Lumbar Muscle Fatigue and Chronic Lower Back Pain

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There currently is a clinical need for an objective technique to assess muscle dysfunction associated with chronic lower back pain. A Back Analysis System for objectively measuring local fatigue in the back extensor muscles is presented. The reliability and validity of this technique was evaluated by testing chronic low-back pain patients and control subjects without back pain. Concurrent surface electromyograms (EMG) were detected from multiple back muscles during sustained isometric contractions at different force levels of trunk extension. Median frequency parameters of the EMG power density spectrum were monitored to quantify localized muscle fatigue. Results indicated: 1) high reliability estimates for repeated trials; 2) significant differences ($P < 0.05$) in median frequency parameters between lower back pain patients and control subjects for specific combinations of contractile force level and muscle site tested; 3) Median Frequency parameters correctly classified lower back pain and control subjects using a two-group discriminant analysis procedure. The applicability of this technique as a treatment outcome measure and diagnostic screening method for lower back pain patients is discussed. [Key words: EMG, median frequency, lumbar musculature, muscle fatigue, low-back pain]

LUMBAR MUSCLE FUNCTION is considered to be an important component of chronic lower back pain (LBP).[1,33,36,44] It has been found that individuals with endurable back muscles and general physical fitness have fewer incidences of back problems than deconditioned cohorts.[8,28,44] Complementary studies have documented compromised muscle function in patients with LBP.[2,8,29,37,45,48,56,62] Although the mechanism associating muscle insufficiency to LBP is not clearly understood, it is commonly held that the passive tissues of the spine are increasingly stressed with increasing functional muscle insufficiency.[9,20,23,54] The high incidence of back injury among workers exposed to fatiguing manual tasks and whole body vibration lend support to this concept.[18] To further understand the relationship of muscle function to LBP, more effective assessment procedures need to be developed and tested for clinical use. Techniques presently available to assess muscle deficiencies are either nonobjective or they lack rigorous clinical validation and reliability.[30,54]

Recent findings suggest that objective evaluation of back muscle fatigue by surface electromyography (EMG) may be of practical importance for the assessment and treatment of muscle deficits associated with LBP.[50,51] These preliminary investigations documented the feasibility of applying quantitative spectral techniques to EMG data collected from multiple surface electrodes on the lower back. It was demonstrated that the dynamic functioning of concurrently active back muscles during sustained contractions can be represented by "fatigue patterns" created by the spectral EMG shift occurring in the different back muscles. Differences in these patterns may be the result of functional disturbances in back extensor muscles associated with LBP.

The study presented here is a continuation of these preliminary results. It describes the reliability of the assessment procedure and the clinical validation of the technique for discriminating muscle function in chronic LBP patients and normal control subjects.

Background

The earliest studies of back muscle fatigue investigated the relation between the EMG signals of the lumbar musculature of the back during static contractions. Morioka[40] observed a decrease in the EMG signal amplitude and an increase in low frequency potentials while subjects performed static lifting. In a similar study, Chapman and Troup[10] observed a decrease in the total electrical activity during the onset of fatigue symptoms. They attributed this decrease to a transfer of activity to other muscles in the trunk; however, in a subsequent study,[61] they found that the integrated EMG signal increased with fatigue. Okada, Kogi, and Ishii[46] studied back muscle endurance for a series of incremented loads sustained until the onset of muscular pain. The EMG signal initially increased with fatigue and then, during the contraction's later stages, it decreased; suggesting that subjects may have altered their posture to alleviate muscular pain. They also observed that the low frequency components of the EMG power spectrum increased consistently throughout the contraction. Andersson et al[3] recorded from many more electrode sites and repeated tests for different trunk postures. Their results showed increases in EMG signal amplitudes at all detection sites of the back when the angle of forward flexion was increased. They observed significant EMG spectral changes in those test positions that also produced high EMG signal amplitudes. Furthermore, they found that an increased level of EMG activity was always accompanied by an increased rate of change of the EMG power spectrum.

A few quantitative studies of back muscle fatigue have included LBP patients. DeVries[17] compared chronic low-back pain patients with normal controls during prolonged stance. Those subjects who developed pain during the experiment showed increased back EMG activity, whereas those who experienced no pain had decreased EMG activity. Similar results were reported by Jayasinghe et al.[26] Recently, Jorgensen and Nicolaisen[28] reported that subjects with prior serious attacks of LBP had less endurance capacity, but similar strength, in the trunk extensors when compared with pain-free control subjects. Endurance was determined by monitoring the time to exhaustion during the performance of sustained, isometric back extension.

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Although these studies have made some contributions to our understanding of back muscle function, they share two major flaws. First, previously reported work on back muscle fatigue has relied on analysis of EMG signals detected from only one or two muscle sites. The back extensor muscles, as described by physiologic and kinesiologic reports, cannot be accurately assessed as one continuous muscle mass. Rather, they are better described as a mosaic of distinct functional units corresponding to the different back muscle groups.[7] Thus, the meager data from only one or two muscle sites can only provide a reduced and oversimplified view of the muscular performance and would preclude meaningful quantification of muscle imbalances or deficits that may occur with LBP. Secondly, the majority of previous studies have relied on the amplitude of the EMG signal as an indicator of muscle fatigue, despite the fact that it is sensitive to many factors other than muscle activity.[14,39] Furthermore, EMG amplitude measures are subject to random variations among separate tests even when all experimental conditions are maintained constant.[58]

A preferred method, which has gained wider acceptance during recent years, is to monitor the modifications to the frequency domain properties of the EMG signal that are associated with fatigue. This technique monitors the compression of the power density spectrum toward lower frequencies as a sustained isometric contraction progresses.[34,41,45,57,59] This well-documented phenomenon has been theoretically and empirically correlated with biochemical and physiologic processes associated with muscle fatigue.[14] Analysis of this time-dependent modification to the EMG signal has led to the characterization of muscle fatigue as a continuous process rather than a single-point event related to failure of force production. From a practical perspective, EMG spectral measurements are more independent of subject motivation and require less demanding contractions, since the subject is not required to sustain a contraction until he can no longer maintain a targeted force level.

