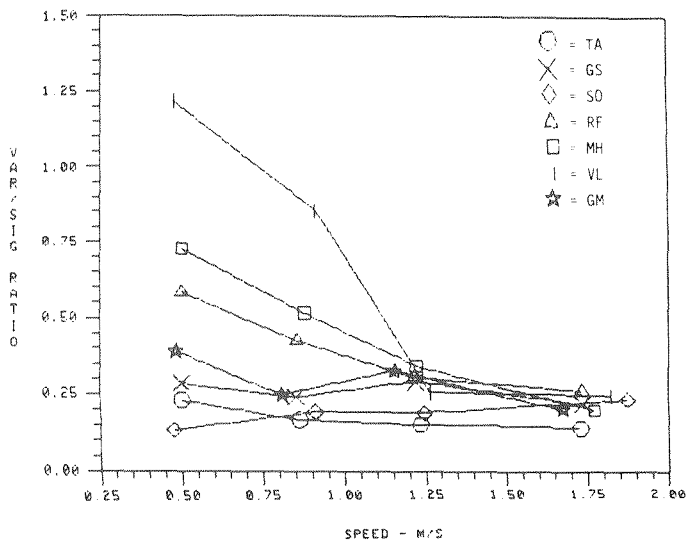


Figure 3.
The variation to signal (V/S) is plotted with respect to average walking speed in each category. Muscle symbols are: tibialis anterior, TA; soleus, SO; gastrocnemius, GS; rectus femoris, RF; medial hamstring, MH; vastus lateralis, VL; gluteus medius, GM.

correlation coefficients of V/S ratio versus speed were calculated to determine the strength of this trend. The correlation coefficients for the tibialis anterior, soleus, medial hamstring, rectus femoris, and vastus lateralis test as nonzero at the 90 percent confidence level. The coefficients for the other muscles test as zero.

Pattern Characteristics: For each muscle the ensemble average (−) and standard deviation (+) for the very-slow and fast speed categories are plotted. The ordinate magnitudes correspond to normalization by pattern average and the time scale is percent of stride. Also indicated are the periods of double leg (DL) and single leg (SLS) support, and swing (SWNG). The results of the ANOVA are presented graphically with these pattern characteristics. The periods of the stride in which the LE test as being significantly different at the 5.0 percent level because of speed are indicated by a star. Discussions are especially focused on these time periods.

The tibialis anterior muscle has the most consistent pattern. The typical biphasic pattern is always manifested (**Figure 4**). The first phase always peaks at 5 percent of the stride; whereas, the timing of the second peak decreases monotonically from 75 to 65 percent as speed increases. Two adjustments in relative intensity occur. The first is that the intensity around 20 percent of the stride is greater at slower speeds and that the intensity during the swing-to-stance transition is greater at faster speeds.

The major portion of gastrocnemius activity always occurs during terminal stance (**Figure 5**). There is a great degree of variability in magnitude and width of this phase as indicated by the large standard deviation during this period. The timing of the peak of activity decreases slightly from 45 to 40 percent of the stride as speed increases, although the amplitude is not indicated as changing significantly. Two significant changes occur that contribute to the formation of a second distinct phase during early stance at fast speed: as speed increases, the activity during the swing-to-stance transition increases, and during midstance the activity decreases relative to the major phase.

The soleus has the second most consistent pattern although strangely its V/S ratio increases with walking speed. Its main phase becomes shorter in duration and a second phase forms as walking speed increases (**Figure 6**).

The pattern of the rectus femoris is biphasic, with the amplitudes of both phases being more variable at slower speeds (**Figure 7**). As speed increases, the relative intensities of these two phases tend to become equal. A very small third phase begins to appear during terminal stance at fast speed. The timing of the peak of the first phase is always at 10 percent of the stride, whereas the timing of the peak of the second phase decreases monotonically from 75 to 65 percent of stride as speed increases.

