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CHAPTER 16

Evolving characteristics of the median frequency of the EMG signal

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INTRODUCTION

In the laboratory and clinical setting, it is often useful and necessary to evaluate the functional capability of a muscle or group of muscles in situ. This may be accomplished by measuring the rate at which the muscle performance deteriorates during a fatiguing contraction. The time-dependent modifications of the electromyographic (EMG) signal detected from the surface of the skin during sustained muscular contractions provides an attractive means of accomplishing this task. Historically, many investigators have reported both an increase in the EMG signal amplitude and a shift in the frequency spectrum when a contraction is sustained. The most commonly used measure of the amplitude is the mean rectified value; however, with the development of functional modules in solid-state circuit devices, the root-mean-square (rms) processor has gained wider acceptance. Although relatively easy to quantify, the amplitude is subject to large random variations even with all experimental conditions maintained constant (Stulen and De Luca, 1978). Other results have shown the amplitude to increase, decrease, or remain constant, depending on the force, velocity and duration of a contraction, and the type of electrode used to record the signal. These results have detracted from the use of EMG signals as a quantitative clinical measure.

Alternatively, spectral measures of the surface detected EMG signal have gained wider acceptance. The power density spectrum of the EMG signal expresses its energy content as a function of frequency. It is well documented that the power density spectrum of an EMG signal is compressed toward lower frequencies as a sustained isometric contraction progresses (Kadefors et al., 1968; Linström et al., 1970; Mortimer et al., 1970; Stulen and De Luca, 1981; Mills, 1982; among others). This measurable phenomenon has been empirically associated with a slowing of the muscle fiber conduction velocity and the associated biochemical events. Therefore spectral parameters are more suitable than amplitude parameters for monitoring fatigue processes within a muscle because they are more directly correlated to the physiological process. Indeed, the increase in amplitude that can be observed during fatigue has been described as a second order effect due to the low-pass filtering effect of the body tissues on the surface EMG signal. The spectral shift which accompanies fatigue places more of the EMG signal energy in the passband of the tissue filter, thereby causing the signal amplitude to rise.

Early attempts at measuring spectral compression passed the EMG signal through a bank of narrow bandpass filters to observe the change in the spectrum (Kadefors et al., 1968; Broman and Kadefors, 1979). It is simpler if a single frequency parameter can be used which effectively represents the spectral change. With the advent of computer-aided signal processing techniques, a number of useful parameters from surface detected EMG signals can now be measured with relative ease. Several researchers have adopted frequency parameters including the mode, mean and median frequencies of the EMG power spectrum. Schweitzer et al. (1979) have investigated the mode frequency of the EMG signal power spectrum, and found that it produces unreliable estimates of the spectral compression. Lindström et al. (1977) have developed a digital means of computing the mean frequency for use in monitoring fatigue. Stulen and De Luca (1978, 1981) have shown, however, that the median frequency – the frequency at which the power density spectrum is divided into portions of equal energy – provides a more reliable estimate of spectral frequency transitions when noise accompanies the EMG signal. Thus the median frequency has been proposed as the preferred parameter for monitoring the spectral shift associated with localized muscle fatigue. An additional advantage to using the median frequency parameter is that it is relatively easy to compute with analog hardware. A computer-assisted device, called the Muscle Fatigue Monitor© (MFM) was devised for this purpose (Gilmore and De Luca, 1985). The MFM consists mainly of an analog circuit which continually computes the median frequency of the power spectrum of an input signal by utilizing sharp cutoff analog voltage controlled filters as depicted in Fig. 1.

First, the EMG signal is detected, amplified and bandpass filtered via a specialized surface electrode and preamplifier. Then it is passed through a fourth-order, modulated low-pass filter and into a true-rms converter. The rms voltage is then compared differentially to the rms voltage obtained by attenuating the original EMG signal by 3 dB, and integrating the difference. The integrator output, in turn, modulates the cutoff frequency of the low-pass filter so that the difference between the rms voltages is driven towards zero. Since the median frequency is defined as the half-power frequency, the integrator output thus corresponds uniquely to the median frequency of the EMG power density spectrum. A primary advantage of using the MFM to compute median frequency rather than a digital, fast Fourier transform method is that the MFM computes and plots median frequency in real time. This advantage is particularly evident when multiple channels of EMG signals need to be processed simultaneously in real time, as in the present MFM configuration.

An alternative approach has been described in which an analog microprocessor (INTEL 2920 and 2921) performs the on-line tracking of median frequency (Merletti et al., 1985a). The operating principle of this technique uses a fixed high-pass digital filter and a shifting EMG spectrum obtained by amplitude modulation of a sine wave carrier of variable frequency F_c. The difference between the carrier frequency F_c and the ideal filter cutoff frequency F_f matches the median frequency when the output power of the filter is 1/8 of the total EMG power. The advantages of this system are that analog circuitry and adjustments are minimized and the computer algorithm is totally software implemented within the EPROM of the INTEL 2920 microprocessor. One reported drawback to this

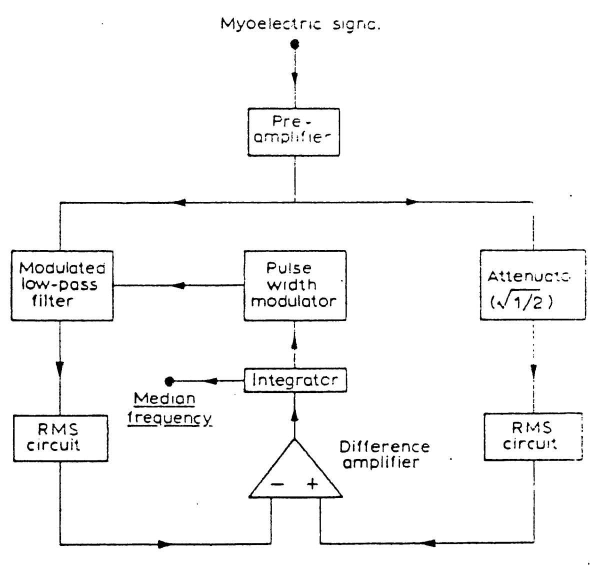


Fig. 1. Block diagram of the median frequency tracking circuitry of the Muscle Fatigue Monitor (MFM) U.S. Patent 4 213 467. (From Stulen and De Luca, 1982)

technique is the increasing quantization error band at low frequencies. Further developments of integrated signal processors in the near future should improve this device.

A different approach is to obtain spectral information from a computer-based fast Fourier transform. Bassano and Ottonello (1986) describe their utilization of a microprocessor to obtain very fast, on-chip evaluation of a 64 point Fourier transform. They also describe the use of commercially available VLSI circuitry that can calculate a 128 point FFT in about 8 ms. For real time displays of the EMG spectrum it is possible to extend the system to a 1024 point FFT with only a slight increase of complexity. The authors report that this system has the advantage of presenting a greater variety of spectral information. The disadvantage, at present, is the comparatively low resolution of the device (± 8 Hz); however, they report that a new prototype based on the same principles can provide improved accuracy (± 2 Hz).

CHARACTERISTICS

Repeatability

The median frequency has been studied with respect to different time durations between tests, different force levels of contraction, different muscle groups and different individuals. Initial median frequency (IMF) measurements were assessed for repeatability within the same experiments and between experiments on the same subject. The IMF is defined as the median frequency estimate that corresponds to the very beginning of a contraction before fatigue processes have modified the EMG signal. In this instance, the IMF was calculated as the y-intercept of an exponential curve fitted to median frequency data plotted as a function of time. Fig. 2 presents the intra- and the inter-experiment variability as a function of contraction level for one subject. The data is from the left (Fig. 2*a*) and right (Fig. 2*b*) tibialis anterior muscle which was tested for brief (10 s) contractions repeated four times each at 20%, 50%, 80% and 100% MVC levels. Each experiment was then repeated on four different days. The figure demonstrates that the variability was greater between experiments than within an experiment and that the intra-experiment variability increased with the force level of the contraction. It is also apparent that the average values of the initial median frequency increases as a function of contractile level. This consistent finding will be discussed later in this chapter.