Despite a renewed interest in EMG spectral measurements and recent technical advances in quantifying spectral fatigue parameters,[21,38] surprisingly little attention has been focused on demonstrating the applicability of power spectral measurement techniques to specific muscle disorders. In response to this need, we developed a Back Analysis System (BAS) to implement this technique for evaluating lumbar muscle function in individuals with LBP.[49,52]

SYSTEM DESCRIPTION

Figure 1 is a schematic of the computer-aided BAS system used in this study. Its individual components are described below in more detail.

Postural Restraint/Force Acquisition Device

The experiments reported in this paper used the restraint device presented in Figure 2. The restraint device stabilizes the pelvis and lower limbs during standing and was designed to ensure that the sustained, isometric muscle activity observed was actually associated with the flexion torque being monitored. The design of the restraint device is crucial. It should isolate the force contribution of the back muscles during extension. Failure to do so can compromise the validity of the tests. The device was constructed from commercially available hexagonal tubing and couplings that allow the apparatus to be tailored to a large variety of subjects and body positions. Specially contoured, adjustable front and rear restraining pads hold the subject securely in a posterior pelvic tilt, and adjustable knee pads provide patellar tendon-bearing surfaces. The torque generated during an isometric contraction is measured with a nylon harness positioned across the shoulder region of the back and attached to two load cells having a compliance of 2.7 μm/kg. Differences in the force computed from the two transducers can provide a measure of the degree to which a "pull" is symmetric or includes a torsional component. A maximal volun-

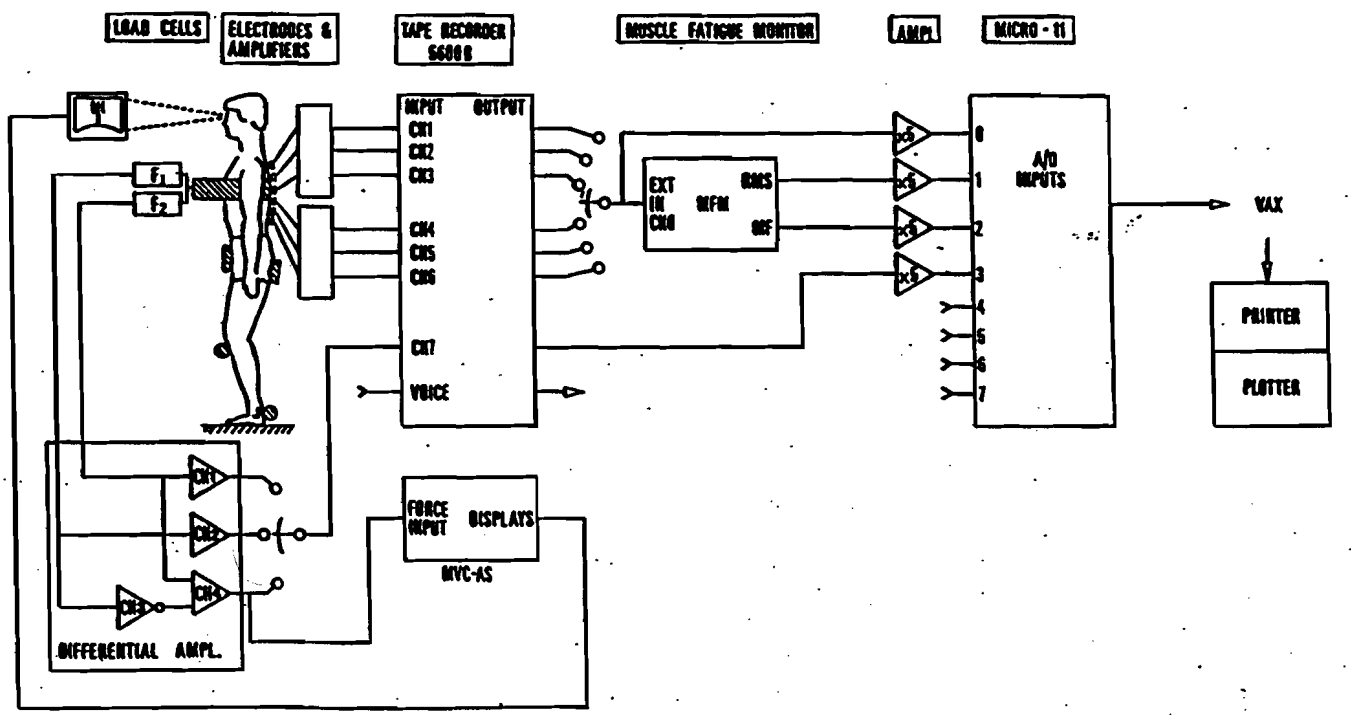

Fig 1. A schematic of the Back Analysis System (BAS) for evaluating fatigue-related information from lumbar muscles using spectral EMG analysis.

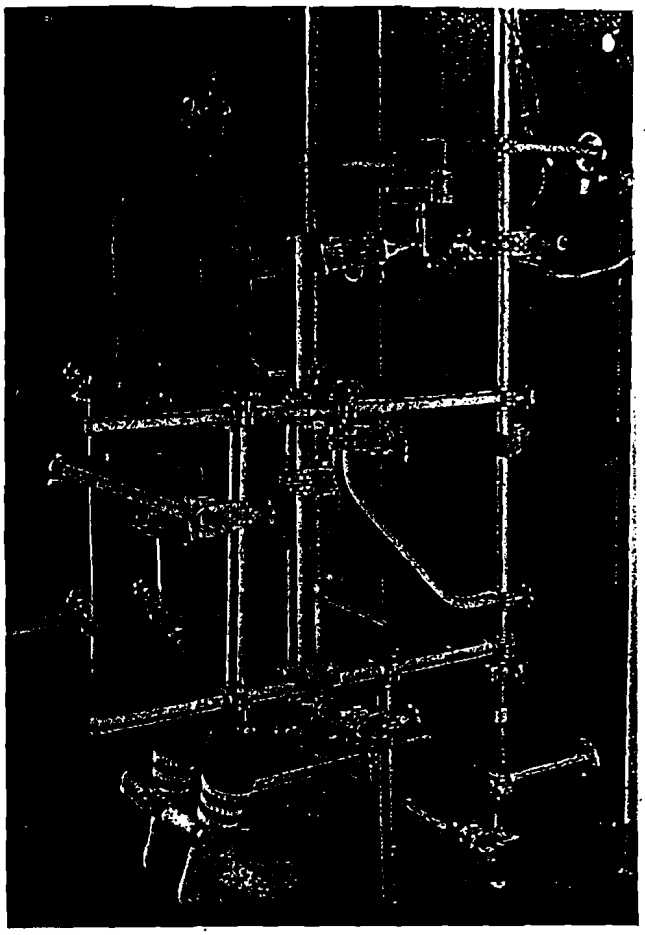

Fig 2. The postural restraint device and force acquisition system used in this study.

tary contraction (MVC) acquisition system also was incorporated in the device via a circuit that calculates the MVC by averaging force values over an adjustable window. The MVC force value is stored and can be used to set a visual display to the desired percentage of the MVC. The feedback display is necessary to help the subject maintain a constant level of contractile force.

