

Effects of muscle kinematics on surface EMG amplitude and frequency during fatiguing dynamic contractions

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Potvin, J. R. Effects of muscle kinematics on surface EMG amplitude and frequency during fatiguing dynamic contractions. *J. Appl. Physiol.* 82(1): 144–151, 1997.—Fifteen male subjects performed a repetitive elbow flexion/extension task with a 7-kg mass until exhaustion. Average joint angle, angular velocity, and biceps brachii surface electromyographic (EMG) amplitude (aEMG) and mean power frequency (MPF) were calculated with each consecutive 250-ms segment of data during the entire trial. Data were separated into concentric or eccentric phases and into seven 20°-ranges from 0 to 140° of elbow flexion. A regression analysis was used to estimate the rested and fatigued aEMG and MPF values. aEMG values were expressed as a percentage of amplitudes from maximum voluntary contractions (MVC). Under rested dynamic conditions, the average concentric aEMG amplitude was 10% MVC higher than average eccentric values. Rested MPF values were similar for concentric and eccentric phases, although values increased ~20 Hz from the most extended to flexed joint angles. Fatigue resulted in an average increase in concentric and eccentric aEMG of 35 and 10% MVC, respectively. The largest concentric aEMG increases (up to 58% MVC) were observed at higher joint velocities, whereas eccentric increases appeared to be related to decreases in velocity. Fatigue had a similar effect on MPF during both concentric and eccentric phases. Larger MPF decreases were observed at shorter muscle lengths such that values within each angle range were very similar by the end of the trial. It was hypothesized that this finding may reflect a biological minimum in conduction velocity before propagation failure occurs.

biceps brachii; power spectrum; concentric; eccentric; electromyography

SURFACE ELECTROMYOGRAPHY (sEMG) has been studied extensively to determine its relationship with muscle fatigue. However, with relatively few exceptions, studies using sEMG for this purpose have done so under isometric conditions where muscle length, velocity, and force were strictly controlled. Fatigue is generally accompanied by increases in sEMG amplitude (8, 31) and shifts in the sEMG spectrum to lower frequencies (spectral compression) (17, 31, 36) during prolonged submaximal contractions. It has been proposed that muscles must respond to fatigue with increased spatial or temporal motor unit recruitment (25, 38) and/or synchronization (28, 29) to compensate for decreases in force capacity and that this response accounts for the observed increases in sEMG amplitude. The spectral compression associated with fatigue is influenced predominantly by decreases in action potential conduction velocity (4, 32).

The interpretation of sEMG signals from dynamic contractions is much more difficult because movement introduces additional factors that affect their character-

istics. Muscle force capacity is highly dependent on fiber length and is also inversely related to shortening (concentric) velocity (15, 39) and directly related to lengthening (eccentric) velocity (16, 19). The sEMG frequency spectrum is also related to muscle length because shifts to higher frequencies have been demonstrated when length is decreased (2, 24). Few studies have analyzed the effects of muscle kinematics on sEMG spectrum characteristics during dynamic contractions. However, some efforts have been made to quantify fatigue during prolonged dynamic movements. These studies, like those using isometric contractions, have demonstrated increases in sEMG amplitude (13, 30, 34) and spectral compression (6, 9, 13) during concentric contractions. However, Tesch et al. (34) found no changes in the sEMG-to-torque ratio or spectral characteristics during repetitive eccentric contractions. Doud and Walsh (9) monitored biceps brachii sEMG and muscle length during slow repetitive concentric contractions (elbow flexion). In subjects under rested conditions, they observed that the relationship between muscle length and sEMG frequency was consistent with isometric studies (2, 24). However, they also found that fatigue-related shifts to lower frequencies were more pronounced at shorter muscle lengths.

sEMG signals have proven to be useful for quantifying muscle fatigue during prolonged isometric contractions. However, relatively little is known about the effects of changing muscle activity, length, and velocity during dynamic movements on the interpretation of sEMG statistics. The purpose of this study was to evaluate the effects of muscle kinematics on biceps brachii sEMG amplitude (aEMG) and frequency characteristics during a repetitive elbow flexion/extension task. Specifically, this study was designed to determine the feasibility and limitations of using sEMG to monitor muscle fatigue during unconstrained, dynamic contractions.

METHODS

Experimental protocol. Fifteen healthy male subjects volunteered to participate in the study. Subjects had an average age of 24.0 ± 3.0 (SD) yr, mass of 73.6 ± 13.9 kg, and height of 1.75 ± 0.04 m. Each subject performed three isometric maximum voluntary contractions (MVC) of the biceps brachii with their right arm restrained and elbow at a 90° angle. After 2 min of rest, subjects got into a standing posture and initiated repetitive flexion/extension movements of the elbow with a handheld load of 7 kg. The rotations ranged from full extension (elbow angle of 0° with the forearm vertical) to full flexion (elbow angle ~140°). Subjects were free to select the cycle frequency, and they were only instructed to 1) maintain the upper arm in a vertical orientation and 2) keep the movements as smooth as possible. Subjects were encouraged

to continue the flexion/extension rotations until they did not feel they could achieve the required movement range or until they experienced unacceptable muscle discomfort.