The repeatability for long duration contractions was also studied. Six subjects (five males, one female) with average age of 24 ± 2.7 performed two sustained isometric contractions of the deltoid

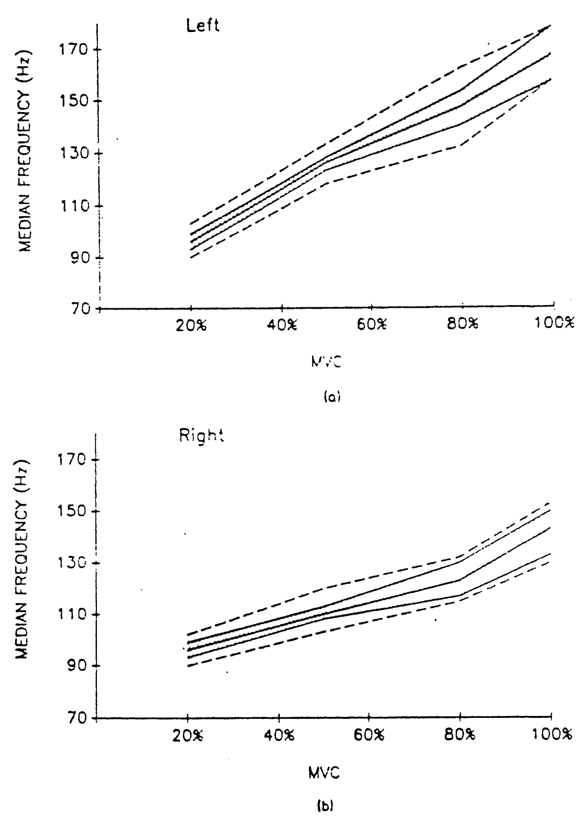


Fig. 2. Initial median frequency from (*a*) the left and (*b*) right tibialis anterior muscle of one subject as a function of the contraction level. Central (solid) line: the means of four experiments; hashed lines: the mean standard deviation within an experiment; broken lines: standard deviation of the four means for the four experiments. (From Merletti et al., 1985b)

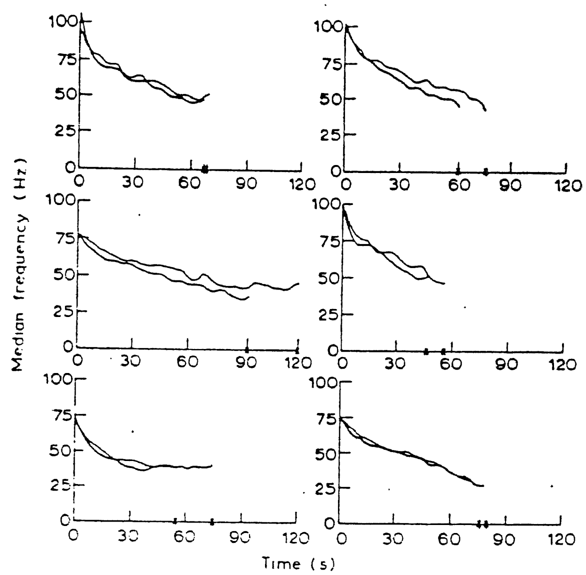


Fig. 3. The variability of the median frequency during a sustained contraction. Each of the six graphs contain two curves obtained from the same subject during a sustained, isometric, constant-force contraction performed at 50% MVC. (From Stulen, 1980)

muscle at a constant force of 50% MVC. These contractions were performed on the same day with prolonged rest periods between contractions.

The results, in Fig. 3 are displayed as six graphs containing two curves each. The curves are the median frequencies obtained from the two contractions at different times in the same day. The curves are consistently similar throughout the duration of the contraction.

Repeatability on the same and different days have also been compared for initial and slope parameters of the median frequency for the erector spinae muscles of the lower back. Eight male subjects with ages ranging from 21 to 40 years performed constant-force, isometric back extension at 80% MVC. These contractions were sustained for 30 s and were performed three times with a 1·5 min rest period between contractions. All eight subjects repeated the same protocol on the next day. The electrodes were removed from day-to-day but their locations were marked with indelible ink. Figs. 4 and 5 demonstrate the repeatability of the IMF and

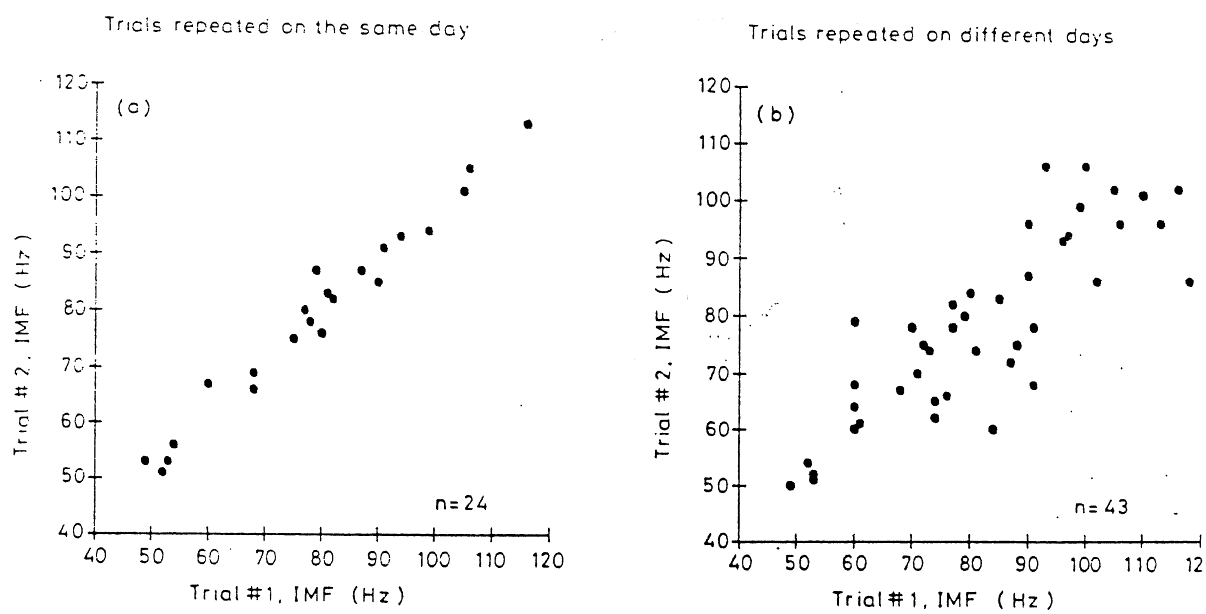


Fig. 4. The results of two identical trials repeated on (*a*) the same day and on (*b*) different days for EMG data from six detection sites on the lower back in four different subjects (n = 24). Data was processed for initial median frequency (IMF) values.

slope parameters for recordings on the same day and different days, respectively. The reliability estimates are summarized in Table 1. Although these results favor the clinical variability of the technique, a slightly higher variability in experimental error was observed for the median frequency slope value and for measurements on different days. These results are most likely attributable to the high sensitivity of the technique to electrode location and to transients in the EMG signal associated with motion artefacts.

