

## Development of new protocols and analysis procedures for the assessment of LBP by surface EMG techniques

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**Abstract**—Spectral parameters of the surface electromyographic (EMG) signal from lumbar back muscles assessed during a fatiguing isometric contraction can be used to classify different categories of low back pain (LBP) subjects and control subjects without LBP. In the test protocol currently used at the NeuroMuscular Research Center at Boston University, subjects contract their back muscles at 80% of their maximal voluntary contraction (MVC) force. This fatigue-based protocol has been successfully applied to persons with subacute or chronic LBP; those in acute pain, however, have not been included because of their inability to perform a maximal exertion. In this paper we will examine the force sensitivity of the currently used EMG parameters and also give an overview of some of our efforts to develop new test procedures. Our goal is to develop force-insensitive surface EMG parameters that can be used for classification purposes in populations of subjects

who develop low trunk extension forces. In addition, the development of a model to predict MVC from anthropometrical measurements will be presented.

**Key words:** *back muscle assessment, classification, EMG, low back pain.*

### INTRODUCTION

Low back pain (LBP) disorders are among the most common musculoskeletal complaints of persons seeking medical care. The incidence of LBP has been reported to be second only to that of the common cold (1), and in the population at large, there is an 80 percent chance that a person will seek medical care for a LBP disorder prior to age 55 (2). The costs to society associated with this condition are staggering. The annual medical treatment cost of LBP in the U.S. was recently estimated to be \$24 billion (3). If compensatory costs and time lost at work are included, the cost would increase to \$72 billion (4). These disorders have proven difficult to evaluate and treat. Thus, the availability of a technique to identify a physical abnormality associated with LBP would be a milestone in terms of assessing the muscular component to LBP.

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The most common way of assessing back muscle function has been through different strength and/or range-of-motion measurements. Strength is an important aspect of muscular function, which has been shown to be related to LBP (5-7). A variety of back dynamometers and range-of-motion tools are available that estimate different types of back muscle performance parameters (8-20); however, the usefulness of strength measurements to discriminate on an individual basis has been criticized due to the wide and overlapping ranges between nonimpaired and LBP subject behavior (20). Marras et al. (21-23) related kinematic parameters extracted during voluntary trunk movements to age, gender, and degree of trunk dysfunction in LBP subjects. In tasks where subjects repeatedly flex and extend their trunk at their maximum preferred speed, kinematic parameters such as movement velocity and acceleration were found to vary with age, gender, and type of back impairment.

The use of surface EMG techniques has played a major role in our understanding of the functional interactions between individual trunk muscles, in both nondisabled and LBP subjects, during specific postures and movements (24-30). A parameter of the surface EMG signal that has been used to assess the function of lumbar muscles is the median frequency (MF). It is commonly associated with fatigability of the back muscles, since it is usually monitored during some type of endurance task. The technique is based on the phenomenon of the compression of the power density spectrum of the EMG signal toward lower frequencies during a sustained contraction. This change in the EMG signal has been associated with muscle metabolic correlates to fatigue; most notably, the accumulation of H<sup>+</sup> ions at the sarcolemma as lactic acid is produced and disassociated (18,28,31,32). The MF or half-power point of the EMG power density spectrum offers a convenient means of monitoring this change.

The rate of the decrease in the MF provides an index of fatigue for the task performed. This analysis technique has been implemented in a device developed at our Center called the Back Analysis System (BAS). In this device, the subject performs an isometric trunk extension in a restrained position, while spectral parameters of the surface EMG from six muscle sites of the lumbar back are monitored. The current BAS protocol is based on the concept of challenging the endurance of the muscles under study, thus fatiguing them. For the fatigue to appear

over reasonably short contraction times, it is important that the muscles are contracted at relatively high force levels. In the standard assessment protocol, the contraction level is 80 percent of maximal voluntary contraction (MVC). Laboratory and clinical studies by our group (29,30,33-36) as well as others (10,37-39) have documented that fatigue measurements based on the EMG median frequency are highly effective in discriminating LBP subjects from the nonimpaired, as well as in monitoring changes in back muscle function following LBP rehabilitation and exercise (10,11,36,40,41).