Data collection. sEMG signals were recorded from the biceps brachii of each subject's right arm. The skin was abraded and cleaned, and bipolar electrodes were fastened over the belly of the muscle with a center-to-center distance of 3 cm. A ground electrode was placed on the right acromion process. The signals were analog processed with a differential amplifier (bandwidth = 15–450 Hz, input impedance = 10 G Ω , Common Mode Rejection Ratio = 130 dB at 60 Hz, and gain = 1,000). A potentiometer was positioned over the elbow and fastened to the forearm and upper arm to monitor the elbow joint angle. The potentiometer was calibrated at 0 and 90° before its application. For each task, the sEMG and potentiometer signals were analog-digital converted (1,024 Hz) on an IBM personal computer with LabVIEW software (National Instruments, Austin, TX). Consecutive EMG and potentiometer data segments of 250 ms (256 samples) were collected, processed, and saved continuously with a double-buffered acquisition system throughout the MVC and fatiguing dynamic contractions.

Data analysis. The average EMG amplitude (aEMG) was calculated with the rectified signal from each 250-ms MVC and dynamic segment. The maximum sEMG amplitude was determined from the MVC contractions. The aEMG was normalized as a percentage of the maximum value (%MVC). For each dynamic segment of EMG data, a fast Fourier transform was performed, and the mean power frequency (MPF) was calculated as the frequency centroid of the spectrum. The potentiometer data were used to calculate the average elbow joint angle (°), and a linear regression was used to determine the average joint angular velocity (AV) as the slope of the joint angles.

The number of dynamic contractions and the total duration of the dynamic trial were recorded for each subject. All trial durations were normalized to the endurance limit. Data segments with average AV magnitudes between -5 and $5^\circ/s$ were removed so that contractions close to isometric were not included in the analysis. The dependent variable data (MPF, aEMG, and AV) were sorted into two velocity directions [concentric (AV_{con}) and eccentric (AV_{ecc})] with seven angle ranges within each direction (from 0–20° to 120–140° in increments of 20°) for a total of 14 groups. Second-order polynomial regression analyses, with normalized time as the independent variable, were used to characterize the time history of each dependent variable throughout the dynamic trial for each combination of velocity direction and angle range. For example, a second-order equation was calculated for each subject to represent the MPF time history for all observations when there was a concentric contraction and the elbow joint angle was between 80 and 100°. For the MPF data, segments with average aEMG levels <5% MVC were removed from the analysis because they were found to be unreliable for characterizing the frequency content of the EMG signals. The intercepts from these regression models were used as an estimate of the rested dependent-variable magnitudes for each velocity-joint angle combination. For each regression line, the intercept was subtracted from the value at the fatigue limit, and this dependent variable was used to represent the fatigue-induced change in MPF, aEMG, and AV over the course of each subject's dynamic trial. The second-order nature of the regression equations allowed for a determination of the curvilinear time history of changes throughout the trials. The models for each individual subject were used to estimate the MPF at the start and at every subsequent 1% interval of the dynamic trial duration. Group

averages were calculated at each interval to determine the average time history of changes.

Statistical analysis. Two-way repeated-measures analyses of variance were used to determine the effects of velocity direction (concentric and eccentric) and joint angle range (from 0–20° to 120–140°) on the intercepts and changes of the MPF, aEMG, and AV dependent variables. Orthogonal means comparisons were used to determine the significance of differences among individual means when significant main or interaction effects were observed. Comparisons were made between concentric and eccentric values at each joint angle range, and the values from the 0–20, 60–80, and 120–140° angle ranges were compared with each angle range, within both the concentric and eccentric velocity directions. Significance was set at $P < 0.05$.

RESULTS

The average number (\pm SD) of elbow flexion/extension cycles was 47.9 ± 18.7 , the average duration of the dynamic trials was 155.3 ± 64.0 s, and the average cycle frequency was 19.1 ± 4.1 cycles/min. Typical raw EMG time histories and spectra are presented in Fig. 1 from rested and fatigued phases, respectively. An example of the reduced kinematic and EMG data from three consecutive flexion/extension cycles is presented in Fig. 2.

Rested conditions. The average rested concentric amplitudes (22.8% MVC) were significantly higher ($P < 0.01$) than the rested eccentric values (12.8% MVC) (Fig. 3, A and B). Angle range was found to have a different effect on concentric and eccentric aEMG (aEMG_{con}, aEMG_{ecc}, respectively; $P < 0.01$). There was a progressive increase in aEMG_{con} with increased angle range. The 0–20° values were lower than all ranges $>80^\circ$ ($P < 0.05$), and the 120–140° aEMG_{con} were higher than those in all other ranges ($P < 0.05$) (Fig. 3A). The aEMG_{ecc} was unaffected by joint angle (Fig. 3B) ($P > 0.5$).