TABLE 1

Reliability estimates obtained from same day and different day recordings

Parameter	Reliability estimates	
	Same day recordings	Adjacent day recordings
Slope of median frequency	$\varrho_1 = 0.94$	$\varrho_1 = 0.73$
Initial median frequency (IMF)	$\varrho_1 = 0.98$	$\varrho_1 = 0.83$

Reliability

Reliability describes how well an instrument or procedure is able to measure the process it intends to quantify. Median frequency measurements from the MFM have been tested for reliability by a number of different methods that utilize either the EMG signal, sine waves, or band-limited white noise sources having known power density spectral shifts. Fig. 6 shows the performance of the MFM for an EMG signal obtained during a sustained, isometric constant force contraction. The power density spectrum of the sampled EMG signal was calculated digitally using a fast Fourier transform algorithm. The median frequency was then calculated directly from the spectrum. As seen in this figure, the estimates of the median frequency determined by the MFM and digital computation were essentially equal. Similarly, the system's estimated median frequency response to a specified input using sine wave and band-limited white noise

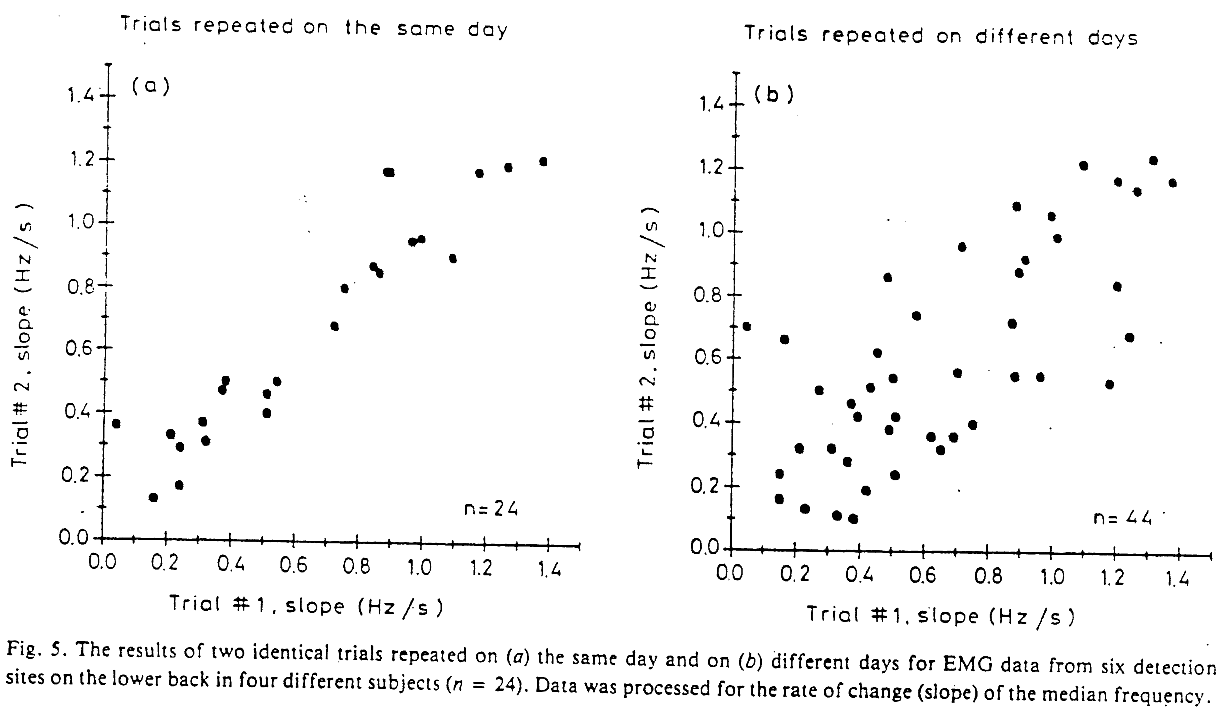


Fig. 5. The results of two identical trials repeated on (*a*) the same day and on (*b*) different days for EMG data from six detection sites on the lower back in four different subjects (n = 24). Data was processed for the rate of change (slope) of the median frequency.

sources compared favorably to those obtained using FFT spectral analysis (Fig. 7). This plot shows a deviation of less than 5% within a range of median frequencies from 20 to 255 Hz.

The issue of reliability is also of concern when measuring spectral parameters from non-static contractions. Since most movements of the human body involve muscle contractions which are dynamic (varying force production and/or changing muscle length), there is considerable need for objective fatigue measurements during dynamic contractions. One sub-category of dynamic contractions is force-varying, isometric contractions in which muscle length is held fixed but the force production of the muscle is variable. One major difference between force varying, isometric contractions and constant force, isometric contractions is that the EMG signal amplitude is variable in the former and essentially constant in the latter. This results from the direct relationship between signal amplitude and muscle force production. To help distinguish between physiological phenomena and circuit response in these instances, it was necessary to further evaluate MFM reliability using control signals with known median frequency and amplitude variations. A signal generation algorithm which produces band-limited noise with specialized frequency and amplitude characteristics was developed for this purpose. Several classes of signals were generated for testing which included constant, step, and ramp median frequency combined with either constant or sinusoidal rms. The range of MF and rms selected was carefully chosen to be in accordance with the physiological range of median frequency values associated with EMG signals. Sinusoidal amplitude variations were used because they are typical of rms curves obtained from EMG signals during repetitive force varying contractions. One particular class of signals tested in which the noise amplitude varied sinusoidally and the median frequency changed linearly, is most representative of EMG signals associated with force varying contractions. Results from these tests led to circuit design improvements in the MFM which virtually eliminated the effects of amplitude changes in the input signal on the median frequency measurements.

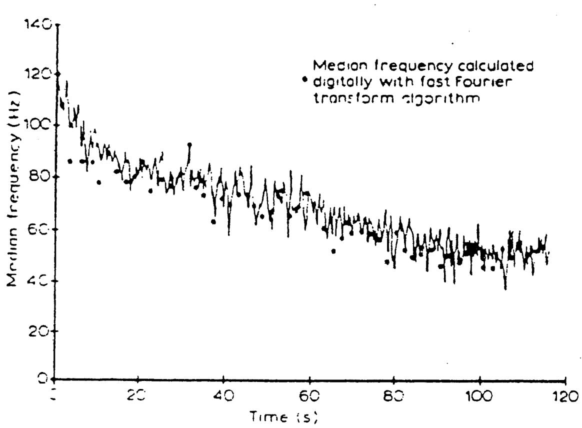


Fig. 6. Comparison of the median frequency determined by digital computations and the muscle fatigue monitor (MFM). (From Stulen and De Luca, 1982)

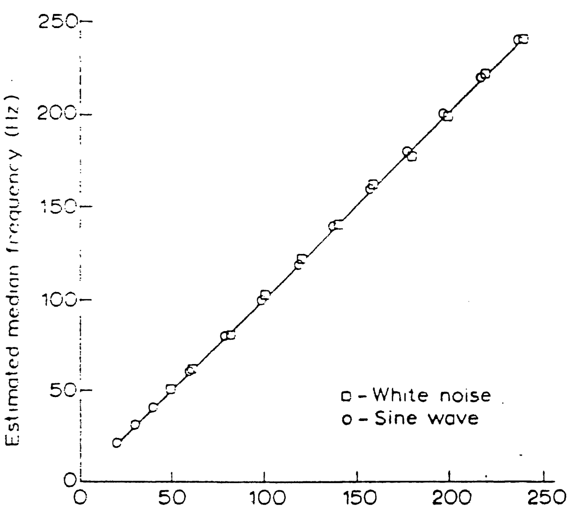


Fig. 7. Graph of the MFM system's estimated median frequency response to a specified median frequency input using sine wave and band-limited white noise sources. (From Gilmore and De Luca, 1980)

Validity studies of the kind described above are useful (and necessary) to clarify the capabilities and limitations which a particular device possesses in processing the types of signals associated with dynamic contractions. However, this does not necessarily mean that the parameter being monitored

– median frequency – yields meaningful information regarding fatigue for these contractions. The rate of force variation, changes in muscle length, movement of the skin over muscle and the problems of signal stationarity all affect the median frequencies recorded during dynamic contractions. Further studies are needed to address these issues.

Electrode location

Recent studies have demonstrated that changes may occur in the median frequency of the EMG signal as a result of either signal detection procedures or the physiological state of the muscle prior to fatigue. Because this technique relies upon surface electrodes to detect the signal, alterations in the shape of the signal can occur as a result of differences in the conductivity of the tissue beneath the electrode and differences in the location of the electrode with respect to the innervation zone and tendon. The spectral content of the signal and its conduction velocity are altered by changes in the observation distance from the signal source, the direction of the signal propagation and the electrical conductivity of the muscle fiber. These effects are well described mathematically for idealized muscle models (Lindström, 1970; De Luca, 1985). Roy et al. (1986b) studied some of these effects empirically, since the methodological factors in vivo may not be the same, or of the same magnitude, as in the theoretical model. Numerous factors related to the nonhomogeneous and anisotropic properties of muscle may modify these results (Cunningham and Hogan, 1981; Gielen and Boon, 1981). Roy et al. described the sensitivity of the median frequency parameter to changes in the location of the detecting electrode with respect to the innervation zone(s) and tendinous portion of the tibialis anterior muscle. Fig. 8 demonstrates the considerable variation in the median frequency as a result of the electrode location reported in this study.