Signal Detection and Recording

Six active, bipolar surface electrodes, similar to ones described by De Luca et al,[15] were used for signal detection. These electrodes were aligned on the muscle so that the parallel silver bars that make up the detection surface are perpendicular to the muscle fibers. Electrodes have a gain of 10 and a −3 dB bandwidth of 20 to 550 Hz, with a rolloff of 12 dB/octave. The six channels of EMG signals were further amplified to achieve an output amplitude of 1 to 2 V peak-to-peak. This data was recorded on 1/2-inch magnetic tape (Honeywell 5600B FM tape recorder, Honeywell, Denver, CO) at a tape speed of 7.5 inches/sec to provide a bandwidth of 1.25 kHz. The force transducer outputs were also recorded on FM tape.

Muscle Fatigue Monitor

This device calculates the median frequency (MF) and the root-mean-square (RMS) of the EMG signal in real-time using analog circuitry similar to that described in a previous publication.[21] Median frequency is defined as the frequency that separates the power density spectrum into halves of equal power. This parameter has been shown by Stulen and De Luca[59] to provide a reliable, consistent, and unbiased measure of the frequency shift of the EMG signal associated with muscle fatigue during sustained, constant-force contractions. The electronic circuitry used to measure muscle fatigue was designed to be integrated with an on-board pocket computer (Sharp PC 1500/CE150; Sharp Electronics, Paramus, NJ) for device control and hard-copy of both graphs and texts. A more recent version of this device has been integrated with an IBM-PC compatible computer, thereby offering greater performance capability.[22] The bandwidth of the input stage is 20 to 550 Hz at a −3 dB with a filter slope of −12 dB/octave. To further ensure error-free EMG detection, a component of the circuit continuously monitors the signal output from the electrodes. Signal levels above and below a preset amplitude range (±20 μV to ±4 mV), or signals containing power line-induced interference, trigger an audible alarm to inform the operator of an error condition.

Data Analysis

Following each experiment, the EMG signals were individually processed by analog circuitry using the muscle fatigue monitor (MFM) to compute the MF value of the signal. This parameter and the force data were further amplified and simultaneously digitized at 100 Hz using a PDP-11/23 microcomputer. A sampling rate of 100 Hz was selected to satisfy the Nyquist criterion, since the fluctuations of the MF and force were below 40 Hz due to the characteristics of the MFM.

The digitized MF records for each of the six electrode locations were simultaneously plotted as a function of time (Figure 3). The software algorithm used a moving average filter with a 2-second window (200 samples) to reduce high-frequency fluctuations in the data. These high-frequency fluctuations are due to the stochastic nature of the EMG signal and are not of interest when observing the time-dependent changes of the MF associated with fatigue. The plot is divided into two sections, corresponding to the left and right sides of the back. The vertical axis represents the MF and the horizontal axis represents time. Quantitative measurements of the time rate of change of the MF of the individual curves is also provided. Linear regression using the method of least squares was used to compute the rate of decrease of the MF. A linear fit to the unfiltered data was chosen since it most consistently represented the time-dependent change of the MF as monitored from the back extensor muscles. The following parameters were investigated: 1) The initial median frequency (IMF) of the curve. This value was obtained by calculating the y-intercept of a straight line regression fit by a "least-squares" method to the MF data. 2) The slope of the regression line for the MF—time function, calculated over the duration of the contraction.

METHODS

A series of preliminary measurements and procedures were followed before the reliability studies and back pain studies described below. To determine the electrode placement, the locations of the motor points were first identified using low-level pulsed electrical stimulation. Six active surface electrodes were positioned bilaterally on the longissimus thoracis muscle at the L1 spinal level, on the iliocostalis lumborum muscle at the L2 spinal level, and on the multifidis muscle at the L5 spinal level (Figure 4). Care was taken not to place the electrodes on the identified motor points to avoid unwanted signal effects related to the innervation zone of the muscle.[51] Subjects then were positioned in the back testing apparatus and, following several practice trials, a maximal voluntary contraction was determined.

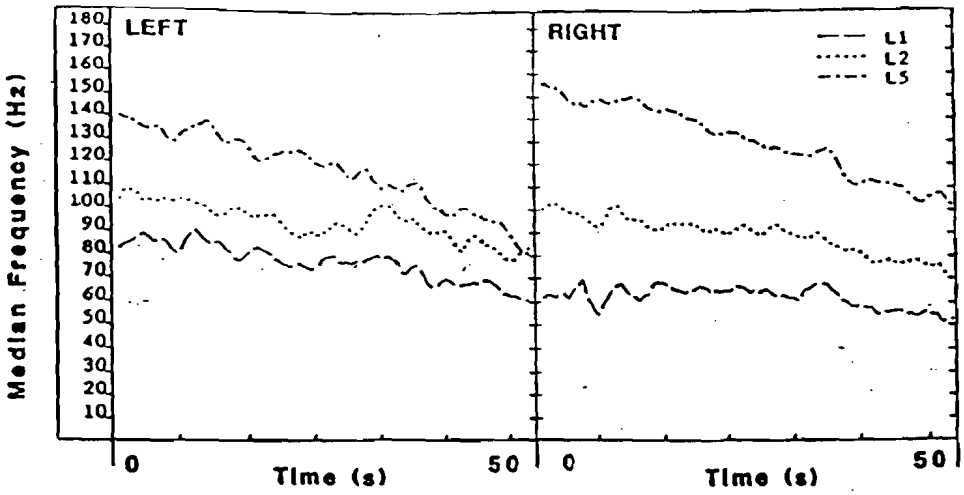

Fig 3. Median frequency plots as a function of contraction duration are depicted for the six lumbar back muscles tested at 80% MVC. Curves are arranged in groups of three, corresponding to left- and right-sided muscles.