The medial hamstring pattern is, by the average, biphasic at very-slow speeds and tends to become

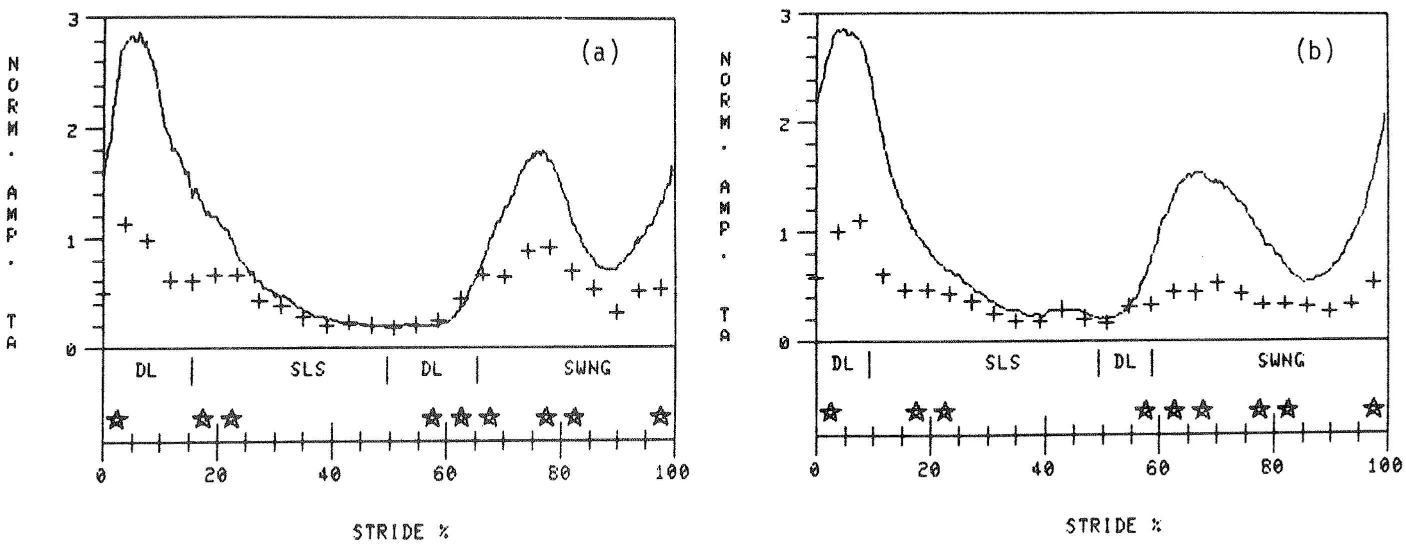


Figure 4.
Ensemble properties of the tibialis anterior muscle for very-slow (a) and fast (b) walking speeds. The average (−) and standard deviation (+) are plotted and the periods of significance are indicated by star symbols.