The highest values of the median frequency always occurred at the region of the innervation zone and tendinous insertion of the muscle, and decreased proportionally with distance from these areas. This finding is consistent with the predictions of theoretical models which describe action potentials of different wavelengths propagating in opposite directions from the innervation zone, forming a complex interference pattern (Desmedt, 1958; Basmajian and De Luca, 1985). The superposition of minutely phase shifted action potentials in the vicinity of the innervation zone provides a relative increase of the high frequency components of the EMG signal (Lindström and Petersen, 1977). When the electrode is placed at either

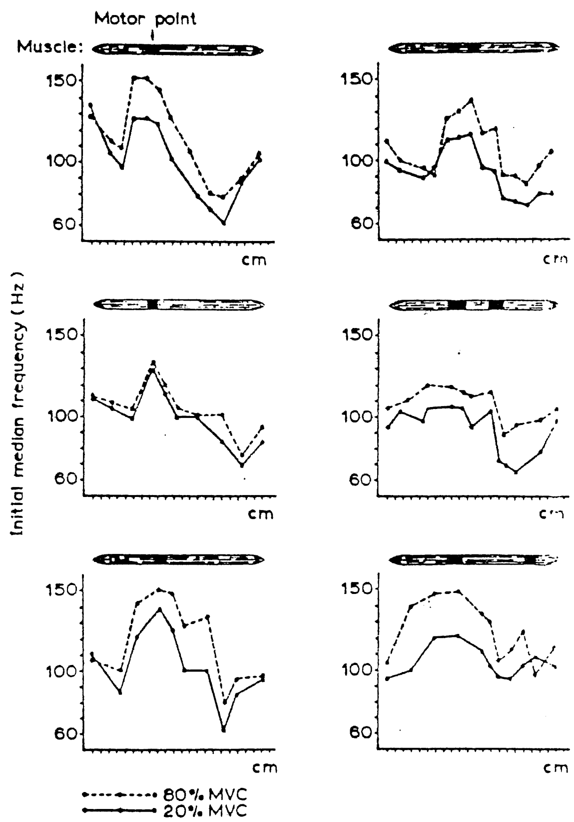


Fig. 8. Median frequency measures from surface electrodes placed along the length of tibialis anterior muscle. Each graph indicates results from a different subject. Data points are average initial median frequency values for three sustained constant-force contractions. Dotted lines connect data points from tests conducted at 80% MVC; solid lines connect data points from tests conducted at 20% MVC. Location of tibialis anterior motor point(s) for each subject is indicated above each graph. (From Roy et al., 1986)

end of the muscle fiber, the relative high impedance of the tendon tissue truncates the action potentials and increases the high frequency components of the signal which results in a relative increase in median frequency. Interestingly, it was found that the rate of change of median frequency was not effected by electrode location (Roy et al., 1986b). This observation was not surprising considering that muscle fiber types are spread homogeneously throughout a muscle and that the surface electrode tends to detect a relatively large domain that normally includes many motor unit populations. Furthermore, the lactic acid produced in any segment of the muscle will diffuse

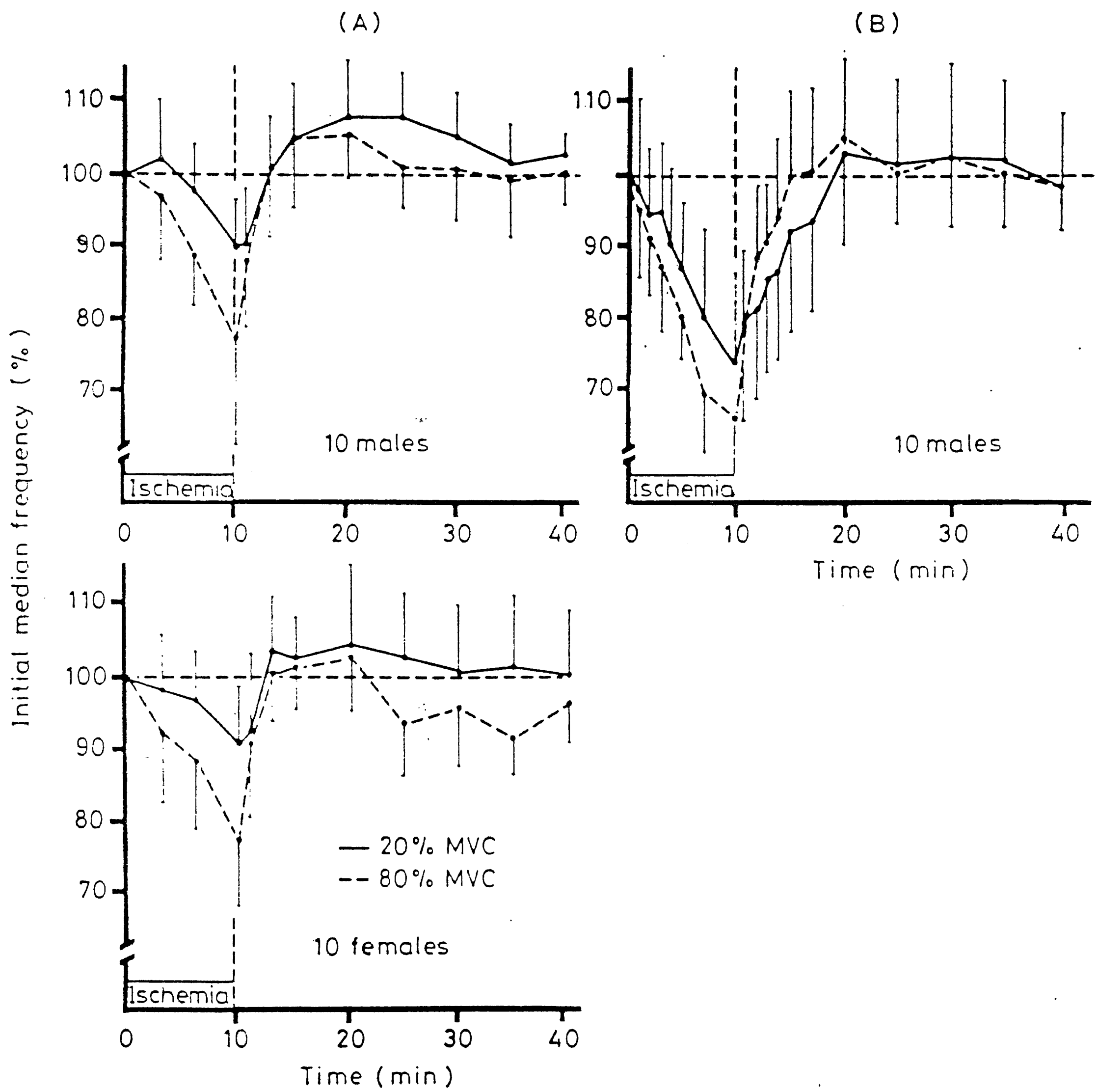


Fig. 9. Behavior of the initial median frequency (IMF) during and subsequent to ischemia involving three contractions (*A*) and six contractions (*B*) during 10 min of ischemia. The contractions were performed separately at 20% MVC and 80% MVC. The diagrams show the average (± SD) from ten females and ten males. IMF values are normalized with respect to control values taken prior to ischemia for each contraction level. (From Merletti et al., 1984)

throughout the interstitial fluid of the whole muscle. This will influence the membrane properties of the muscle fibers throughout the muscle.