There was a progressive increase in concentric MPF (MPF_{con}) with increased angle (decreased muscle length) (Fig. 3C). The 0–20° MPF_{con} values were lower than those of all other ranges ($P < 0.05$), and the 120–140° MPF_{con} values were higher than those for all other ranges ($P < 0.05$) (Fig. 3C). Joint angle had a somewhat different effect on the eccentric MPF (MPF_{ecc}; $P < 0.01$). There were no differences between MPF_{ecc} values for the 0–20°, 20–40°, and 40–60° ranges and then a progressive increase with further elbow flexion. The 120–140° MPF_{ecc} values were higher than all other ranges ($P < 0.05$) (Fig. 3D).

The mean rested AV_{con} was observed to increase from low levels at full extension to a peak of 130.3 ± 11.2 (SE) %/s in the 60–80° range and then decrease again to low levels at full flexion (Fig. 4). A similar pattern was followed during the eccentric phases, although the peak AV_{ecc} magnitude of $123.0 \pm 9.3^\circ/s$ occurred in the 80–100° range (Fig. 4B). The AV in the 0–20° and 120–140° ranges were overestimations of the real values because segments with AV magnitudes $<5^\circ/s$ were not analyzed, and most of these segments occurred in the end ranges of motion.

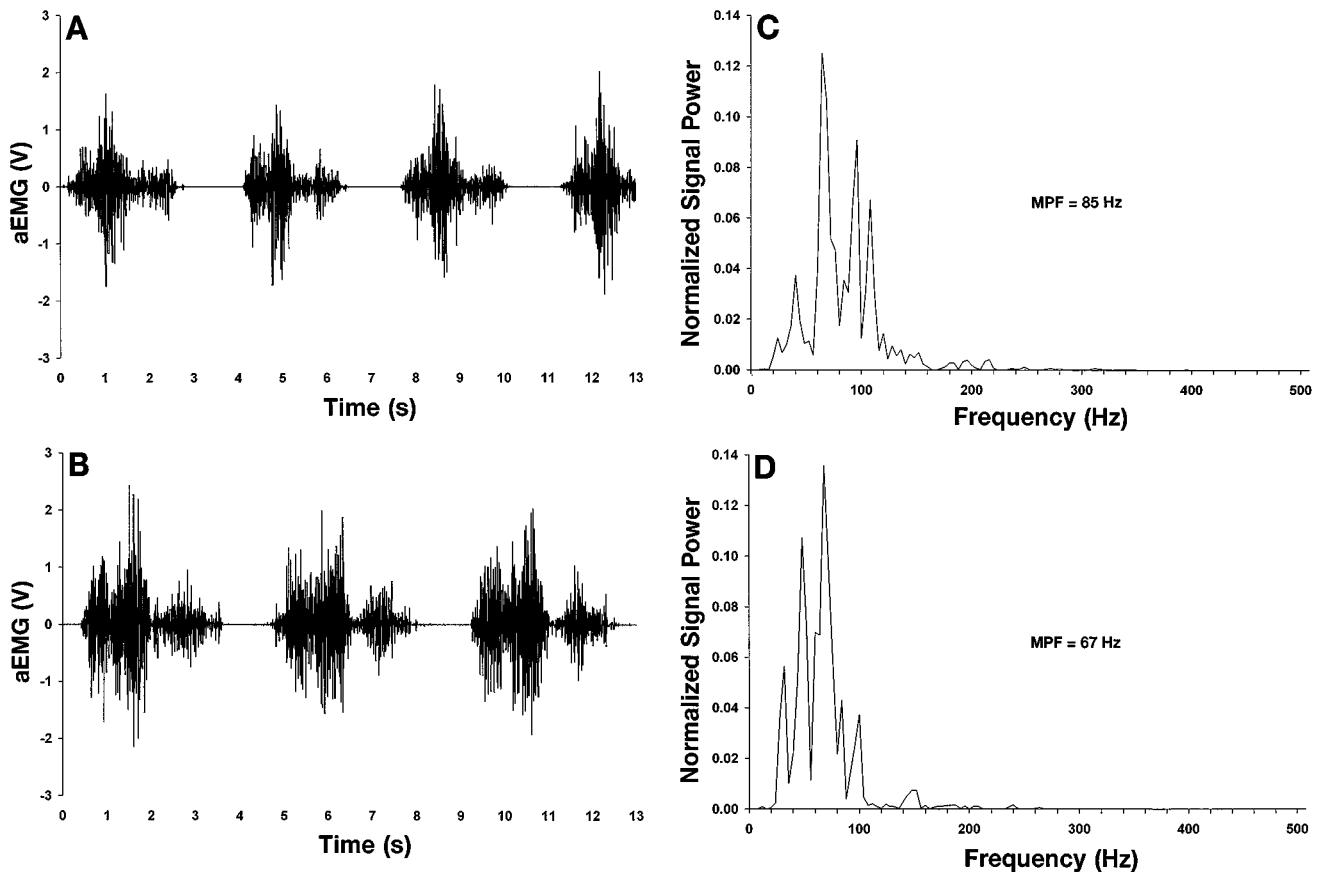


Fig. 1. Typical raw EMG time histories during 13 s of rested (A) and fatigued (B) phases of trial. C and D: EMG power spectra for rested and fatigued phases, respectively, for 250-ms segments taken during 2nd cycles of concentric contraction between 80 and 100°. MPF, mean power frequency; aEMG, biceps brachii surface EMG amplitude.