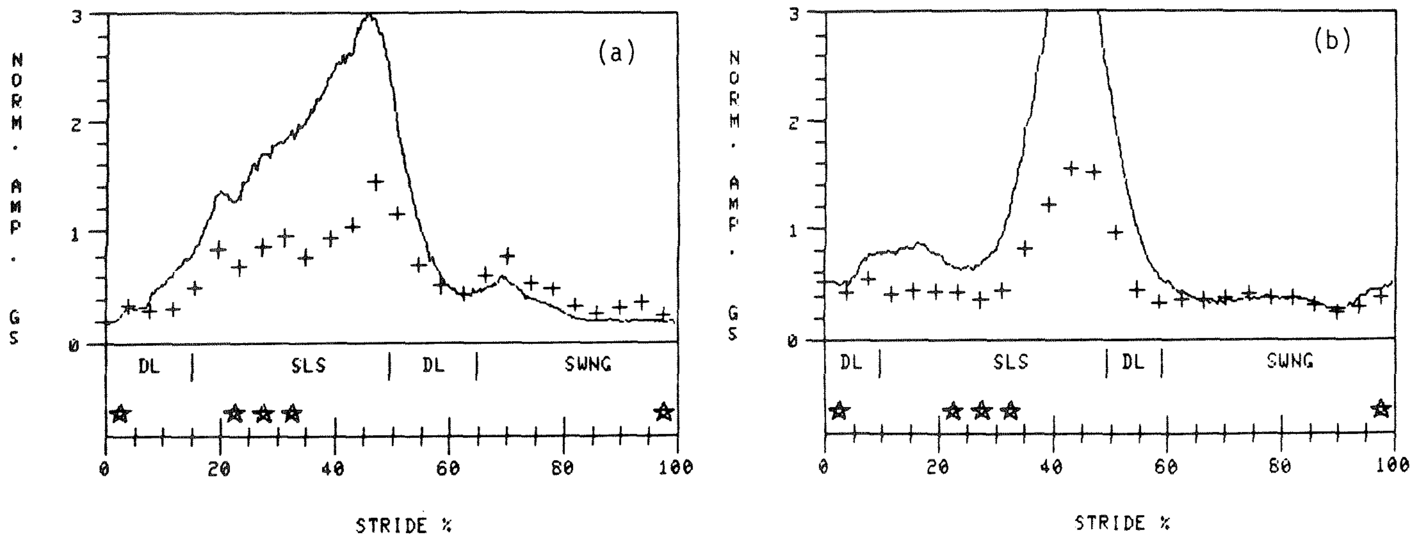


Figure 5.
Ensemble properties of the gastrocnemius muscle for very-slow (a) and fast (b) walking speeds. The average (–) and standard deviation (+) are plotted and the periods of significance are indicated by star symbols.

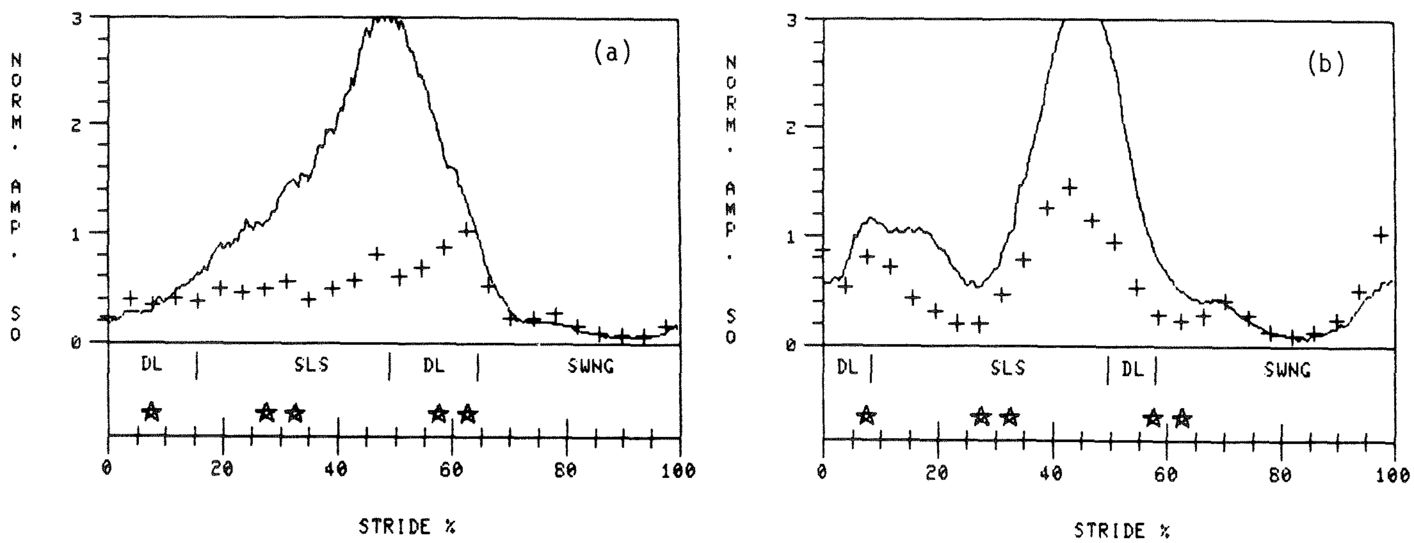


Figure 6.
Ensemble properties of the soleus muscle for very-slow (a) and fast (b) walking speeds. The average (–) and standard deviation (+) are plotted and the periods of significance are indicated by star symbols.

monophasic at fast speeds as shown in **Figure 8**. The important aspect here is the large standard deviation at very-slow speed. It indicates a spread of values so large that the second phase may or may not exist in some subjects. Perusal of the individual patterns verifies this conjecture. The swing-to-stance transition phase always exists. The timing of its peak activity changes gradually as speed increases. At very-slow and slow speeds it peaks during loading, whereas at the faster speeds it peaks during late swing.