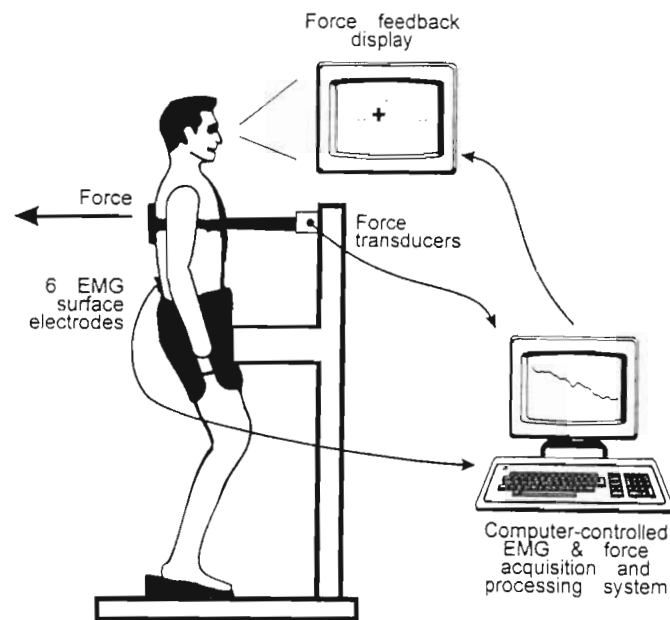
A problem with most back muscle function tests is that they are based on the interpretation of the individual of what his/her maximum performance of strength, speed, or range of motion is. In addition, subjects have to be motivated to reveal their maximum performance. Hence, a highly motivated individual interested in knowing the actual performance state of his/her muscles would perform differently from an individual with less motivation. Such lack of motivation could be due to low pain tolerance in combination with a fear of re-injury when performing the task. Thus, results from such tests will inevitably be contaminated by the conscious and/or unconscious motivation to perform to the full extent of physical capability. Consequently, any assessment test based on the concept of a maximum performance, be it muscular strength, endurance, range of motion, or certain kinematic parameters of voluntary movements, shares the common flaw that it is cognitively perceived by the subject and thus can be voluntarily regulated in a manner that can influence its outcome. This makes these tests unsuitable for subjects who are in acute pain or just unwilling or unable to cooperate during the test. Therefore, it is of paramount importance that the assessment of muscular function in such subjects be based on a test that requires a minimal performance level and that the extracted information cannot be voluntarily regulated by the subject.

The aim of this paper is to investigate the force sensitivity of currently used fatigue EMG parameters as well as to present some of our efforts to develop new surface EMG parameters that may be used in situations when the subject is unable or unwilling to develop the minimal force required to be tested with the fatigue protocol. In addition, the development and initial evaluation of a model to predict MVC from anthropometrical measurements is presented.

## METHODS

### The Back Analysis System (BAS)

Figure 1 shows the BAS in use: the subject is secured by the postural restraint apparatus of the device, which stabilizes the pelvis and lower limbs during the test and is designed to ensure that the sustained isometric muscle activity is actually associated with the extension torque being monitored. Specially contoured, adjustable front and rear restraining molds hold the subject securely in a posterior tilt, and adjustable knee pads provide partial weight bearing and comfort. The generated isometric forces are measured with a nylon harness positioned across the shoulder region of the back and attached to two low-compliance force transducers ( $3.70 \text{ N}/\mu\text{m}$ ). Differences in the forces computed from the two load cells provide a measure of the degree to which the contraction is symmetric or includes a torsional component. The sum of the forces from the load cells are displayed to the subject as a visual feedback "target" to help maintain the desired level of muscle contraction force during the test. Six active surface EMG electrodes (34) are placed bilat-



**Figure 1.** An overview of the BAS showing the different parts of the system. The force feedback display provides the subject with real time information regarding force level and right/left symmetry.

erally over sites at L1, L2, and L5 levels of the lower back corresponding to the longissimus thoraces, iliocostales lumborum, and multifidus muscles. The electrodes have a gain of 10 with a 3 dB bandwidth of 20 to 400 Hz. The EMG signals are further amplified and then processed on-line by customized hardware and software in a personal computer. The amplitude (RMS) and MF of the surface EMG signals are sampled at 10 Hz and all 12 signals, RMS and MF, from each of the 6 EMG electrodes, are stored on hard disk for further off-line processing. A more detailed description of the BAS may be found in Roy SH et al. *Classification of back muscle impairment based on the surface electromyographic signal* immediately preceding the present article.

### Subjects and Protocol

To investigate the relationship between force and EMG-based fatigue parameters, a group of 10 nonimpaired male subjects were tested in the BAS at 20, 30, 40, 50, 60, 70, and 80 percent of MVC. These subjects were selected since they were very familiar with the testing situation and had previously been tested in the BAS several times. The characteristics of the subjects are given in **Table 1**. One of the subjects was left-handed. Following a period of low force level warm-up exercises, the subjects were vigorously encouraged by the experimenter to perform a "true" MVC. They were allowed at least three attempts to perform an MVC. Additional attempts were given if the two best trials differed by more than 5 percent. The best attempt was used as a measure of MVC. The 80 percent MVC contraction was performed first of the test contractions. The remaining trials were performed in a randomized order to cancel out any effects of fatigue across the group means. In addition, a rest period of 3 min was given between consecutive contractions, to ensure that any fatigue effect between the different trials was minimized. Each contraction level was maintained for 30 s.

A post-hoc analysis was performed in a group of eight chronic LBP subjects that had previously been tested at 40 and 80 percent of MVC. These subjects were specifically selected for having pain at the left side of their back during the BAS test; their characteristics are given in **Table 1**. Pain intensity was measured in the LBP subjects with a visual analog scale graded from 0 to 100 (42). Pain location was assessed from a pain drawing on which the subject identified the location of his/her pain as accurately as possible (43).