Physiological inferences

Further investigations have been conducted which are relevant to understanding the sensitivity of the median frequency to other physical conditions of the muscle. Blood flow and temperature are two measurable influences that are known to significantly affect the measurement of myoelectric spectral parameters during fatigue (Hara, 1980; Mills, 1982; Merletti et al., 1984). In fact, from a theoretical basis, these factors are crucial for determining the resultant shape of the EMG signal since they directly influence the conduction velocity. Experiments involving the first dorsal interosseous (FDI) were conducted by Merletti et al. (1984) that demonstrated a significant reduction of the IMF in contractions performed under ischemic conditions; upon release the IMF recovered quickly (Fig. 9). It can be seen in the figure that at 80% MVC, a greater decrease of the IMF was recorded when compared to the 20% MVC. These results are consistent with the fact that when the blood is occluded, acidic by-products accumulate in the environment of the muscle fiber membrane and decrease the conduction velocity of the muscle fibers. In this same study, the median frequency was also shown to be affected by the muscle temperature. Their results demonstrated a linear decrease of median frequency with the reduction of muscle temperature by external cooling (Fig. 10). The gradual recovery of the IMF is demonstrated in the figure. These observations are consistent with the fact that conduction velocity of muscle fibers is proportional to its temperature.

There is compelling evidence in the literature favoring the role of the myoelectric conduction velocity in determining the shape of the EMG

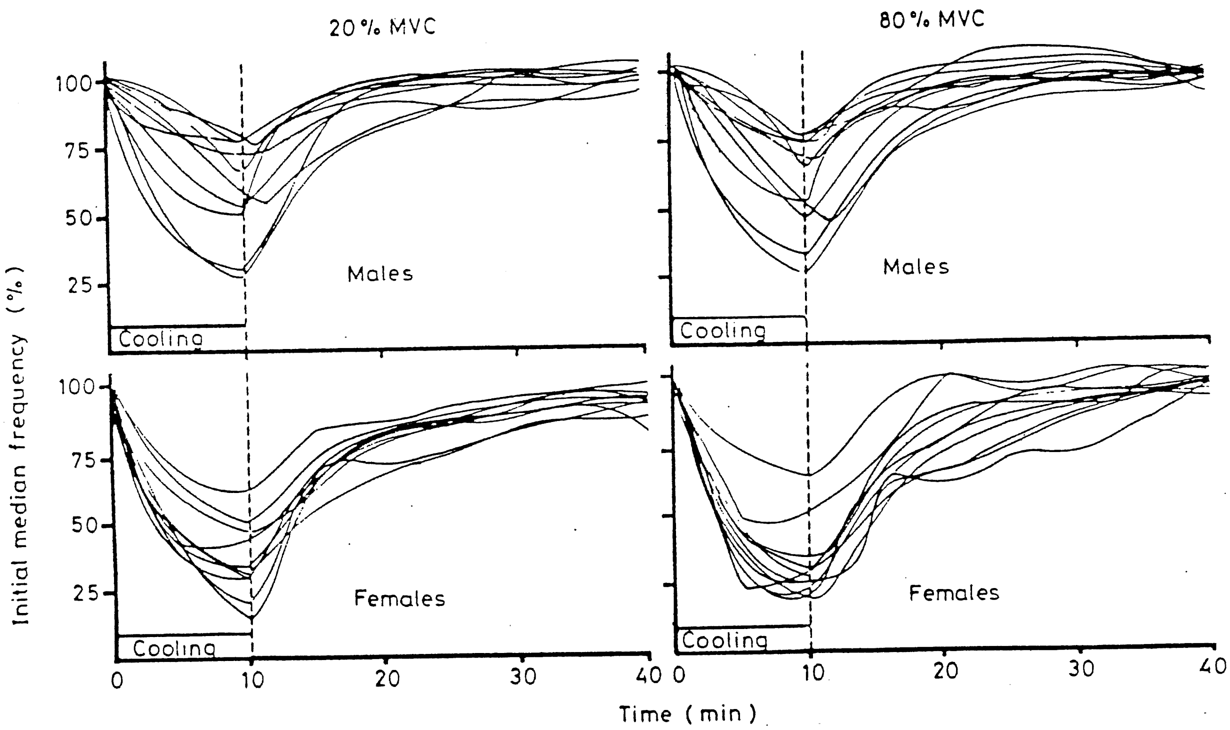


Fig. 10. The initial median frequency (IMF) as a function of time during and subsequent to cooling of the first dorsal interosseous muscle. Muscle contractions were elicited at 20% MVC and 80% MVC. IMF values are normalized with respect to control values taken prior to cooling. Each trace represents one subject. Results are from ten males and ten females. (From Merletti et al., 1984)

signal (for a review see De Luca, 1984). There is, however, no conclusive interpretation of the power spectral changes observed during prolonged, forceful contractions (see chapter 9 of this volume). There are, for instance, persuasive theoretical arguments based on mathematical models of the EMG that describe a linear relationship between the conduction velocity and the median frequency (Lindström, 1970; Stulen, 1980; Stulen and De Luca, 1981). The data that has provided empirical support for this relationship has typically described a greater decrease in median frequency than conduction velocity when measured concurrently during sustained, isometric contractions. Fig. 11 demonstrates this behavior for data from the tibialis anterior muscle in which conduction velocity and median frequency estimates were measured. The regression line was calculated as:

$$MF = (23.1)(CV)$$

where CV is in m/s and MF is in Hz; the regression had a value of $R = 0.84$. Similar linear relation-

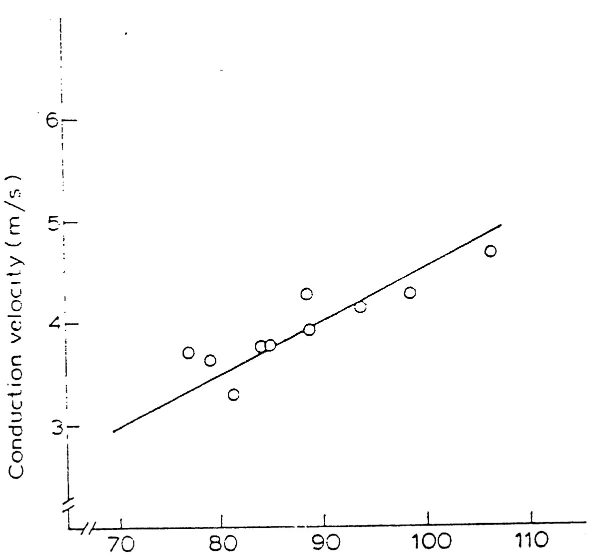


Fig. 11. Relationship between conduction velocity and median frequency for data obtained from electrode site associated with highest cross-correlation values for each subject. (From Roy et al., 1986)

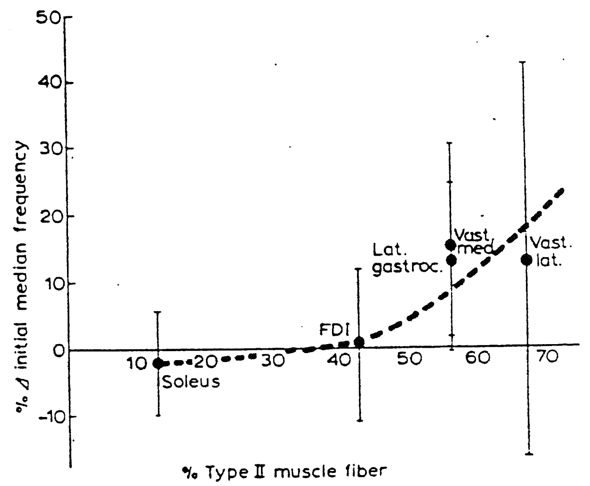


Fig. 12. The percentage change of the initial median frequency (from 20% MVC to 100% MVC) is shown to increase exponentially with the percentage composition of type II muscle fibers in a particular muscle. The intersubject variations are probably due to the difference in the percentage of type II fiber composition. (From De Luca et al., 1983)

ships between CV and MF have been reported elsewhere (Stulen, 1980; Broman et al., 1985). The differences in the behavior of these parameters suggests that other mechanisms, such as motor unit firing characteristics, may contribute to the modification of the EMG signal observed during fatigue inducing contractions (Bigland-Ritchie et al., 1981; Jones and Lago, 1982).