Reliability Assessment. Four subjects participated in the experiment to determine the repeatability of the back assessment procedure. All were male subjects with no reported previous history of back pain. The subjects' ages ranged from 20 to 38 years (24.1 ± 5.3). The experimental protocol consisted of two repeated 80% MVC contractions sustained for 30 seconds, with a 15-minute rest period between contractions. The subject was not removed from the apparatus between trials and the electrodes were kept in the same location for each trial.

A single-factor analysis of variance (ANOVA) procedure was used to compute the reliability estimate.[1] This procedure calculates the ratio of the variance of the true measurement divided by the sum of the variance of the true measurement and the variance of the experimental "error" associated with each measurement. Thus, for example, a reliability estimate of 0.8 indicates that on the average, 20% of the variability in the measurements can be attributed to experimental error. Data from all six electrode sites were included in the analysis.

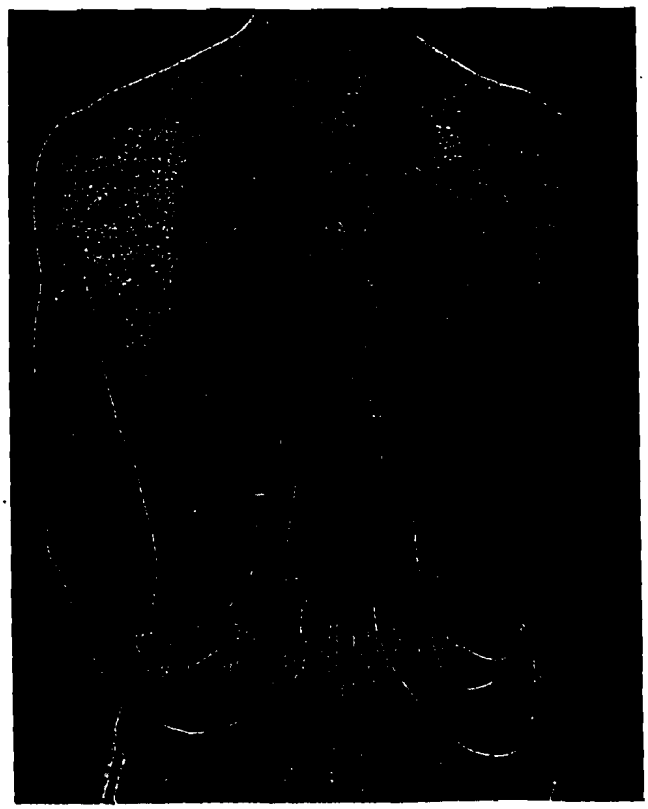

Fig 4. A typical location of six EMG surface electrodes are shown corresponding to the approximate bilateral locations of longissimus (L1), iliocostalis (L2), and multifidus (L5) muscles.

Back Pain Study. Twelve patients (32.5 ± 13.6 years) with a history of chronic back pain were evaluated and compared with 12 healthy subjects (27.4 ± 6.8 years). All participants were male and right-handed. Grouping was determined by the presence or absence of a documented history of chronic back pain. Chronicity was defined as persistent or frequently recurring pain over a period of at least 1 year. The average duration of a LBP history was 5.2 years for the patient group (range, 1.5–13 years). Patients with acute exacerbation of back pain were excluded. Those subjects with previous back surgery or radiographic evidence of structural disorders of the spine also were excluded.

Following the determination of the subject's MVC and a 5-minute rest, the subject performed three constant-force contractions at 40% MVC, 60% MVC, and 80% MVC for a duration not to exceed 1 minute. A 15-minute rest period between contractions allowed for full recovery of the MF parameters.[53]

A three-factor ANOVA was performed (BMDP, 1987: Health Sciences Computing Facility, UCLA, California) to investigate whether MF parameters differed between LBP and control subjects, and to discern the interactive effects of load and muscle site on the MF parameters. One grouping factor was used to classify back patients and controls and another grouping factor distinguished the results obtained from the left and right sides of the back. The trial factors represented the three contraction levels performed during the experiments (ie, 40% MVC, 60% MVC, and 80% MVC). This analysis was repeated for each of the three muscle groups investigated.

A two-group, stepwise discriminant analysis was conducted to investigate the ability of the MF parameters to discriminate LBP from control groups.[1] This analysis included the IMF and slope parameters from the six electrode sites and was performed separately for each of the three force contraction levels. All parameters were first prescreened for multicollinearity by computing a correlation matrix and eliminating those variables that were highly correlated (a correlation coefficient greater than 0.80). For more subtle patterns of correlation, no variable was entered into the classification function unless it could pass a tolerance limit of 0.01.

We also tested the hypothesis that multichannel electrode configurations are better able to characterize the complex performance of the back muscles and are better predictors of LBP than proce-

dures relying on fewer electrode configurations. This test was implemented by conducting the discriminant analysis separately for each of the bilateral recordings at L1, L2, and L5. All three contraction force levels were analyzed in this way for a total of nine separate classification functions. The percent correct classification resulting from each of these functions then were compared with similar calculations that combined data from all six electrode sites.

RESULTS

The reliability estimates calculated from the ANOVA analysis were 0.98 and 0.94 for the IMF and MF slope, respectively. This indicates that the error associated with the measurement of IMF is 2%, and with the measurement of MF slope, 6%. The results of the reliability trials are represented in Figures 5 and 6 for estimates of IMF and MF slope, respectively. Each figure presents a scatter plot of the data for repeated trials (Trial 1, Trial 2) executed 15 minutes apart. The unity-slope, zero-intercept line representing perfect reliability falls within the 95% confidence limits of the data.

For the LBP and control subjects, multiple unpaired Student *t* tests were performed to determine if group differences existed with respect to age, weight, height, or maximum voluntary contraction level. The results are summarized in Table 1. Of the anthropometric group parameters studied, significant group differences were present solely with regard to body weight ($P > 0.05$). The LBP group had a mean body weight that was 13.7 kg (30 pounds) higher than the control group. No isometric strength differences were observed as measured by the MVC values.

The summary statistics for IMF and MF slope measures from the LBP study are plotted in Figures 7 and 8, respectively. Each figure contains data for the longissimus (L1), iliocostalis (L2), and multifidus (L5) muscle detection sites. In each plot, the data from the left and right sides of the back are further subdivided, as are the force levels of the contraction. This graphical representation identifies a number of important results, which are summarized below:

1) The IMF decreases for increasing levels of contractile force. This relationship was present for all muscle groups tested and in both LBP subjects and control subjects.