The vastus lateralis is unquestionably the most variable of the muscles studied. It is also, on the average, biphasic at very-slow speeds and tends to become monophasic at fast speeds (**Figure 9**). The large standard deviations at very-slow speed indicate a large spread of relative intensities in both phases. Perusal of the individual patterns reveals that there is a mixture of monophasic and biphasic patterns and that the loading phase always exists. As speed increases, the loading phase becomes dominant and the monophasic pattern consistent.

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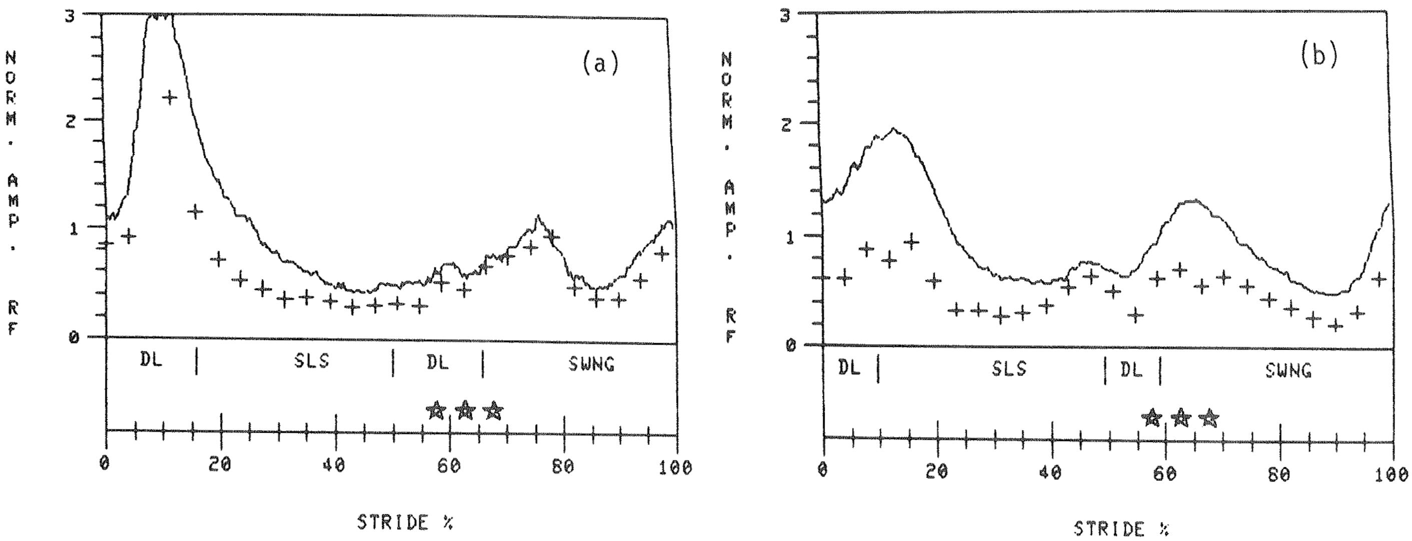

Figure 7.
Ensemble properties of the rectus femoris muscle for very-slow (a) and fast (b) walking speeds. The average (−) and standard deviation (+) are plotted and the periods of significance are indicated by star symbols.