The direct and indirect relationship(s) of the median frequency shift to physiological, anatomical and biochemical events within the muscle make it a prime candidate for a noninvasive technique of muscle fiber typing. However, to date, very little work of this nature has been reported in the literature. In one preliminary study (Rosenthal et al., 1981), the surface EMG signal was detected from the soleus, gastrocnemius, vastus medialis, vastus lateralis, as well as the FDI muscles of six males. These muscles were specifically selected on the basis of their different fiber type composition (assessed by reports in the literature). Fig. 12 indicates that the percentage difference of the median frequency for brief contractions at 20% MVC and 80% MVC are exponentially related to the percentage of Type II muscle fibers within these muscles. Much further work is needed before this technique can be considered as a noninvasive alternative to muscle biopsy.

TIME AND FORCE BEHAVIOR OF THE MEDIAN FREQUENCY

The time and force behavior of the median frequency has been studied extensively for sustained, constant force isometric contractions (for a review, De Luca, 1985). However, little is known about the behavior of the median frequency for sustained, periodic isometric force contractions. These cyclic contractions, at a particular duty cycle, may more closely approximate the development of localized muscle fatigue for repetitive tasks. As such, their study may have direct applicability to the work environment or to sports medicine. Preliminary work was conducted in our laboratory to measure and compare the behavior of the median frequency for both periodic and sustained contractions that differed according to the desired force level (Stulen, 1980). Six male subjects with average age 26 ± 3.2 years performed a series of isometric contractions of the FDI and deltoid muscles. Three different periodic force contractions were performed between force levels of 80% and 20% MVC. These contractions were performed by both the FDI and deltoid muscles. The contractions were denoted by 2H-2L, 4H-4L and 2H-6L where the number indicates the time in seconds contracting at 80% MVC (denoted by H), and at 20% MVC (denoted by L). In addition to these contractions, a sustained, constant force contraction at 80% MVC was tested for comparison. The median frequency was computed by the MFM only for sections of the data corresponding to the 80% MVC contractions. A simple exponential decay was used to fit the sections of the median frequency obtained at 80% MVC (Fig. 13).

The averages and standard deviations for the time constants of the contractions are presented in Table 2. A linear regression was performed between the time constants obtained from the 2H-6L and the 4H-4L contractions and the percentage of

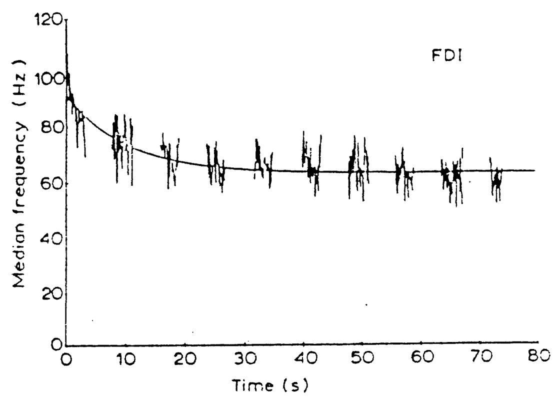


Fig. 13. Typical curve-fit to the median frequency obtained after a sustained, isometric, periodic-force contraction. (From Stulen, 1980)

TABLE 2

Average values and standard deviations of the time constants of the decay in the median frequency during sustained, isometric, periodic force contractions (From Stulen, 1980)

	Contractions			
	4H-4L	2H-2L	2H-6L	80% MVC
FDI	15.2 ± 12.2	6.19 ± 3.98	37.8 ± 27.9	2.79 ± 1.80
Deltoid	22.3 ± 22.5	20.5 ± 15.4	27.6 ± 25.8	13.5 ± 25.0

Notes: 80% MVC, sustained constant-force isometric contraction. #H − #L, # is time in seconds contracting at 80% MVC (H) or 20% MVC (L).

time spent at 80% MVC. The time constants from the sustained, constant force contractions performed at 80% MVC were also used in this analysis. Fig. 14 is a plot of the average and individual time constants of the FDI and deltoid muscles corresponding to the time spent at 80% MVC. In this figure it can be seen that as the percentage of time spent at 80% MVC decreased, the time constant increased. The time constants for the 2H-2L contractions were also compared to the values obtained from the 4H-4L contraction to determine the effect of the period of repetition (4 and 8 s). The average time constant was longer for the 4H-4L contraction than for the 2H-2L contraction for both the results obtained from the FDI and deltoid. However, this difference was not significant at $P = 0.01$ in either case.

The percentage decrease in the median frequency was also considered. The average decreases for the three sustained, periodic force contractions were all significantly less than the average decrease in the median frequency obtained from the constant force contraction performed at 80% MVC. Yet, no difference was observed between the decreases of just the periodic force contractions.

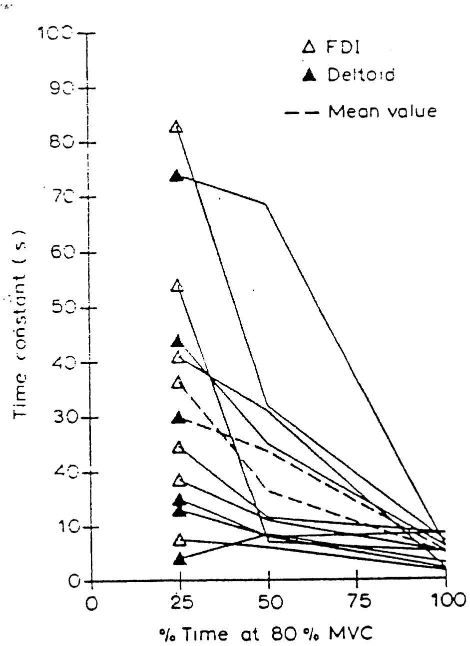


Fig. 14. The average (broken line) and individual (solid line) curves of the time constants of the single exponential curve fit of the sustained, isometric, periodic force contractions versus percentage of time at 80% MVC (duty cycle). (From Stulen, 1980)

The return of the median frequency to its initial value after a sustained contraction provides additional information about the fatigue-related, physiological processes occurring within a muscle. The recovery process is most closely related to the ability of the arterial blood flow to purge the extra- and intra-cellular muscle environment of the metabolic by-products that accumulate during a contraction. We tested a series of short contractions during recovery from a sustained constant-force contraction.

After each of the sustained contractions at 80%, 50%, and 20% MVC were performed, subjects periodically contracted briefly (5 – 10 s) at 20% MVC during a 10 min recovery period. In addition to these contractions, a brief 20% MVC contraction was performed 1 min prior to the sustained contraction to establish a baseline for the recovery phase. Six male subjects with average age 26 ± 3.2 years were tested for both the FDI and deltoid muscles.

A simple rising exponential curve was used to fit the median frequency obtained during the recovery. The initial value of the curve was forced to correspond to the average ending value of the sustained contraction. The results were plotted for each subject as a function of recovery time in Figs. 15 and 16 for the data obtained from the FDI and deltoid respectively. A linear regression was performed for the time constant of recovery versus the relative force of the sustained contraction. When the data for both muscles are considered together, the slope of the regression is significantly different than a slope of zero ($P < 0.002$). The recovery time constant is therefore dependent on the relative force of the sustained contraction. The time constant measured in seconds may be estimated from the relative force by the following relation:

$$\text{time constant} = 93.3 - 0.577\ (\%\text{MVC})$$

The average values and standard deviations of the time constants for the three relative force levels are given below in Table 3. The grand average of all the time constants is 66 s. Therefore, the average time for the median frequency to fully recover is

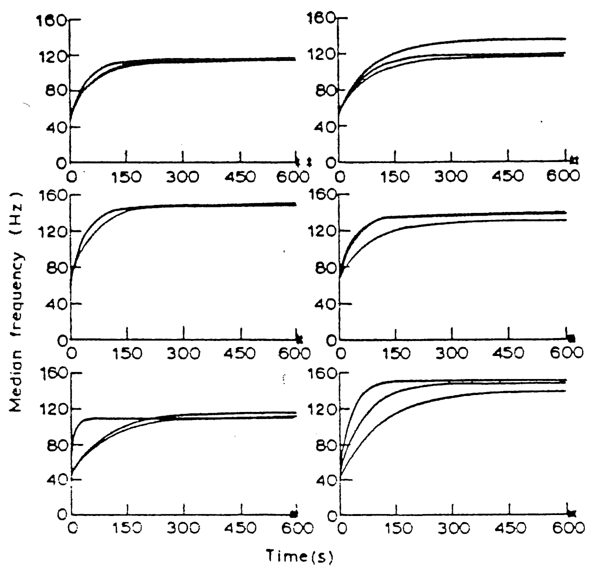


Fig. 15. The recoveries of the median frequencies obtained from six subjects following the sustained, isometric, constant-force contractions of the FDI muscle. (From Stulen, 1980)