2) Left–right differences are present in both LBP and control subjects for each of the parameters studied.

3) For any given contraction force level, the average IMF and MF slope are greater at the L5 detection site than at either the L1 or L2 detection sites for both LBP and control subjects.

4) The IMF is significantly lower for LBP subjects compared with control subjects across all force levels of left and right longissimus, L1 electrode location.

5) Low-back pain subjects exhibit significantly higher MF slope values than control subjects at 80% MVC for the L2 and L5 recording sites.

The results of the discriminant analyses are presented in Tables 2 and 3. A maximum of six noncorrelated variables were entered into these discriminant analyses. For the data in which all muscle sites were included in the analysis, the best classification results corresponded to trials conducted at 40% MVC and 80% MVC.

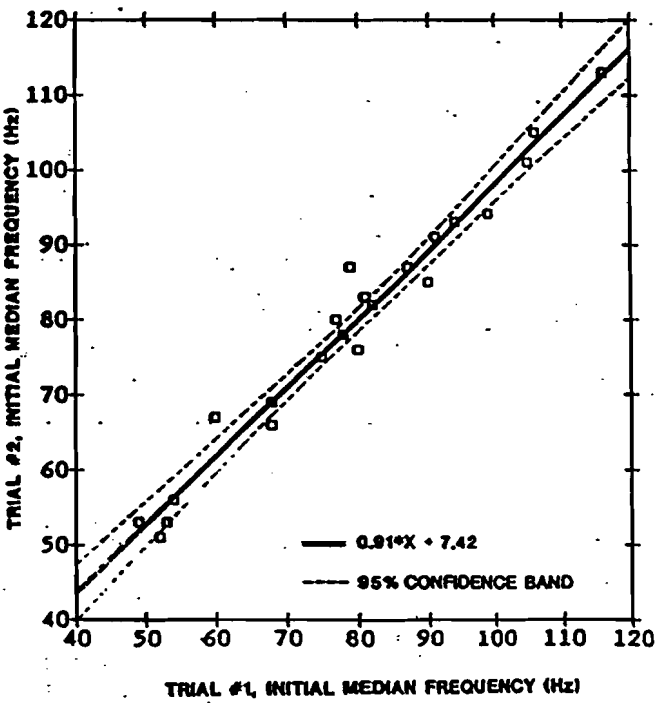

Fig 5. A scatter plot of initial median frequency (IMF) data for repeated trials (Trial 1, Trial 2) tested 15 minutes apart. Data is for all six electrode sites in four subjects. A "least-squares" linear regression and 95% confidence band are indicated.

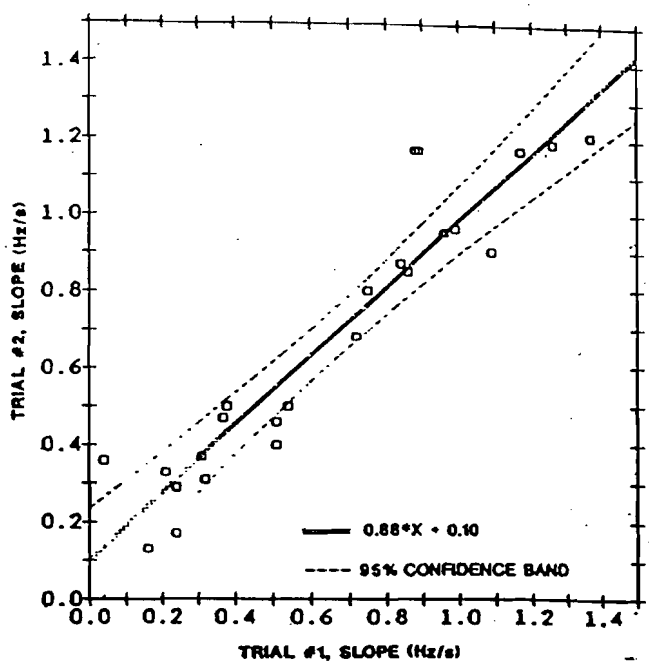

Fig 6. A scatter plot of median frequency slope data for repeated trials (Trial 1, Trial 2) tested 15 minutes apart. Data is for all six electrode sites in four subjects. A "least-squares" linear regression and 95% confidence band are indicated.

Table 1. Characteristics of Subjects: Mean Values (Standard Deviation)

Characteristic	Control subjects (n = 12)	LBP subjects (n = 12)
Age (yrs)	27.3 (7.2)	32.4 (14.3)
Height (cm)	169.8 (17.0)	177.9 (10.1)
Weight (kg)	71.1 (5.9)	84.8 (14.3)*
MVC level (kg)	105.8 (24.9)	112.1 (34.5)

LBP = low-back pain; MVC = maximal voluntary contraction.
*$P < 0.05$, by Student unpaired two-tailed *t* test.