Fatigue-related changes. Fatigue resulted in significant increases in aEMG ($P < 0.01$), although the nature of these changes was different across joint angles ($P < 0.01$) and between the concentric and eccentric phases ($P < 0.01$). The average increases in aEMG_{con} and aEMG_{ecc} at the endurance limit were 34.6% MVC (158% of the rested value) and 10.5% MVC (84%), respectively. The largest average increase in aEMG_{con} was 53.8% MVC in the 40–60° range, and the smallest increase was only 5.9% MVC in the most flexed range (Fig. 3C). Changes in aEMG_{ecc} were <6% MVC for angles <60°, and the largest increase of 20.7% MVC occurred near full flexion (Fig. 3D).

No significant differences were observed between the MPF changes during the concentric and eccentric phases. When data were pooled, fatigue resulted in MPF decreases in each angle range ($P < 0.01$), and the magnitude of this decrease became progressively larger as flexion angle increased ($P < 0.01$). The decreases at the endurance limit ranged from 10.6 Hz (17.3% of the rested value) to 28.5 Hz (35.0%) as elbow flexion angle increased from the 0–20° to the 120–140° range. The MPF changes in the 0–20° range were significantly lower than in all ranges >60°, and those in the 120–140° range were higher than those in all other ranges (both $P < 0.05$) (Fig. 3, C and D). Figure 5 illustrates the average polynomial regression curves, pooled across

subjects, for the MPF_{con} within each angle range. The joint angle effect on MPF_{con}, observed under rested conditions, was greatly diminished as fatigue progresses during the trial.

The average decrease in AV_{con} (39.5°/s) was significantly larger than the decrease in AV_{ecc} (13.5°/s; $P < 0.01$). The AV_{con} decreases were at least 19.5°/s for all angles <120°, with the largest decrease being 66.9°/s in the 80–100° range (Fig. 4). The AV_{ecc} did not change for angles <60°, and the largest decrease of 32.7°/s was observed in the 80–100° range (Fig. 4B).

DISCUSSION

The main finding of this study was that muscle length and velocity interact to affect the magnitude of fatigue-related changes in biceps brachii sEMG during repetitive flexion/extension movements. Fatigue was observed to have a larger effect on force-generating capacity at high concentric velocities while having only a small effect during all eccentric muscle actions. The effect of fatigue on MPF was observed to be similar for concentric and eccentric phases of the task. The effect of muscle length on MPF was prominent under rested conditions but diminished as fatigue progressed.

Activation under rested conditions. The observation that rested aEMG_{con} values were significantly higher