**Table 1.**  
Anthropometrical data for the different groups of subjects involved in this study.

	Nonimpaired (n=10)	Nonimpaired (n=17)	LBP (n=8)
Age (years)	24.9 ± 5.2	25.7 ± 5.0	40.3 ± 9.6
Weight (kg)	77.7 ± 12.3	76.0 ± 8.3	85.3 ± 9.9
Height (cm)	180.0 ± 7.2	179.6 ± 9.1	1.73 ± 0.10
MVC (N)	1111.2 ± 223.6	1106.9 ± 181.2	680 ± 216
BMI (kg/m <sup>2</sup> )	24.0 ± 3.3	23.6 ± 2.1	28.6 ± 4.1
Months of LBP	—	—	7.8 ± 6.5
VAS Pain Score	—	—	29.4 ± 21.6

To study the relationship between anthropometrical properties and muscle strength, a series of measurements was collected from a group of 17 nonimpaired male subjects. Their isometric trunk extension MVC was measured in the BAS according to the procedure described above. If MVC can be predicted from anthropometrical measurements, this information could be used to determine whether the subject makes a valid effort during the test. If the performed MVC deviates substantially from the predicted value, the subject may not be suitable for a BAS test using the fatigue-based protocol. The degree of deviation in such a prediction needs to be related to the force sensitivity of the fatigue-based EMG parameters (cf. above). We chose to implement a method previously described by McArdle et al. (44), initially developed for estimation of body composition. It has the advantage of being very simple to perform and repeated measurements are highly correlated (>0.95). Twelve anthropometrical measures representing six muscular and six nonmuscular circumferences are taken with a flexible tape measure as the average of two consecutive measurements around the shoulders, chest, biceps, forearm, wrist, waist, umbilicus, hips, thigh, knee, calf, and ankle. The characteristics of the subjects are presented in **Table 1**.

## Data Analysis

### Fatigue Parameters

Linear regression analysis was performed on the MF signal at all force levels for the 10 nonimpaired male subjects that were tested in the BAS. Previous studies have indicated that the MF signal decreases in a linear fashion throughout a 30 s fatiguing contraction (38); however, at the lowest force levels used in this study, there were no electromyographical signs of fatigue as indicated by con-

stant MF values. The data were analyzed between the 3rd and the 30th s to avoid initial variations in the collected data. The slope and the y-intercept of the fitted line was extracted from the analysis for each subject as MF slope and initial MF (IMF), respectively. A second linear regression analysis was performed on the MF slope and IMF values for all subjects across all force levels (%MVC).

The MF slope and IMF parameters for each subject were entered into a previously developed discriminant function (36). This function calculates a standardized score for classification of lumbar back muscle behavior. The function classified subjects into LBP and control groups with 86 percent and 89 percent correct classification on a "learning" and a "hold-out" sample, respectively. The score is calculated so that a negative value indicates nonimpaired behavior and a positive score indicates a behavior consistent with impaired back muscle function. The discriminant function is based on contractions at 80 percent of MVC. To evaluate the sensitivity of the function to variations in MVC, a classification score was calculated for each contraction level (20–80 percent MVC) and a regression line was calculated for each subject (classification score vs. %MVC).

### Imbalance Parameters

A set of parameters was calculated to monitor spectral imbalances between muscles of the lumbar back during different levels of sustained isometric contractions. The parameters were derived by taking a sample-by-sample ratio between contralateral (right and left L1, L2, and L5) as well as ipsilateral (right and left L1/L2, L1/L5, and L2/L5, respectively) lumbar levels of the MF and RMS signals of interest. Thus, these parameters reflect both relative changes in the bandwidth (MF) and power (RMS) of the surface EMG spectrum between the two muscle sites. To correct for the problem that occurs when comparing two ratios where one is larger than 1, and the other smaller than 1, the initial ratios were transformed according to the following procedure. If the direct ratio was smaller than 1, it was first inverted (e.g., 0.5→1/0.5=2), followed by subtraction of 1 and finally a multiplication with  $-1((2-1) \cdot (-1) = -1)$ . If the direct ratio was larger than 1, the value 1 was simply subtracted (e.g., 22-1=21). This gave a set of corrected ratios which were centered around 0. For example, a value of 1 for the corrected ratio of A/B would mean that A was twice the value of B and vice versa for a ratio of -1. This procedure was applied for each

ratio on a sample-by-sample basis through the whole time series. The mean and standard deviation for each ratio between 3 and 30 s was calculated.

#### Anthropometrical Measurements

To verify the validity of the measurements, they were used to estimate the body mass according to the procedure described by McArdle et al. (44). All 12 measurements were then entered as independent variables into a stepwise multiple regression analysis, with MVC as the dependent variable. The analysis was limited to three steps to avoid over-fitting the data.