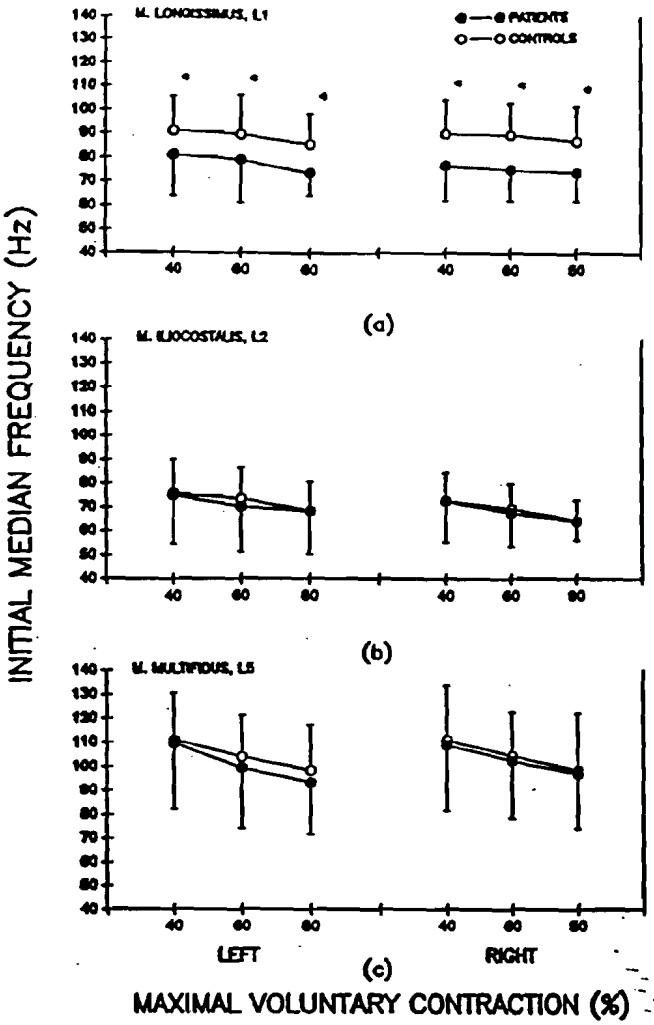

Fig 7. Mean initial median frequency (IMF) for left and right muscle groups for three %MVC forces tested in patients (●) and control (O) subjects. Plots are presented for the (a) longissimus (L1), (b) iliocostalis (L2), and (c) multifidus (L5) muscles. *$P < 0.05$ for significance of difference between means.

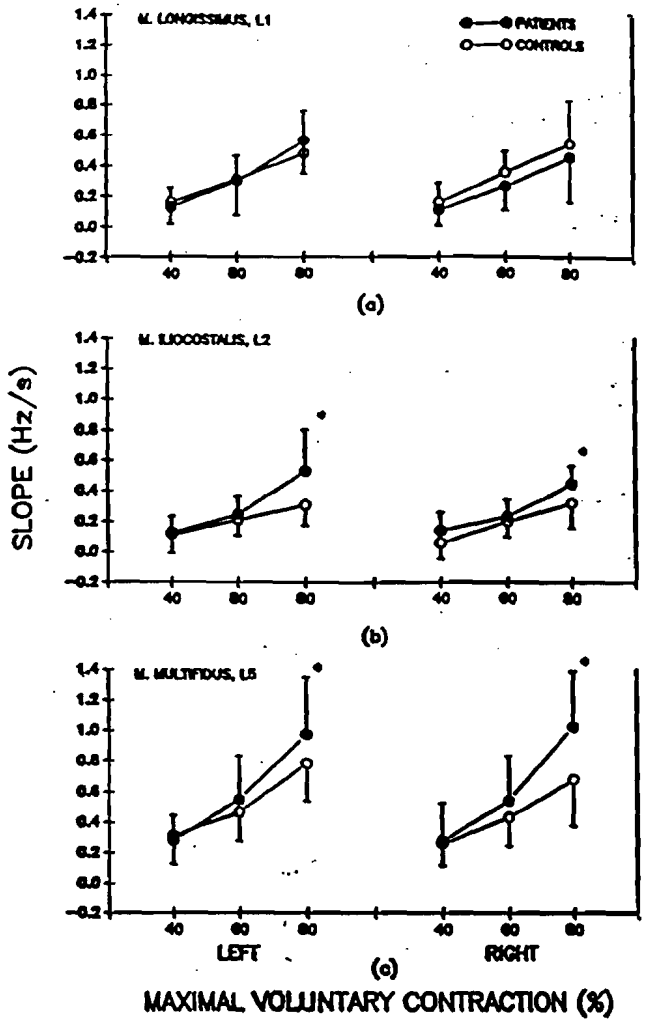

Fig 8. Mean slope of median frequency for left and right muscle groups for three %MVC forces tested in patients (●) and control (O) subjects. Plots are presented for the (a) longissimus (L1), (b) iliocostalis (L2), and (c) multifidus (L5) muscles. *$P < 0.05$ for significance of difference between means.

The analysis correctly predicted 92% of the LBP group and 82% of the control group for the 40% MVC trials. This corresponds to one false positive and two false negative identifications. Similarly, for the 80% MVC trials, the analysis correctly identified 84% of the LBP group and 91% of the control group. This corresponds to two false positive and one false negative classifications. The percent correct classification by the discriminant analysis therefore was nearly the same for identifying LBP subjects as it was for identifying control subjects. Performance suffered somewhat for the 60% MVC trials, particularly when identifying control subjects. The variables entered into the discriminant function were primarily the IMF at L1 and the MF slope measurements at L2 and/or L5. The results show a higher percentage of correct classification for analysis in which all electrode sites are included in the discriminant analysis (Table 2) than for the analysis in which muscle groups were treated separately (Table 3). The data analyzed for separate muscle groups demonstrated only one combination of lumbar level and contractile level (L5 at 80% MVC) in which LBP and control groups were identified correctly for at least 75% of the cases. (R)Slope and (L)IMF, in that order, were the discriminating variables for this two-group analysis.

Table 2. Results from Discriminant Analyses—All Lumbar Levels

Contractile level (percent MVC)	Percent correct classification: NLBP (n = 12) (%)	Percent correct classification: LBP (n = 12) (%)	Variables used in classification (in order)
40	92	82	(R)IMF,L1 (R)SLOPE,L2 (L)SLOPE,L2 (L)SLOPE,L1 (L)IMF,L2 (L)IMF,L1
60	67	75	(R)IMF,L1 (L)SLOPE,L1
80	84	91	(L)IMF,L1 (R)SLOPE,L5 (L)SLOPE,L5 (L)SLOPE,L2

(R), (L) = right side, left side; LBP = with low-back pain; NLBP = without low-back pain; IMF = initial median frequency; SLOPE = median frequency slope.

Table 3. Results from Discriminant Analyses—Individual Lumbar Levels

Lumbar level	Contractile level (Percent MVC)	Percent correct classification: NLBP (n = 12)	Percent correct classification: LBP (n = 12)	Variables used in classification
L1	80	70	73	(L)IMF
	60	70	75	(R)IMF, (L)SLOPE
	40	70	75	(R)IMF, (L)SLOPE
L2	80	83	70	(L)SLOPE, (R)SLOPE
	40	75	70	(R)SLOPE, (L)SLOPE
L5	80	75	75	(R)SLOPE, (L)IMF
	60	75	50	(R)SLOPE

(R), (L) = right side, left side; LBP = with low-back pain, NLBP = without low-back pain; IMF = initial median frequency; SLOPE = median frequency slope.