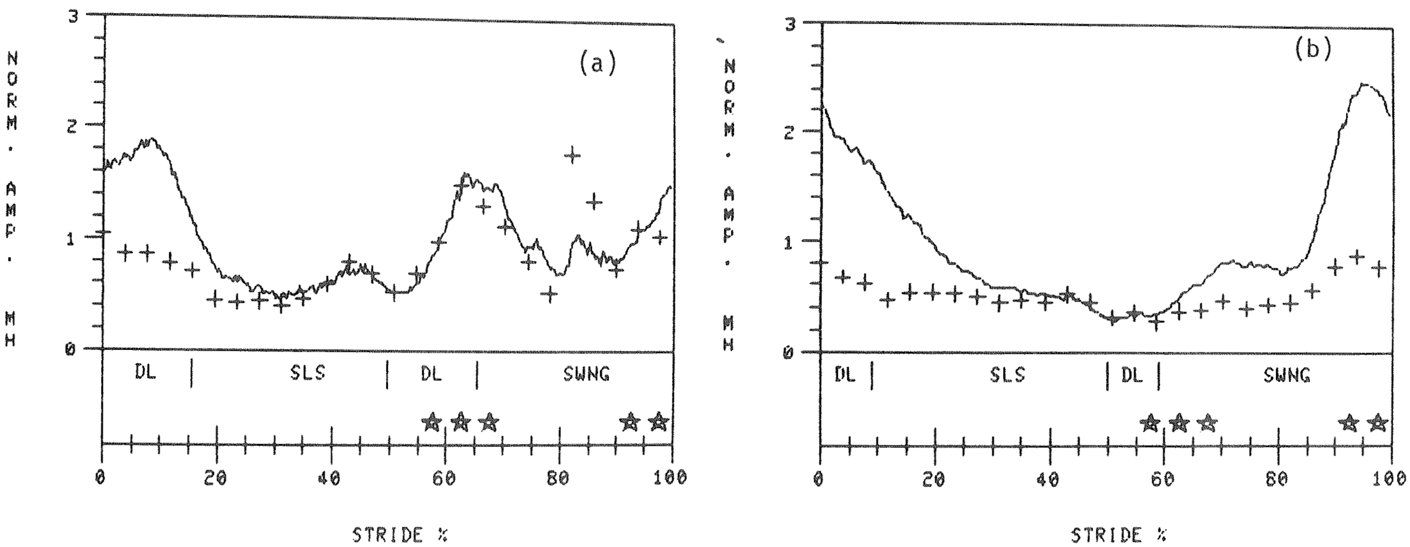

Figure 8.
Ensemble properties of the medial hamstring muscle for very-slow (a) and fast (b) walking speeds. The average (−) and standard deviation (+) are plotted and the periods of significance are indicated by star symbols.

The gluteus medius shows a lot of relative amplitude changes with walking speed (**Figure 10**). It has phases of activity during stance and swing. At slower speeds the stance phase is dominant with activity peaking around 12 percent of the stride. As speed increases, the relative intensities change in two ways: the activity during terminal stance becomes a third phase of activity, and the swing phase activity increases in intensity to be greater than that in terminal stance. The peak of activity during early stance shifts to 8 percent of the stride.

DISCUSSION

The population variability of patterns in all the muscles studied, except for the soleus, decreases as walking speed increases. In two muscles this trend is not monotonic. Nonetheless, the tendency is for the patterns to become more consistent as walking speed increases. That is, at slower speeds the muscles' patterns are much more individualized and muscles are responsive to individual motion requirements. But at faster speeds, some type of