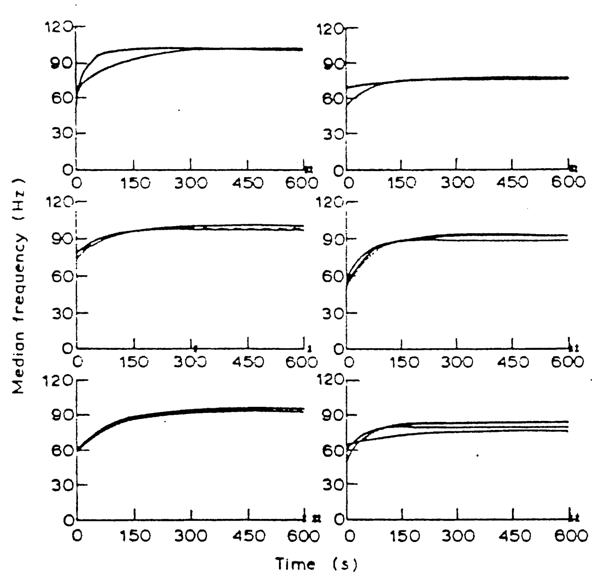


Fig. 16. The recoveries of the median frequencies obtained from six subjects following the sustained, isometric, constant-force contractions of the deltoid muscle. (From Stulen, 1980)

266 s. The time constant was also tested as a function of the percentage decrease in the median frequency of the previous sustained contraction. There was no significant relation at $P < 0.01$ between the time constant and decrease in the median frequency for the results obtained from either the FDI or deltoid muscles.

The relationship of the recovery time of the median frequency to physiological correlates of fatigue needs further investigation. Stulen (1980) attempted to measure intramuscular pH with a probe in the FDI muscle of the hand while detecting the surface EMG during a prolonged 50% MVC contraction and during recovery. Although the median frequency and pH measures appeared closely correlated, the pH measurements using the probe had unacceptable measurement error. Recent advances in ^{31}P MR (magnetic resonance) spectroscopy may provide the noninvasive, in vivo measures of pH and other fatigue correlates needed to continue these important investigations. Miller et al. (1985), for instance, have successfully combined surface electromyography with ^{31}P MR spectroscopy. They reported time histories of pH decrease and recovery during sustained 100% MVC contractions of the adductor pollicis muscle that closely approximates median frequency findings by other investigators for similar contractions. The present rapid growth of physiological imaging technologies and spectroscopy techniques will undoubtedly assist basic research efforts by electromyographers studying muscle fatigue (Kushmerick, 1986).

TABLE 3

Average values and standard deviations of the time constants of the rise in the median frequency during the recovery phase as a function of the force of the previously sustained contractions (From Stulen, 1980)

	Contraction level (% MVC)		
	20%	50%	80%
FDI	80.7 ± 17.6	57.9 ± 18.3	41.2 ± 25.2
Deltoid	86.9 ± 28.9	66.8 ± 32.5	58.2 ± 20.0
Both	83.8 ± 23.0	62.3 ± 25.5	50.5 ± 23.1

APPLICATIONS

Effect of physical exercise

Continued preferential use of selected muscle groups may be expected to induce biochemical and structural modifications to the muscle fibers (Fugl-Meyer et al., 1982; Schantz et al., 1983; Tanaka et al., 1984). The frequent preferential use of the dominant hand in daily activities may over the years act as a form of endurance training for the muscles of the hand. It, therefore, follows that fatigability of the muscles in the dominantly used side may differ from that of corresponding contralateral muscles. The median frequency of the EMG signal should provide an appropriate measure of these modifications since it is known to be influenced by the muscle fiber size and type (De Luca, 1985). However, the use of the EMG signal to compare the performance of corresponding contralateral muscles has been meagre (Coogler, 1983). An investigation was therefore conducted to measure the sensitivity of the MF parameter to the effects of hand dominance (De Luca et al., 1986).

A total of 35 normal adult subjects were studied; 16 were females with an average age of 25.6 ± 3.0 years, 19 were males with an average age of 27.3 ± 3.7 years. 18 subjects were right handed and 17 were left handed. The FDI muscle for each hand was tested during isometric, constant force abduction of the index finger. Each subject performed three contractions each at 20%, 40%, and 80% MVC held for a maximum of 60 s. Two parameters were measured: the initial median frequency and the rate of decrease (slope) of the median frequency of the EMG signal. When the data are plotted as a function of the force level at which the contractions were performed, no distinction in the median frequency is seen as a function of force (Fig. 17). In contrast, the slopes of the median frequency do show a distinction (Fig. 18).

The linear regression line for the data from the right hand of right handed subjects has a slope value which is lower than those of the other three sets of data in Fig. 18. More importantly, in the right handed subjects, the values of the slope of the median frequency from the right FDI are lower in a statistical sense than the corresponding values from the left FDI. This indicates that the median frequency of the dominant FDI muscle in right handed subjects decreases more slowly than that of the non-dominant muscle. This clearly noticeable and significant distinction suggests that the accumulated preferential usage of the dominant hand during the course of a lifetime might have

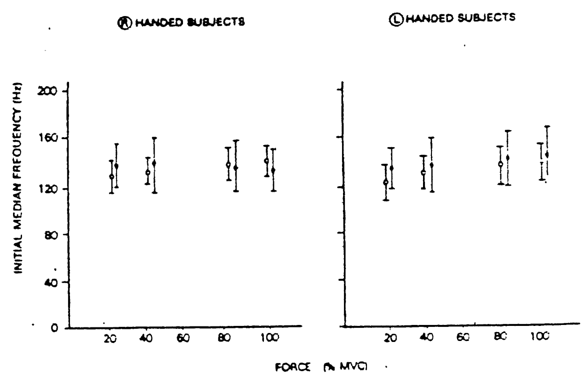


Fig. 17. Mean value and standard deviation of the initial median frequency (IMF) of all the trials as a function of contraction force level. Data from the right (●) and left (○) first dorsal interosseous muscles are compared for right and left handed subjects. (From De Luca et al., 1986)

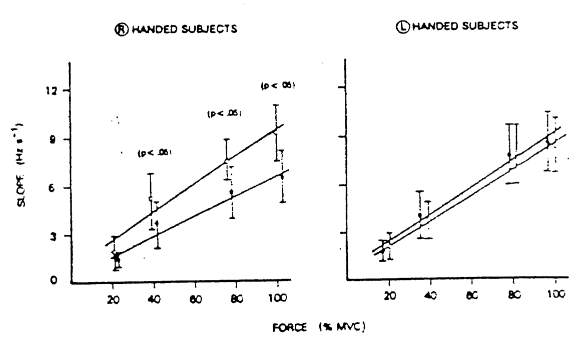


Fig. 18. Mean value and standard deviation of the slope of the median frequency of all the trials as a function of contraction force level. Data from the left and right first dorsal interosseous muscles are compared for right and left handed subjects. Separate regression lines for data from the left [○] and right [●] hand are represented for right handed subjects and left handed subjects. A t test was used to calculate the statistical significance of the difference between mean values at each of the four force levels. (From De Luca et al., 1986)