DISCUSSION

Reliability

Any new technique that is introduced as a means of evaluation must have its reliability demonstrated. This is particularly true for surface EMG techniques because there is a need to overcome the bias that has evolved from a history of unfulfilled expectations. Two issues that effect reliability must be addressed: the specificity of the muscle generating the detected signal and the location of the surface electrode on the muscle.

The issue of muscle specificity associated with EMG detection is relevant to this and other studies that attempt to extrapolate results to characterize specific muscle groups. The question often raised is whether the detected EMG signal includes myoelectric activity from neighboring, or even distant muscles. Such a signal, which may be erroneously interpreted as generated by fibers near the electrode, is usually referred to as "cross-talk". We minimized cross-talk by carefully selecting the most superficial back extensor muscles with relatively large muscle cross sections. A number of dissected cadavers were studied to help select electrode locations where the major erector spinae muscle groups were not overlapped by neighboring muscles. The selective use of electrical stimulation and palpation in subjects prior to electrode placement helped to isolate these muscle groups. In addition, the electrode's bipolar design, small (1.0 cm) interelectrode separation, and incorporation of parallel detection bars oriented perpendicular to the muscle fibers all favored high muscle selectivity.[7] Despite these precautions, it still can be argued that some cross-talk may still have been present. Recent studies conducted in our laboratory and elsewhere[16,19] have shed some light on this issue. These reports have concluded that up to 16.6% of a signal (peak-to-peak amplitude) detected above a neighboring muscle may be due to cross-talk rather than to activation of the muscle below the detecting electrode. These studies were conducted on leg muscles only; however, similar investigations are now being conducted on lower back muscles.

The second issue that effects reliability pertains to the location of the electrode on the muscle. This is generally not a problem when the amplitude of the EMG signal is used as a measure of muscle activity. It does become important when EMG frequency parameters are used. Our past work[51] has shown that the spectrum of the EMG signal is sensitive to the placement of the electrode in the vicinity of the motor point. Our empirical findings are consistent with the predictions of theoretical considerations. Thus, we chose to place the detection electrode away from the motor point and highly recommend this procedure for all surface EMG detection.

The alternative to our method of electrode placement is to locate the electrodes at fixed distances from vertebral or other bony landmarks. We did not select this methodology, nor do we advise it, since this procedure offers little possibility of a functional interpretation of the EMG and will in all likelihood contribute greater variability to the data. When fixed bony landmarks are used, the individual differences in body size and anatomy will determine which muscle groups are near the detection electrodes as well as the position of the electrode relative to the motor point.

Few published EMG studies of back function have examined the reproducibility of the measurements.[30] In this study, estimates of measurement error for IMF and MF slope were 2% and 6%, respectively; well within the limits of acceptability for clinical applications. In this experiment, the electrodes were not physically removed between trials, hence the measurement errors could be caused mainly by two factors: a slight movement of the electrode on the skin surface and a variation in the force output of the back muscles in the two contractions. Previous studies have provided a theoretical basis[4,7,34] for, and an empirical verification[51] of, the sensitivity of EMG signal parameters to minor displacements of the electrode. This concern is particularly relevant when spectral parameters of the EMG signal are calculated because, although the average amplitude of the EMG signal may remain insignificantly affected, the frequency spectrum can be altered due to variations in the spatial filtering properties of the tissue between the active fibers and the detection electrode. The second possible cause of the error, the variation of force output, will effect the MF values by altering the contribution of motor units to the EMG signal and thus modifying the frequency spectrum of the EMG signal.

The precautions taken in locating the electrode on the muscle, especially distancing the electrode from the motor points, renders the IMF in this instance primarily related to the average cross-sectional area of the muscle fibers being detected.[14] This relatively stable determinant of IMF can be contrasted with the more unpredictable processes of metabolite production, re-utilization and vascular flow, which are the primary determinants of the MF slope.[14] These different physiologic correlates may explain the differences in reliability between IMF and MF slope parameters. Further studies are planned to measure the reliability of this technique within the LBP population.

Normal Back Muscle Performance

One consistent finding observed in this study was the decreasing trend in IMF with increasing force level of the contraction. This finding is not seen in limb muscles, where the opposite behavior has been universally reported.[14,53] The decreasing trend in IMF with increasing force level seen in Figures 7A to C suggests that smaller sized muscle fibers are recruited at high contraction force levels. This observation is consistent with recent reports from autopsy and biopsy studies that the later recruited, Type II muscle fibers have smaller mean diameters than Type I fibers in human trunk extensor muscles.[6,55,60] This is an obvious, and rather profound, departure from the norm for almost all other human skeletal muscles.[24]

A trend toward increasing fatigue rates with increasing force of contraction also was characteristic of the lower back muscles tested. This observation also is commonly found in limb muscle studies,[7,11,53] and is most likely due to the increased rate of metabolite accumulation associated with an increasing contractile force level. This can occur when a predominance of Type II motor units are active and the pressure within the muscle reduces the blood flow to the muscle.[14]

Median frequency parameters displayed inconsistent left-to-right differences that did not appear to be related to the presence or absence of LBP. It is entirely reasonable to expect that left-to-right differences may be more representative of patients who have a unilateral mechanical deficit,[27] but none of our patients suffered from skeletal deficits in the back. Furthermore, we did not overtly control the amount of trunk rotation that may have been introduced into the trunk extension task. Inconsistent trunk rotation could result in an assymmetrical "pull" that would produce inconsistent contralateral results. Our findings are in agreement with those obtained by Jayasinghe et al and Collins et al,[12,26] who observed inconsistent left-to-right differences in both patients and healthy individuals, especially during exertive tasks. In contrast, Hoyt et al[25] observed larger absolute left-to-right differences in activity levels in back patients as compared with control subjects. Cram and Steger[13] observed left-right imbalances in patients, but they did not test healthy subjects.

The differences in the MF parameters for the three lumbar levels selected may be directly related to differences in mechanical advantage and other biomechanical factors that determine the amount of force a muscle needs to generate. Previous biomechanical modeling studies of the lumbar musculature during standing or slight forward flexion have predicted greater forces for the lower lumbar musculature than the upper lumbar musculature.[63] This difference in force distribution may explain the higher MF slopes observed at L5 than at either L1 or L2 electrode sites. Similarly, the higher mean IMF values at L5 compared with upper lumbar muscles may result from larger muscle fiber cross-sectional areas often associated with increased force-generating capacity. Reported differences in muscle function for the iliocostalis, longissimus, and multifidus muscles[35] may also account for the differences we observed in the MF data as a function of muscle site.