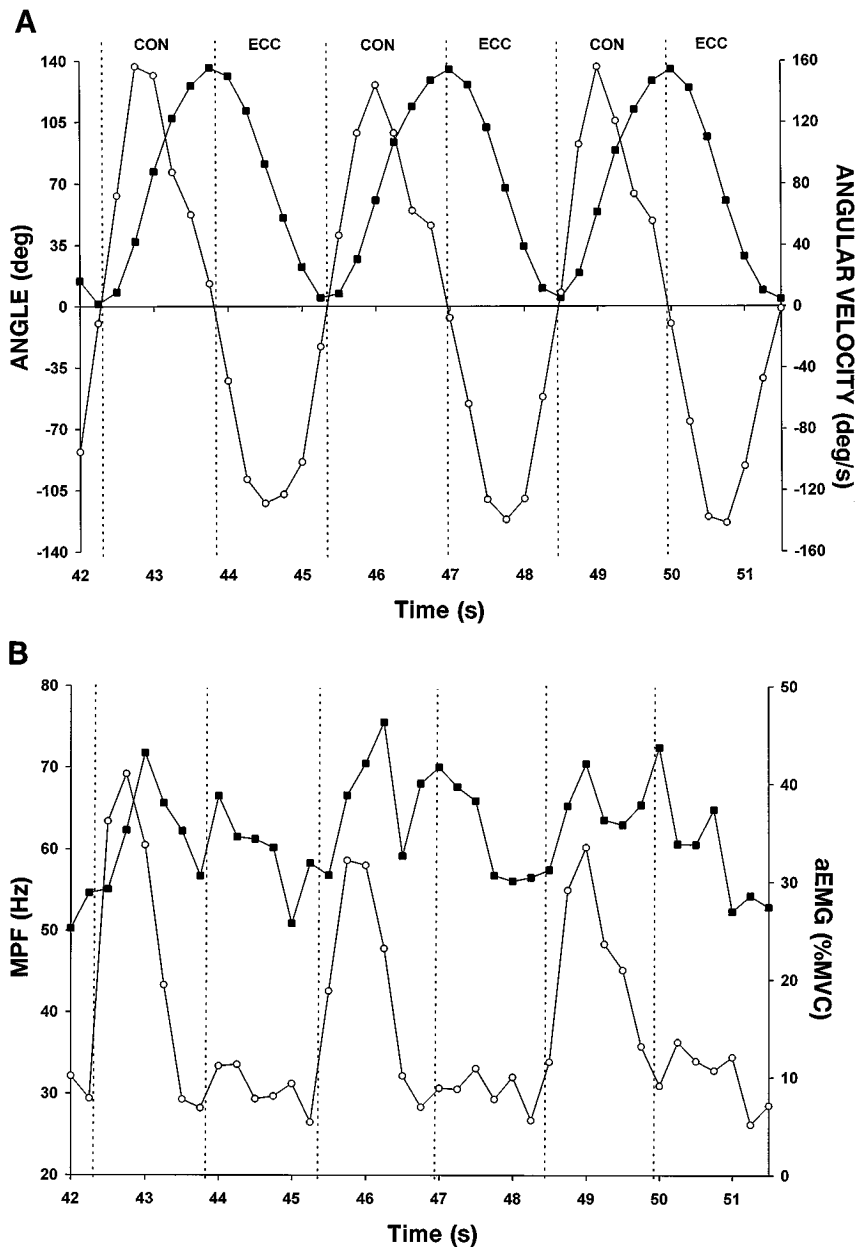


Fig. 2. Example of reduced kinematics and EMG data from 3 consecutive flexion/extension cycles. Each point represents average of 250-ms segment of data. Concentric (Con) and eccentric (Ecc) phases are indicated between vertical lines. *A*: kinematic data with elbow joint angle (■) and elbow joint angular velocity (○). *B*: EMG data with MPF (■) and aEMG (○). MVC, maximal voluntary contraction.

than eccentric values (Fig. 3, *A* and *B*, respectively) was consistent with previous dynamic studies (19, 24). These differences were due to the increased capacity to generate force during muscle lengthening (16, 19) and decreased capacity as shortening velocity increases (15, 39). Joint angle was shown to have no effect on aEMG during the eccentric actions while causing some increase in aEMG_{con} in the more flexed ranges. Given the nature of the task, the required activation at any time would have been dependent on a complex combination of the load moment, muscle length, velocity magnitude, and velocity direction. It appears that this combination resulted in relatively constant activation levels throughout the full range of eccentric actions and for most of the concentric ranges. The higher aEMG_{con} levels observed in the most flexed postures were likely due to the decrease in the contractile component strength (12)

and elimination of the passive elastic contributions (22) that would have been associated with the shorter muscle length.

MPF under rested conditions. MPF values were observed to increase significantly with decreased biceps brachii muscle length (increased flexion) for both the rested concentric and eccentric phases (Fig. 3, *C* and *D*, respectively). These results were consistent with previous studies of concentric biceps brachii contractions (9) and isometric contractions of the biceps brachii (2, 24). The frequency characteristics of sEMG are highly dependent on the conduction velocity of action potentials along the muscle fibers (5, 32). Muscle diameter increases at shorter lengths, assuming constant volume (1, 10), and muscle diameter have been shown to be directly related to conduction velocity (14, 18). Therefore, some authors have proposed that the higher