## RESULTS

### Fatigue Parameters at Different Force Levels

The results from the regression analysis of MF slope and IMF over the different levels of MVC performed on the data for the group of 10 males are presented in **Table 2**. There was a strong and significant correlation between MF slope and %MVC at all electrode sites. MF slope changed significantly faster at the L1 and L5 sites as compared to the L2 site ( $p < 0.01$ , **Table 2**). The linear regression equations were used to calculate at what force level fatigue started to develop as indicated by a negative slope. This occurred at 23, 29, and 20 percent of MVC for the left L1, L2, and L5 sites, respectively. The corresponding values for the right side were 20, 26, and 20

percent of MVC, respectively. The IMF tended to decrease with an increase in force. However, a statistically significant decrease in IMF with increasing force level was found only for the two L5 sites, with a trend for a decrease in the IMF with increasing force for the L1 and L2 sites as well (**Table 2**). The correlation between IMF and force was generally weak ( $-0.14$  to  $-0.38$ ).

### Classification Scores at Different Force Levels

The regression lines between classification score and force level calculated for each of the 10 male subjects are presented in **Figure 2**. Three subjects were correctly classified at 35 percent of MVC. At 58 percent of MVC, 8 of the 10 subjects were correctly classified. This suggests that these subjects would have been correctly classified even if they would have produced just over 70 percent of their true MVC, since the actual test is performed at 80 percent of the measured MVC (80% of 70% = 56%). At 68 percent, all but one subject were correctly classified. This one subject was always misclassified as LBP with the current classification function. It is interesting to note that this subject was a highly trained international endurance athlete (800 m runner) and the only competitive athlete of the group.

### Prediction of MVC and Body Mass from Anthropometrical Measurements

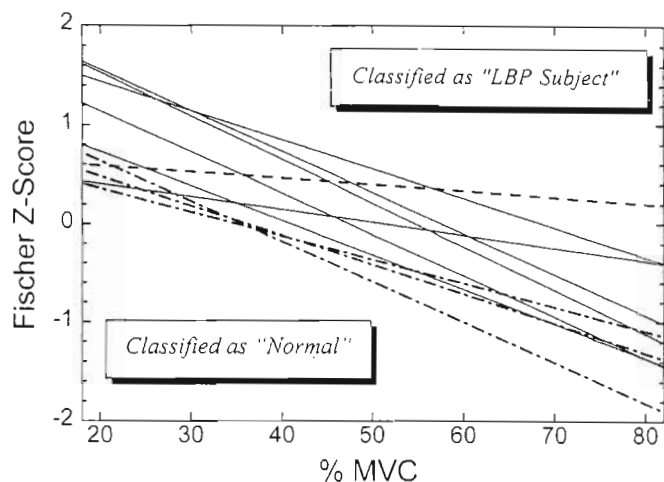
The results from the anthropometrical measurements conducted on 17 nonimpaired young males for the

**Table 2.**

Results from linear regression analysis for MF Slope (Hz/s) against %MVC and IMF (Hz) against %MVC. Significant correlation coefficients are noted in bold.

MF Slope-%MVC	Slope	Intercept	r	p
rL1	-0.015	0.24	<b>-0.75</b>	<0.001
rL2	-0.009	0.23	<b>-0.69</b>	<0.001
rL5	-0.019	0.37	<b>-0.80</b>	<0.001
rR1	-0.018	0.42	<b>-0.75</b>	<0.001
rR2	-0.010	0.29	<b>-0.67</b>	<0.001
rR5	-0.018	0.35	<b>-0.74</b>	<0.001
<b>IMF-%MVC</b>				
rL1	-0.13	101.3	-0.14	0.24
rL2	-0.11	75.2	-0.19	0.12
rL5	-0.27	119.3	<b>-0.38</b>	<0.01
rR1	-0.09	96.6	-0.20	0.097
rR2	-0.22	82.8	-0.23	0.054
rR5	-0.27	119.1	<b>-0.26</b>	<0.05





**Figure 2.** Regression lines between %MVC and classification score for the 10 control subjects. The three dot-dash lines indicate the subjects who were correctly classified already at 35 percent of MVC, and the dashed line indicates the subject who was an elite athlete and never correctly classified with the current classification function.

prediction of MVC are presented in **Table 3**. To verify the validity of the measures, they were used to calculate body mass according to the procedures presented by McArdle et al. (44). There was a strong and highly significant correlation between the measured and calculated body mass ( $r=0.98$ ,  $p<0.001$ ). The largest deviation between estimated and measured body mass for any one subject was 5 percent. All measurements were then entered into a stepwise multiple regression procedure. The analysis was limited to three steps to avoid effects of over-fitting the function. The analysis selected the shoulder, hip, and thigh circumference measurements as the best predictors of MVC in this population of subjects. The correlation coefficient was 0.94 with an adjusted  $R^2=0.86$ . This indicates that 86 percent of the variation in MVC was explained by variation in the three anthropometrical measures. The largest deviation between observed and estimated MVC in any one subject was 10.4 percent. These results suggest that MVC can be accurately predicted from anthropometrical measurements.