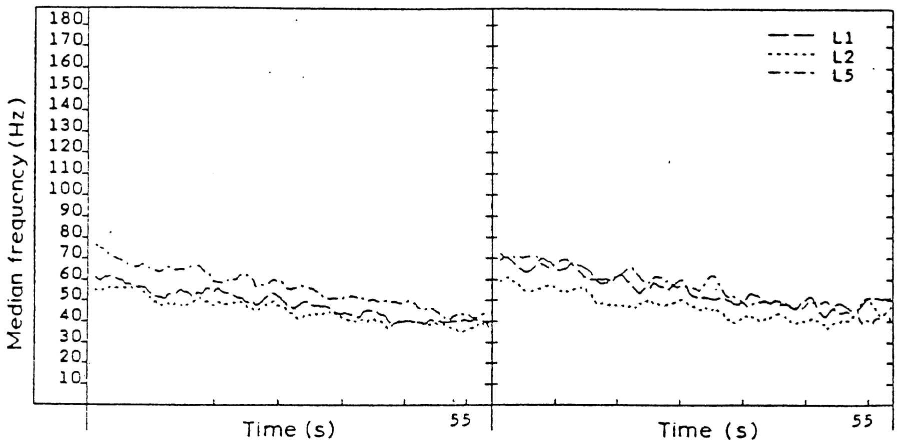
altered the fiber type composition. In these subjects, the dominant FDI displayed a lower rate of decrease of the median frequency than the corresponding contralateral FDI during a sustained contraction. This may indicate a decreased rate of lactate production which, according to the results of Tesch and Karlsson (1977), suggests that there are relatively more type I, slow-twitch, aerobic fibers in the FDI of the dominant hand than the non-dominant hand. The fact that Fig. 18 demonstrates a pronounced distinction in the behavior of the median frequency slope at higher force levels is consistent with the fact that as the force increases, the internal pressure in the muscle occludes the arterioles, shutting off the blood return. Thus, the greater amount of lactic acid generated in the non-dominant FDI would accumulate and have a greater effect on the median frequency of that muscle. An explanation is necessary as to the observation in Fig. 17 that no consistent statistically significant distinction was noted in the values of the initial median frequency as a function of hand dominance. This is contrary to the notion that the initial median frequency value reflects the size of the muscle fibers. However, it should be noted that the validity of this statement must be qualified by the proviso that the EMG signal should not be affected by disturbing influences from the innervation zone and tendons. The FDI muscle used in this study has dimensions that are too small for the 1 cm electrode; hence, these conditions result in ambiguous measurements of the initial median frequency.

Back pain evaluation

The technique of monitoring the frequency shift of the EMG signal for the purpose of measuring localized muscular fatigue has several advantages that make it well suited for clinical use: it is non-invasive; it may be performed on muscles in situ; it may be performed in real time; and it provides information relating to events which occur inside the muscle. These features favor the use of the technique for a variety of applications such as athletic training, physical therapy, respiratory therapy, ergonomics, and the diagnosis and prognosis of neuromuscular disorders. Recent studies in these areas are summarized by De Luca (1985). Many of these applications are still conceptual, and although others have been put into practice, further experimental verification is needed.

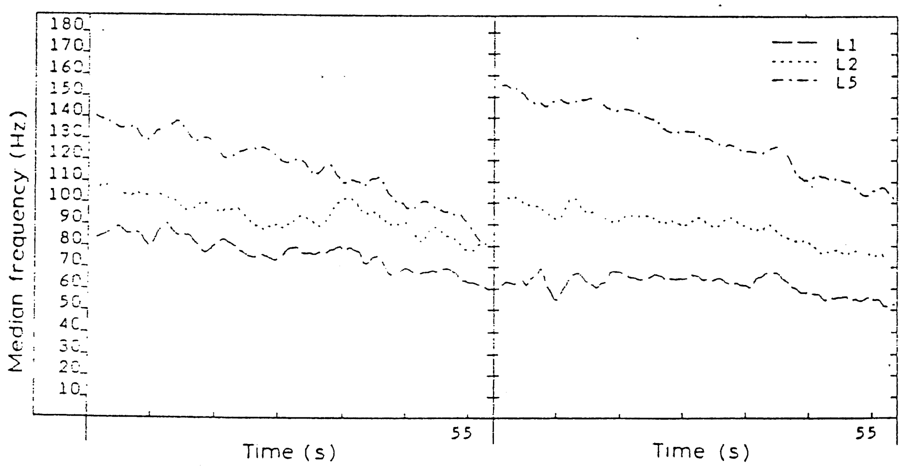
Carefully controlled studies are being initiated in which the clinical usefulness of this technique is being investigated. For instance, lower back pain, one of the most common, and least understood, musculoskeletal disorders is being studied on the basis of EMG spectral information from back muscles (Roy et al., 1986a, 1987). Six electrode sites from bilateral erector spinae muscles at L1, L2 and L5 spinal levels were selected to monitor EMG signal activity during sustained isometric extension of the trunk. Their protocol tests different force levels at 40%, 60% and 80% MVC each held for a maximum of 60 s. Examples of the results for a normal control subject and a chronic back pain patient are presented in Figs. 19*a* and 19*b*. The back pain patient had greater than 2 year history of back pain, had no X ray evidence of structural abnormality to the vertebral complex, and was in a period of remission at the time of testing.

Each figure displays the simultaneous time history of the median frequency for the six electrode sites, with the left and right side of the back plotted separately. These *muscle fatigue patterns* may prove to be a unique and effective way of describing the coordination and efficiency of synergistic muscle systems. In the data presented in Figs. 19*a* and 19*b*, the most obvious differences between these plots are the much higher initial values and the steeper decay of median frequency for the back pain subject compared to the control subject. These preliminary data suggest that back pain patients may have muscles with altered physiological characteristics. Further data is being collected to conclusively state whether this is a consistent finding related to the presence or absence of specific lower back disorders. Muscle fatigue patterns were also analyzed for evidence of asymmetries contralaterally and between muscles at a different spinal level. It was determined, for instance, that



L1:	SLOPE=−0.43 R=0.86 IMF= 60 FMF= 36 NORM SLOPE= −0.71	SLOPE=−0.43 R=0.83 IMF= 67 FMF= 44 NORM SLOPE= −0.64 L/R= 0.99
L5:	SLOPE=−0.57 R=0.93 IMF= 72 FMF= 40 NORM SLOPE= −0.79	SLOPE=−0.59 R=0.91 IMF= 72 FMF= 39 NORM SLOPE= −0.82 L/R= 0.96
L2:	SLOPE=−0.36 R=0.82 IMF= 55 FMF= 35 NORM SLOPE= −0.66	SLOPE=−0.30 R=0.68 IMF= 56 FMF= 39 NORM SLOPE= −0.54 L/R= 1.19

(a)



L1:	SLOPE=−0.44 R=0.85 IMF= 88 FMF= 64 NORM SLOPE= −0.50	SLOPE=−0.17 R=0.46 IMF= 66 FMF= 57 NORM SLOPE= −0.25 L/R= 2.62
L5:	SLOPE=−0.99 R=0.96 IMF=143 FMF= 89 NORM SLOPE= −0.69	SLOPE=−0.99 R=0.97 IMF=158 FMF=103 NORM SLOPE= −0.63 L/R= 1.00
L2:	SLOPE=−0.42 R=0.80 IMF=105 FMF= 82 NORM SLOPE= −0.40	SLOPE=−0.50 R=0.89 IMF=103 FMF= 75 NORM SLOPE= −0.49 L/R= 0.84

(b)

Fig. 19. Median frequency *fatigue patterns* for (*a*) control subject and (*b*) back pain subject. The six curves in each plot represent data from six electrode sites at L1, L2 and L5 spinal levels bilaterally.

amongst the normal subjects tested, the steepest fatigue curves corresponded to the lower lumbar sites (L5) rather than at the upper lumbar areas. This consistent finding was not the case for the back pain subjects. Many of the back pain subjects displayed relatively steep median frequency curves in the upper lumbar areas as well as the lower lumbar area. These differences existed even when back extensor strength, as measured by the 100% MVC, were functionally the same.

EMG spectral measures of fatigue for the back extensor muscles are also being combined with a mathematical model describing the individual force contributions of each muscle (Murthy et al., 1987). The spectral measures from these multiple EMG detection sites are combined with an anatomical database for optimizing the redundancy inherent in the model. The long term goal for this unique method of measuring the muscular defects associated with low back pain is to formulate a computer-aided design approach for prescribing exercises to correct these muscular defects (Roy et al., 1987).

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