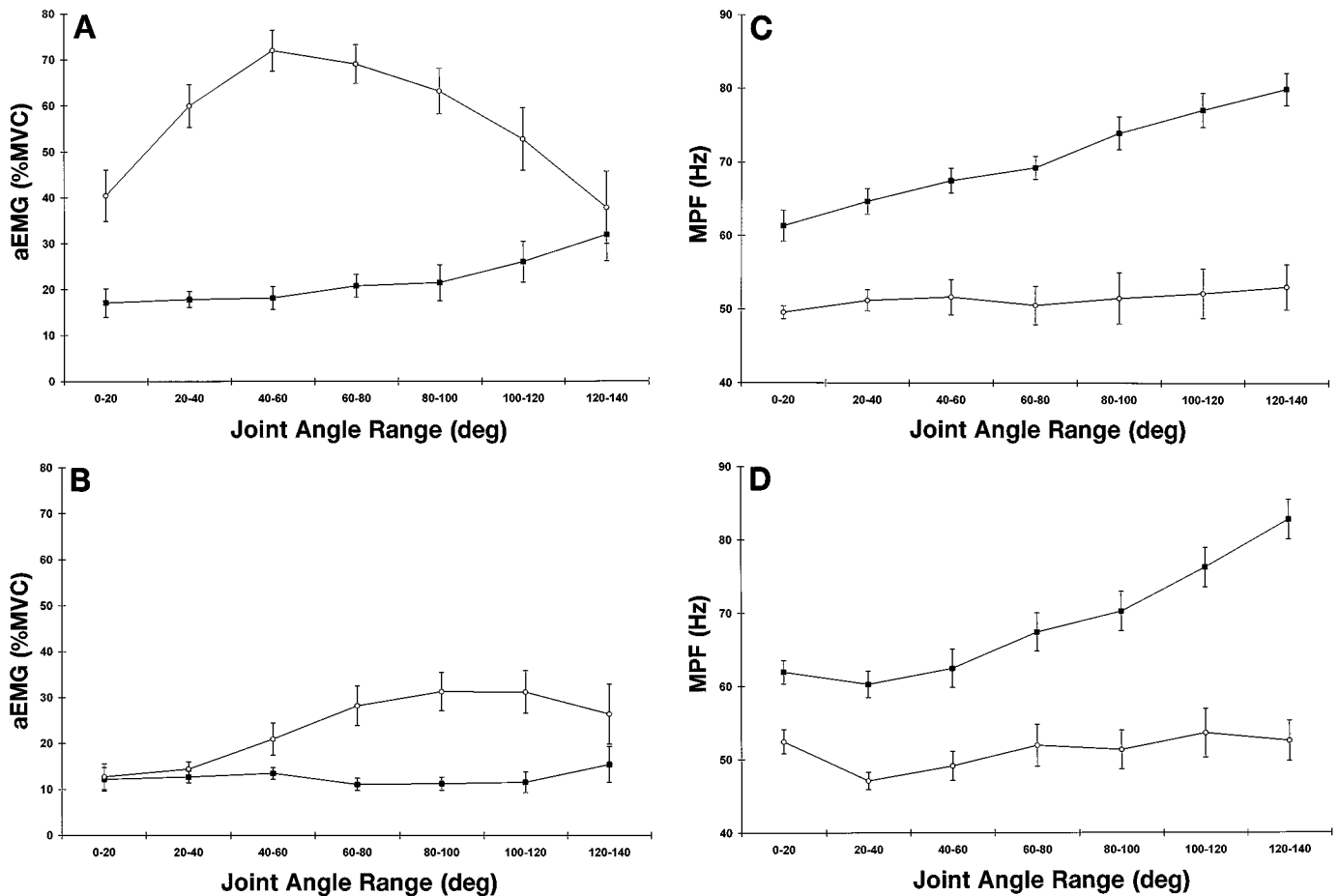


Fig. 3. Biceps brachii EMG data under rested (■) and fatigued (○) conditions for each of 7 angle ranges. A: aEMG during Con. B: aEMG during Ecc. C: MPF values during Con. D: MPF values during Ecc. Values are means \pm SE for each angle range ($n = 15$ subjects).

MPF values associated with increased elbow flexion result from shorter muscle lengths having higher conduction velocities (1, 33). The present results are consistent with the proposed relationship among muscle length, conduction velocity, and MPF through the entire range of elbow flexion.

The observation that rested MPF_{ecc} were slightly lower than MPF_{con} for three of the seven angle ranges may have been due, in part, to the lower eccentric activation levels (Fig. 3, C and D). However, the average rested MPF_{con} were only 1.8 Hz (3%) higher than MPF_{ecc} , and the highest difference within any angle range was only 5.1 Hz (8%). Previously, no difference was found between biceps brachii MPF_{con} and MPF_{ecc} values when the elbow flexion was near 90° (6, 24), although MPF_{con} has been observed to be 11 Hz higher than MPF_{ecc} in more flexed postures (24). In the present study, the largest concentric/eccentric differences were observed to occur in the more extended positions. The lower activation levels observed during the eccentric phases would be expected to result in relatively lower MPFs. However, it has been suggested that eccentric actions are associated with a preferential recruitment of fast-twitch fibers (26, 27) that results in increased MPF (20). This may account for the similar length effects on MPF_{con} and MPF_{ecc} , even though

$aEMG_{ecc}$ levels were significantly lower than $aEMG_{con}$ at all angles $>20^\circ$.

Fatigue-related changes in activation. The fatigue-related increases in $aEMG_{con}$ and $aEMG_{ecc}$ were consistent with results from previous repetitive dynamic studies of the biceps brachii (13) and other muscles (30, 34). Similar findings have also been reported in numerous studies using prolonged isometric contractions (8, 31). These aEMG increases have generally been attributed to increased motor unit recruitment (25, 38) and/or synchronization (28, 29). Increased recruitment was considered to be a likely explanation for the present results because the initial activation levels were low and the biceps brachii has been shown to recruit new units up to 85% of maximum force (21).