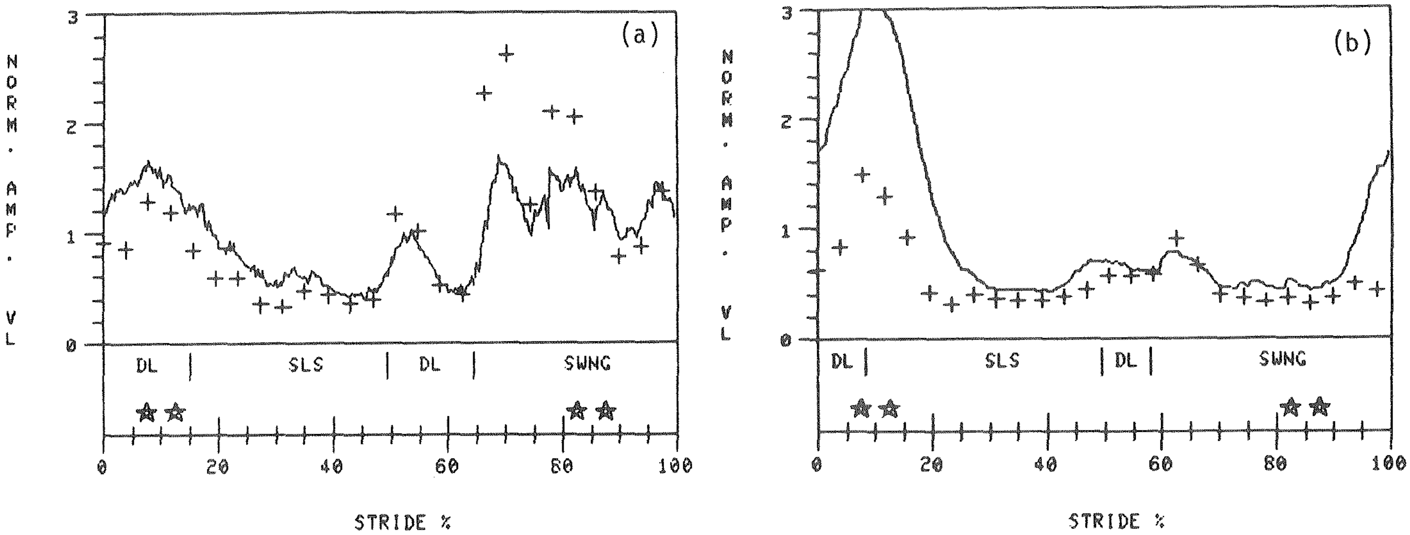

Figure 9.
Ensemble properties of the vastus lateralis muscle for very-slow (a) and fast (b) walking speeds. The average (−) and standard deviation (+) are plotted and the periods of significance are indicated by star symbols.

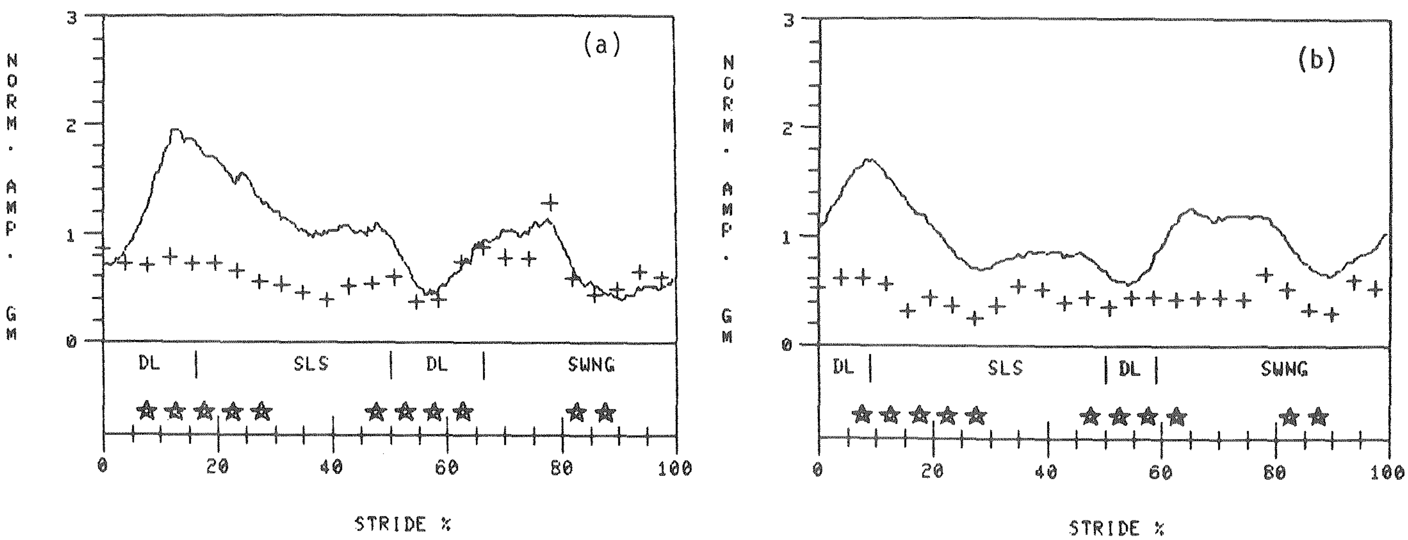

Figure 10.
Ensemble properties of the gluteus medius muscle for very-slow (a) and fast (b) walking speeds. The average (−) and standard deviation (+) are plotted and the periods of significance are indicated by star symbols.