### Imbalance Parameters

#### *Nondisabled Subjects*

The surface EMG-based imbalance parameters, defined as ratios between pairs of contralateral and/or ipsilateral MF and RMS signals, displayed some very interesting properties, which contrasted them to the

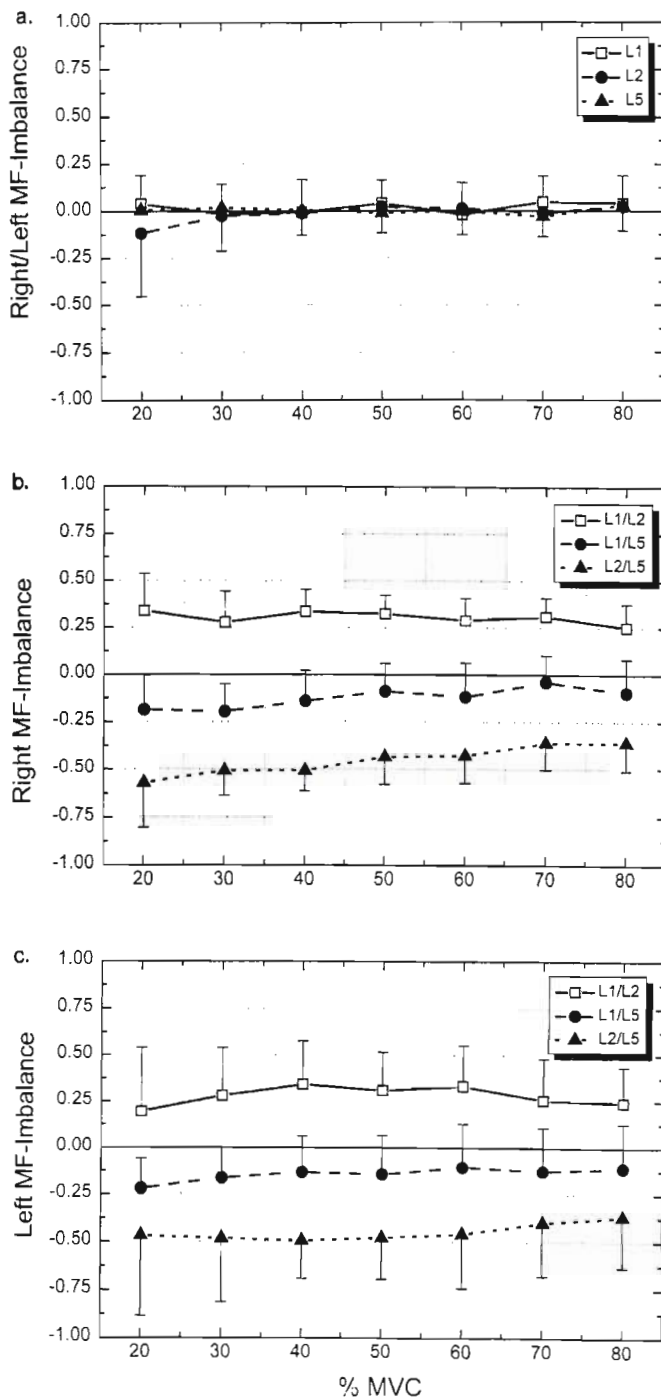
**Table 3.**

Anthropometrical measures used in stepwise multiple regression analysis for prediction of MVC in the group of 17 nonimpaired subjects.

Anthropometrical Measures	Mean (n=17±SD)
Shoulder (cm)	114.1 ± 4.7
Chest (cm)	97.6 ± 5.2
Biceps (cm)	31.8 ± 2.2
Forearm (cm)	27.4 ± 1.3
Wrist (cm)	16.7 ± 0.6
Abds-1 (cm)	79.4 ± 5.3
Abds-2 (cm)	82.1 ± 6.2
Hips (cm)	95.9 ± 5.5
Thigh (cm)	56.1 ± 3.7
Knee (cm)	37.5 ± 1.8
Calf (cm)	37.4 ± 2.0
Ankle (cm)	24.4 ± 2.8
Estimated weight (kg)	76.1 ± 9.1

fatigue-based parameters. The MF-based imbalance parameters for the 10 males are presented in **Figure 3**. The contralateral parameters are shown in the top graph with the ipsilateral left and right parameters in the middle and lower graphs, respectively. The values represent means and standard deviations (SD) for all individuals over all levels of MVC. The overall behavior of these parameters was also consistent within the individual subjects. Note that SDs are plotted in only one direction for better clarity. All MF imbalance parameters displayed a very consistent behavior over the whole force range, especially up to 60 percent of MVC. At the two highest force levels, there was a small deviation for the ipsilateral parameters involving the L2 site, indicating a proportionally smaller decrease in MF associated with fatigue during the contraction at these force levels (**Figure 3** and **Table 2**). This resulted in a significant difference between the right L2/L5 MF imbalance at 20 %MVC versus the 70 and 80 %MVC values, respectively. No other significant differences were found between MF imbalance parameters across different force levels.