The increases in $aEMG_{ecc}$ were more pronounced in the higher flexion ranges (Fig. 3B). Tesch et al. (34) observed no changes in the sEMG/torque ratios during repetitive isokinetic eccentric actions of the biceps brachii. This may mean that fatigue does not result in direct increases in $aEMG_{ecc}$. However, in the present study, fatigue was associated with decreased extension velocity magnitude for the same angle ranges that demonstrated increases in $aEMG_{ecc}$ (Figs. 3B and 4B). With angles between 60 and 140°, the decreased lengthening velocity of the biceps brachii would have been

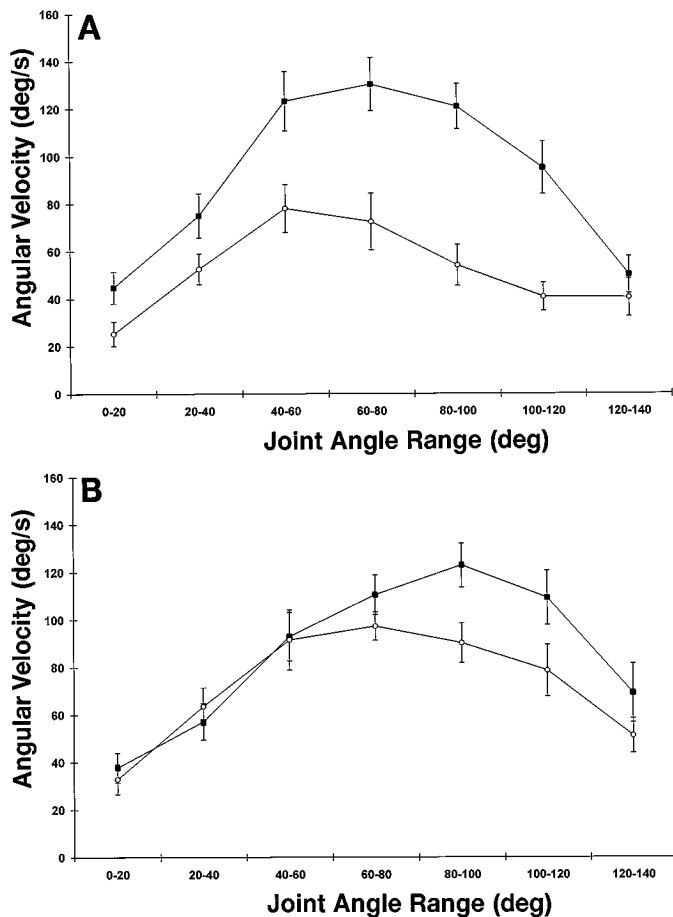


Fig. 4. Elbow joint angular velocities under rested (■) and fatigued (○) conditions for each of 7 angle ranges during Con (A) and Ecc (B). Values are means \pm SE for each angle range ($n = 15$ subjects).

associated with decreased eccentric strength (16, 19) and a subsequent need for increased activation levels. This velocity effect was not observed by Tesch et al. (34) because AV was set to be constant at $180^\circ/s$.

The increases in $aEMG_{con}$ from rest were as high as 54% MVC and significantly larger than eccentric changes, despite substantial decreases in concentric velocity that should have contributed to enhanced force capacity (15, 39). The strength-reducing effects of fatigue may have dominated the strength-enhancing velocity effect. However, it is not fully understood why the concentric $aEMG$ increases were so much larger than eccentric increases. It may be that the effects of fatigue on the excitation/contraction-coupling process are more pronounced when the muscle shortens. The observation that concentric $aEMG$ increases were highest in the ranges where joint angle was changing most rapidly (Figs. 3A, 4A) may also indicate that the peripheral effects of fatigue on force capacity are more prominent at higher shortening velocities. This proposed mechanism is supported by the data of Beelan and Sargeant (3), who observed larger relative decreases in cycling power output as velocity was increased. They hypothesize that this effect may be due to a selective fatigue of the fast-twitch fibers that are more susceptible to fatigue.

Fatigue-related changes in MPF. The decrease in sEMG MPF observed throughout the dynamic trials (Figs. 3, C and D, and 5) was consistent with previous dynamic studies of the biceps brachii (6, 9, 13). The MPF decrease associated with fatigue may be due to a temporal elongation of the motor unit action-potential waveform due to increased lactate levels (35) and/or impairment of the ionic pump (23) and a subsequent decrease in muscle fiber conduction velocities (4, 32). However, it appears that MPF decreases may not be entirely explained by conduction velocity decreases (4, 32) and that subsequent MPF recoveries are more rapid than lactate removal (30, 37). Tesch et al. (34) observed MPF decreases during concentric trials, but no changes were observed during repeated eccentric muscle actions. They stated that this difference may have resulted from an absence of fatigue during the eccentric trial. In contrast, the present concentric and eccentric phases demonstrated similar fatigue-induced decreases within each angle range. Both concentric and eccentric actions were performed within the same session, and the level of fatigue would have been very similar during each cycle. These results demonstrate that MPF will decrease during eccentric muscle actions if the muscle is sufficiently fatigued.