motor programming is dominant and everyone tends to utilize the same patterns (2). There is a definite distinction between calf and thigh muscles. At very-slow and slow speeds the variability of the thigh muscles is at least twice as great as those of the calf muscles: at free and fast speeds all muscles had variabilities with the same order of magnitude. These general results are the same as those found in childhood gait (17).

An interesting problem arose when the normal subjects were requested to walk at very-slow speeds of approximately 0.5 meters per second. They acknowledged that they had lost their automaticity and had to concentrate and practice in order to walk smoothly. They could ambulate with long steps, short steps, or even pause in position during the stride. In order to maintain uniformity, the subjects were trained or coached to perform walking strides that used shorter stride times and lengths and were performed smoothly. It was impossible for anyone to walk slower than 0.34 meters per second without shuffling or pausing in motion.

The integration of these results describing the behavior of the EMG LE profiles with those previously reported is somewhat difficult because of the variety of experimental paradigms (14). This is especially true knowing the effect that EMG threshold has on the on-off pattern (9,20). However, if one studies carefully the standard deviation of the ensembles, the findings are consistent.

The variability in the EMG patterns was demonstrated by two general modes of accommodation or adaptation; the use of different phases of activity or the modulation of intensity of phases. The modulation mode is one in which the basic shape of the patterns does not change but the relative amplitudes of the phases of activity change with walking speed. The tibialis anterior, gastrocnemius, soleus, and gluteus medius muscles exhibited this type of behavior. The major implication is that the control strategy for these muscles does not change greatly but only undergoes minor modifications. Yang and Winter (21) also observed this behavior in the tibialis anterior and soleus muscles. Although Murray, et al. (11), studied trends in EMG activity with walking speed, this type of behavior could not be ascertained because of their case study approach.

Several muscles displayed additional or fewer phases of activity at different walking speeds. At slower speeds the medial hamstring and vastus lateralis had another phase of activity during stance-to-swing transition and early swing, respectively. Perhaps the reason this phase of activity is not always observed is that most investigators concentrate on activities at free and fast walking speeds. The observation that the stance-to-swing transition activity of the rectus femoris becomes more prevalent with an increase in walking speed agrees with the consensus of previous investigators (14). In addition, distinctly low- amplitude phases of activity are beginning to form at faster speeds in the gastrocnemius, soleus, and gluteus medius. The implication is that the control strategies change for different biomechanical demands.

An additional comment concerning the gluteus medius muscle is warranted. It is usually reported as a stance phase muscle. However, swing phase activity is observed and has been reported in a few other investigations (11,14).

Among the muscles whose patterns display a modulation, another behavioral feature exists: the timing of certain phases occurs earlier in relative stride time as speed increases. This is clearly demonstrable in the tibialis anterior, gastrocnemius, and rectus femoris muscles. Since it is known that stance phase decreases as speed increases (15), this argues for normalizing the stance and swing phases to predetermined percentages to further reduce variability within the population of patterns.

SUMMARY

The profiles of the linear envelopes of electromyographic patterns of seven major muscles were studied as a function of walking speed. All 30 subjects were adults and the electromyograms were measured with surface electrodes. The speeds ranged from 0.34 to 2.13 meters per second. The ensemble statistical properties were used to quantitate the characteristics of the patterns. Two general types of pattern changes were observed: one was that the fundamental phasing of muscle activity never changed but that relative amplitude of the phases was modulated as speed increased; the second was that different phases of activity existed for different walking speeds. The timing of most phases as expressed in percentage of the stride decreased as speed increased. This suggests that the time base should be further normalized by stance and swing phases.

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