Muscle Performance Associated with LBP

The results of the two-group discriminant analysis provided, to our knowledge, the first examples in which the clinical validation of spectral estimates of muscle fatigue has been quantified. The MF parameters correctly separated the LBP population from the control population with only one or two false identifications. These findings imply that LBP subjects and individuals without back pain can be classified correctly on the basis of an objective measure of muscle function that does not directly involve the psychological aspects of performance. Furthermore, in comparing the differences in percent classification between Tables 2 and 3, it is apparent that the array of electrodes sampling multiple sites provided more accurate results than any individual bilateral site. There is no definitive explanation for the poor categorization by the 60% MVC data. Perhaps the muscle deficits are more apparent at the extremes of the contractile force range, where all of the muscle fibers are active.

Of the two miscategorized subjects, there was one consistent false positive. Follow-up interviews and reexamination of subject records revealed that in the year following these experiments, this subject developed a nonspecific LBP disorder that has continued for several months. The one consistent false negative case identified by the analysis was a college varsity swimmer who had one of the highest MVC values (176.5 kg) of all of the LBP subjects tested. It is possible that he may have been misclassified because of his athletic background, which may have resulted in highly endurant back muscles despite a chronic LBP history. This possibility reinforces our belief that separate data bases are needed for different subcategories of LBP patients and different levels of conditioning.

In comparing the IMF values obtained from patients and controls, significant group differences were observed solely for data obtained from the longissimus muscle site at spinal level L1 (Figure 7A). At this level, the mean IMF values were found to be 10 to 15% higher for controls than for patients, and this difference was consistent at all three %MVC levels. Considering the significance of the IMF discussed previously, this observed difference may be due to the possibility that 1) the longissimus muscle at level L1 has a greater proportion of smaller diameter, Type II fibers in patients than in controls, and/or 2) LBP patients have decreased muscle fiber sizes, possibly resulting from generalized muscle atrophy associated with disuse.

Significant differences for MF slope measures were reported between the LBP and control groups for specific electrode sites and force contraction levels. The data from the multifidus (L5), and iliocostalis (L2) muscle sites demonstrated consistently higher fatigue rates at the 80% MVC level for LBP subjects compared with control subjects. Significant group differences of MF slope were not present at the 40% or 60% MVC levels. No significant group differences were discerned at either of the three contraction levels for the longissimus (L1) muscle. These findings support our previously stated concern that the back extensor muscles cannot be assessed properly by relying on only one or two sites for EMG signal detection. It appears that the behavior of the MF is muscle-specific as well as load-dependent. These factors may explain the poor reliability and conflicting data that have characterized previous attempts at studying back muscle fatigue by surface EMG. The higher rate of fatigue observed in the LBP group as compared with the control group in our study is consistent with previous observations made by others.[17,26,28,31,44] Again, this observation may be attributed to a greater proportion of Type II muscle fibers in LBP patients than control subjects. Others have theorized that excessive back muscle fatigue is associated with high precontraction metabolite levels resulting from persistent muscle spasm and prolonged muscle tension.[5] A further possibility that is particularly relevant to our study is that some back muscles may contribute a proportionately greater component of force than other back muscles when a chronic LBP condition is present. This likelihood precludes identifying whether the fatigue-related difference observed between LBP and control subjects is due to a physiologic change in the muscle (ie, a different metabolic profile) or whether the muscles are coordinated differently so that the actual muscle forces that comprise the net extensor moment of the trunk are distributed unequally. We are developing a mathematical model[32,42,43] to estimate the individual force contributions of trunk muscles and thereby gain a clearer understanding of the relation between muscle force distribution and muscle fatigue.

From the characteristics of the two groups selected, it appears that these groups were closely matched except for body weight. The higher weight of the LBP subjects may reflect the lower level of physical activity that is commonly associated with chronic LBP. Isometric back muscle strength was not affected, since back patients were able to generate MVC force levels similar to that of control subjects. This result is consistent with the observation of others,[28,37] who also found no strength deficits associated with LBP. There are, however, conflicting reports from a number of other studies that identify back extensor weakness associated with LBP.[1,8] This discrepency may be attributable to the different dynamometers used and whether pain inhibited the response at the time of testing.

We deliberately selected patients in remission to increase the possibility that differences in muscular performance, if present, were the result of physiologic factors rather than psychogenic fac-

tors. We therefore would expect that an individual with acute LBP at the time of testing would have very different MF parameters than the LBP subjects tested in this study. Similarly, athletes or individuals who have exercised regularly despite having a LBP disorder also should be categorized separately. One possible alternative to requiring separate subcategories of LBP patients is to look toward more effective ways of normalizing the EMG tests rather than relying only on the subject's MVC value. The MVC value may change considerably as a result of pain or lack of motivation and may not always be a true representation of the force-generating capacity of an individual's muscle. One possibility is to express the MVC as a percentage of body weight.[30] This would normalize muscle performance with respect to an anthropometric measure that is assumed to be related to muscle strength. In this instance, a normative data base of MF parameters plotted as a function of the contraction force level (expressed as a percentage of body weight) would need to be determined. Unfortunately, body weight has been shown to be a poor predictor of muscle strength.[44] Other anthropometric, or even biomechanical, data may provide a better approximation of the force-generating capacity of the muscle. Future work is needed to investigate these possibilities for improving normalization procedures for clinical applications to a variety of populations.

CONCLUSION

A new back assessment procedure is presented that uses an array of surface-detected EMG signals to objectively characterize the muscle fatigue insufficiencies associated with nonstructural, chronic LBP. The technique was shown to be highly reliable by repeated trials conducted on a group of healthy subjects ($n = 4$). Tests involving chronic LBP ($n = 12$) and healthy control subjects ($n = 12$) demonstrated that specific median frequency parameters of the EMG correctly classified individuals into the LBP and control categories by a discriminant analysis procedure. The initial value of the median frequency and the slope of the median frequency were significantly different between LBP and control subjects ($P < 0.05$) for specific target force contraction levels and muscle groups tested.

These results validate the use of surface EMG spectral parameters as an objective measure of back muscle function in chronic LBP patients.

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