One of the most interesting findings of this study was that the effect of length on MPF was progressively diminished over the course of the fatiguing trials. This

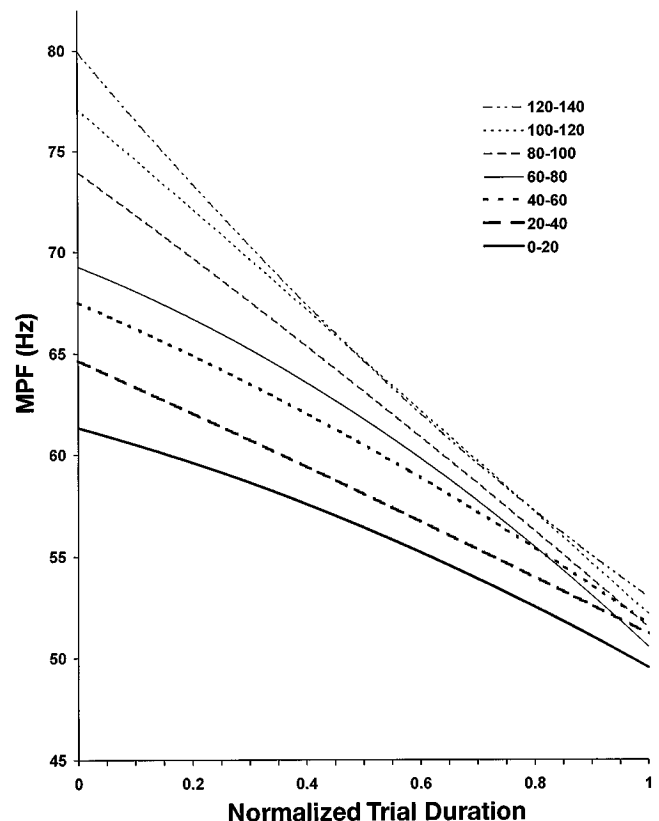


Fig. 5. Average time history of concentric MPF changes during dynamic trials. Time is normalized to trial durations. Averages were calculated with regression curves of each subject ($n = 15$) pooled within angle ranges at start and every 1% of trial duration.

finding was consistent with those of Doud and Walsh (9), who also used a regression-based method to monitor MPF changes in the biceps brachii during repeated slow (30–35°/s) controlled concentric contractions. They observed that the rates of decrease in MPF were similar at each length for the first 75% of the trials and that the shorter lengths had proportionately larger decreases only in the last 25% of the trial duration. In contrast, Fig. 5 demonstrates that the trends in the present study were essentially linear within each angle range and that the slope of these trends became more negative as joint angle decreased. These differences may have been due to the substantially larger AV used in the present study.

Under rested conditions, increases in elbow joint angle from full extension to maximum flexion caused average MPF_{con} and MPF_{ecc} increases of 19 and 22 Hz, respectively. However, fatigue caused larger MPF decreases in the higher flexion ranges (Fig. 3, C and D), and the MPFs were observed to be relatively independent of joint angle at the end of the trials (Figs. 3, C and D, and 5). Under fatigued conditions, the average MPF_{con} and MPF_{ecc} differences between the 0–20° and 120–140° angle ranges were only 3.5 and 0.1 Hz, respectively. This convergence of MPF values with progressive muscle fatigue may indicate that there is some biological minimum for the action potential conduction velocities that dominate the sEMG spectrum. Recent data from Cupido et al. (7) support this hypothesis. They stimulated the biceps brachii at 20 Hz and concluded that propagation failure would occur beyond the point where conduction velocity had decreased ~50%. This decrease in conduction velocity appears to result from a slowing or complete impairment of fast-twitch motor unit action potential transmission (11). With long fiber lengths, the conduction velocity would already be relatively slow such that fatigue would only cause small decreases in velocity before transmission failure. However, in shorter fibers the initial conduction velocities would be much higher and could experience larger fatigue-induced decreases before conduction would no longer be possible.

Summary. Under rested conditions, the concentric sEMG amplitudes were observed to be higher than eccentric values due to the influence of velocity on force-generating capacity. MPF values were similar for concentric and eccentric phases of the movement, although values increased with decreased muscle length. This finding was attributed to the increase in action potential conduction velocity that results when a muscle becomes shorter and its diameter increases. Fatigue resulted in an increase in the concentric sEMG amplitudes, although this increase was not uniform across elbow joint angles. Larger increases were observed in the midrange of concentric movement, and it was hypothesized that this was related to the higher velocity magnitudes that occurred in this range. There appears to be a disproportionate decrease in concentric force capacity with increased velocity and a subsequent need for higher activation. No concentric/eccentric differences were observed in the fatigue-induced changes

in MPF. Larger decreases were observed at shorter muscle lengths such that final values fell within a small range. It was hypothesized that this finding may reflect a biological minimum in conduction velocity before propagation failure occurs.

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