A similar behavior was seen for the contralateral RMS-based imbalances (**Figure 4**). All three contralateral RMS imbalances stayed constant over the whole force range. No significant differences were seen between force levels. Interestingly, the L5 values were significantly larger than zero ( $p<0.01$ ), indicating higher RMS on the right than on the left side for the L5 lumbar level



**Figure 3.** Contralateral (a) and ipsilateral (b and c) MF imbalance parameters at different %MVC for the 10 control subjects. Data represent mean and SDs for all subjects over %MVC levels. SDs are sometimes only given in one direction to increase clarity.

(Figure 4). This behavior was most dominant in two of the subjects. A negative imbalance value at the L5 level was unusual in this group of subjects and was seen, inconsistently, in only two of the subjects.

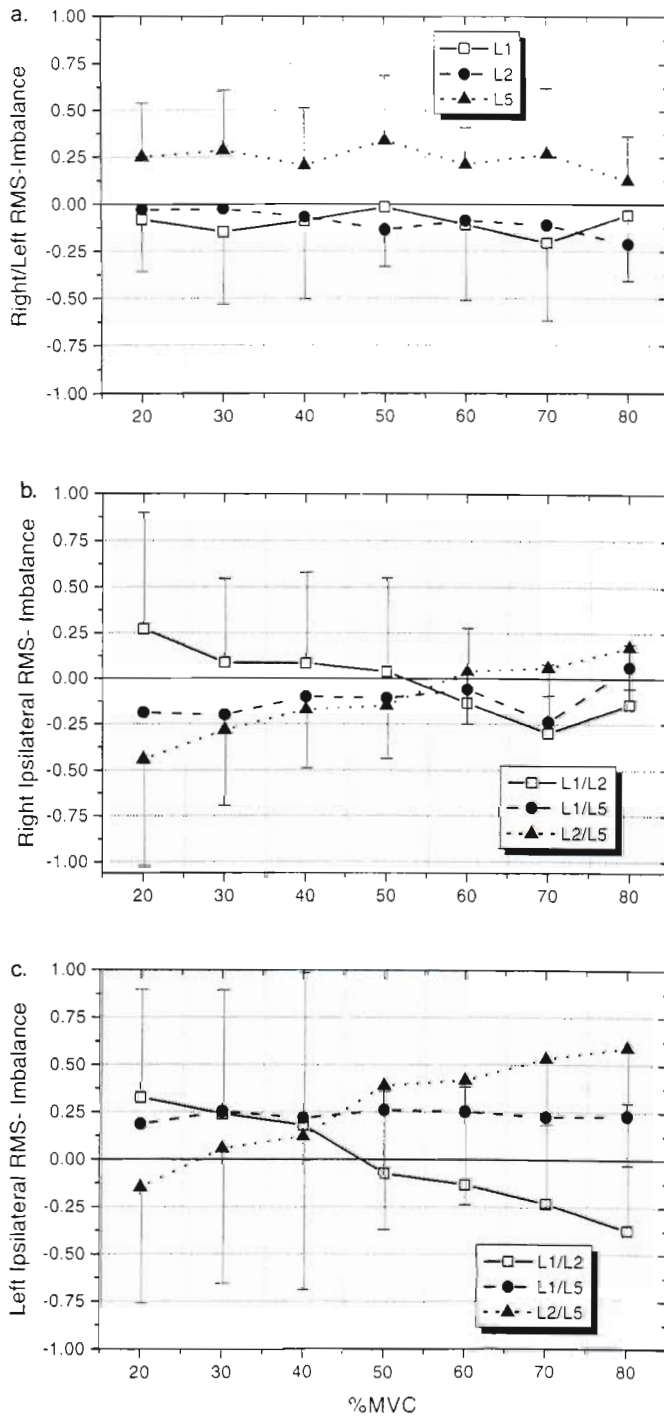
The right and left ipsilateral L1/L5 RMS imbalances were also constant over the whole force range due to similar changes in RMS for the L1 and L5 sites with increasing force. Thus, the L1/L5 RMS imbalance was close to zero for both the left and right sides, respectively (Figure 4). A different behavior was seen for the ipsilateral L1/L2 and L2/L5 RMS imbalances. The RMS increased proportionally more with force at the L2 sites as compared to both the L1 and L5 sites. This resulted in a continuously decreasing L1/L2 ratio and an increasing L2/L5 ratio, respectively (Figure 4). In general, there was a larger variability in the RMS parameters as compared to the MF parameters as indicated by the SDs in Figures 3 and 4, respectively.

#### Subjects with Low Back Pain

Contralateral MF imbalance values for the chronic LBP subjects who had pain on the left side of the lumbar back at the time of testing are presented in Figure 5. Their visual analog pain scale score during activity ranged between 10 and 85. Only the contralateral MF imbalances for L1, L2, and L5 are presented here. The asterisks indicate the five subjects who had a well defined pain site that, according to their pain drawing, happened to coincide with one of the EMG electrode sites. Note that these deviations were always toward a negative imbalance, indicating that the left (in this case the injured side) consistently displayed higher MF frequencies than the right, noninjured side.

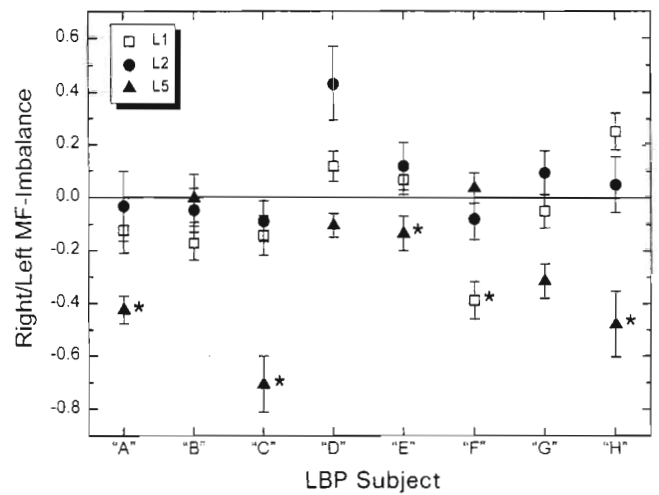
## DISCUSSION

The aim of this paper was to provide an overview of our efforts to develop new surface EMG parameters that can be used for classification of impaired low back muscle function as well as to investigate the force sensitivity of our fatigue-based spectral EMG parameters. We have now demonstrated that subjects must be able to develop a certain minimal force level to be correctly classified in the fatigue-based test protocol. This minimal force level appears to vary somewhat among subjects. Some individuals become correctly classified at very low force levels, while others require higher contraction levels to be correctly classified. This variation may be related to individ-



**Figure 4.**

Contralateral (a) and ipsilateral (b and c) RMS imbalance parameters at different %MVC for the 10 control subjects. Data represent mean and SD for all subjects over %MVC levels. SDs are sometimes only given in one direction to increase clarity.



**Figure 5.**

Contralateral MF imbalance parameters for the eight chronic LBP subjects with pain on the left side of the back at the time of testing. Asterisks indicate subjects who had a pain site that coincided with the electrode site. Note how well the site of pain coincides with an increased negative deviation in contralateral MF-imbalance.

ual differences in conditioning level as well as muscle fiber type composition and muscle fiber size (45). However, it appears from the results of this study that the safety margin for most subjects is rather large. For eight of the subjects in this study, an MVC of just over 70 percent of a "true" MVC would have been high enough for a correct classification. Most of the subjects would even have been correctly classified at 50 percent of a true MVC. We are currently in the process of investigating these findings in a larger population of subjects. These results also indicate that certain categories of subjects, such as extremely conditioned athletes, may never be correctly classified with the currently used discriminant function. The one subject in this study who was never correctly classified differed in two important ways from the other subjects. Firstly, he appeared to develop minimal fatigue, as defined by an MF slope, even at the highest force levels. In addition, his IMF was almost constant over the whole force range, even for the two L5 sites, which parameter showed a significant decrease with increasing force for the group as a whole. It may be argued that the population of fibers this subject recruits at higher force levels, have been conditioned through many years of training to a high lactic acid buffering capacity and/or that he has a very high proportion of fatigue resistant fibers (Type I and/or Type IIa). A specific discriminant function may have to be developed for successful classification of such subjects.



We further demonstrated in this study that the MVC can be accurately predicted from a series of simple anthropometrical measurements. However, the subjects we used represented a rather homogeneous group of young males, so we may not find such a strong relationship in a larger group of subjects with more varying characteristics. On the other hand, in a larger group of subjects more factors could be entered into the analysis, which may compensate for larger variance within certain parameters. Hence, we interpret these results as encouraging. It would also be important to consider nonlinear regression models for the prediction of MVC. The MVC is the measure of a torque produced by the muscles, that is, their force multiplied by length. Since the force produced by the muscles is proportional to their cross-sectional area, the torque would be proportional to the third power of length ( $m^3$ ). The circumference measures used in this study are only proportional to the first power of length ( $m$ ). Thus, adjustments for this effect in the measures used in the regression analysis may further improve the prediction power of the anthropometrical measures.

We plan to use this information to see whether a subject to be tested in the BAS is suitable for classification with the fatigue-based protocol. Thus, in a clinical setting, the anthropometrical measures would be taken from the person before conducting the BAS test. If the subsequently measured MVC is within a certain "safety margin" when compared to the predicted MVC, the subject can be classified from the fatigue-based parameters. It is important to note that this limitation to the fatigue-based protocol only relates to the *classification* of back muscle performance as being nonimpaired or impaired. In a rehabilitation or treatment outcome situation, a subject could still be tested with the fatigue-based protocol in spite of a low contraction level, as long as the comparison is done with previous tests on the same person. In this situation, the subject will act as his/her own control and the parameters from the test would solely be used to monitor treatment outcome. If the subject cannot develop the minimal force required for the fatigue-based test, he or she cannot be classified with these parameters. However, in this study we also presented a set of new parameters, which appear to have properties that make them attractive for classification purposes at minimal contraction levels.

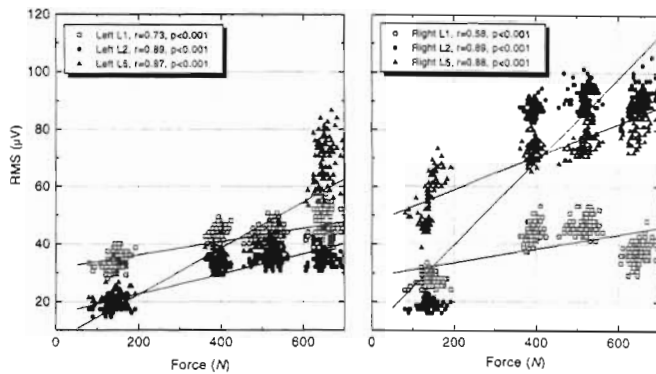
#### *Surface EMG Imbalance Parameters and Aspects of Load Sharing*

These parameters were calculated as ratios between the RMS and MF from pairs of muscles of the lumbar

back. The results have indicated that these parameters were insensitive to the exerted force in a group of control subjects. We further propose that these parameters reflect aspects of "load sharing" between pairs of muscles of the lumbar back. It is well known that the RMS has a close relationship with the force developed by a muscle. However, factors such as temperature, fatigue, subcutaneous tissue, and relationship between active muscle fibers and electrode site/configurations may influence this relationship substantially. Nonetheless, in the experimental setup used here with standardized and symmetrical electrode sites, it is likely that many of these factors are similar for the different muscle sites, especially for the contralateral ratios. Thus, it could be claimed that a ratio between the RMS from two of these muscle sites is a fair reflection of the relationship between the forces developed by underlying muscles.

*In vitro* studies of rat muscles conducted at the NeuroMuscular Research Center have indicated that the MF of the EMG signal contains information regarding the size as well as the type of fibers recruited during a contraction (45). Thus, all else being equal, the MF at a given contraction level would carry information related to the population of fibers being activated, and a ratio between the MF from two similar muscle sites would reflect differences between the population of fibers being activated at the two sites with respect to fiber size and/or type. It is conceivable that when the back muscles contract they share the load in a balanced way, proportional to the force each muscle group can contribute and the function they have during the task. Thus, if a symmetrical force is developed, there should be an equal balance between muscles of the right and the left sides of the spine. Similarly, ipsilateral muscles would need to share the load during a trunk extension exertion, and there is likely a certain optimal load sharing behavior displayed by these muscles for a given level of force exerted. In addition, for gradually increasing force levels there should also be a balanced increase in activation level for all muscles involved in the task.

We have found that nonimpaired subjects usually display a balanced and proportional increase in RMS with force for all muscle sites and for both left and right sides. The increase is close to linear over the force range with correlation coefficients usually higher than 0.90. However, subjects with impairments display a very different type of behavior. An example from such a subject is shown in **Figure 6**. There is a general increase in RMS amplitude with increasing force, but it is different



**Figure 6.**

Relationship between force and RMS amplitude in a person with LBP performing four different contractions. The RMS from the first 10 s of each contraction level are plotted in the graph. Note the imbalanced increase in RMS within as well as between the two sides.

between the muscle sites. For this subject, the L1 sites increased activity between the first and second force level and then appeared to “saturate” (i.e., no further increase was seen in spite of a two-fold increase in force). At the highest force level there was even a *decrease* in activation level at the right L1 site. It is suggested that this imbalance behavior with nonproportional changes in activation level with increasing force production, commonly seen in certain categories of LBP subjects, reflects a characteristic signature of the impairment of the individual person.

#### *Surface EMG and Pain*

The MF imbalance parameters also appeared to reflect changes in the activation of lumbar muscles associated with the presence of pain, as was seen in the five LBP subjects whose pain site coincided with an EMG electrode site. It should be noted that this was a post-hoc analysis of previously collected data. The subjects were selected solely on the basis of their pain drawings and subjective estimation of pain from a visual analog scale score. This should not be misinterpreted to mean that surface EMG signal reflects the sensation of pain. The individual perception of pain is extremely subjective; from a clinical point of view, it is very complex with psychological, physiological, and social components between which it may be very difficult to discriminate.

However, there are specific physiological mechanisms that may explain these findings. A soft tissue injury triggers the release of several endogenous algescic substances. Among them is bradykinin, a very potent activator of group III and IV afferent nerve endings, the

afferents that signal pain to the CNS. When the pain receptors are stimulated, the nerve will release substance P. Substance P stimulates the release of histamine in the tissue that, by acting on the endothelium of the venules, will cause a flow of proteins and white blood cells from the blood vessel to the injured tissue. These typical signs of an acute inflammatory process, which are absent during a chronic inflammation, initiate a healing process in the tissue at the same time as they cause a dramatic change in afferent information from the injured site back to the CNS (46,47).

There is some evidence to suggest that this afferent information influences the activation balance of muscles of the injured area. Such mechanisms, causing alterations in the activity level of injured muscles, are not well understood. However, it is known that reflexively induced muscle inhibition can follow lesions in muscle, joint, ligament, and bone (48–50). Furthermore, afferent information from joint receptors may project to gamma motoneurons, thus causing alterations in muscle tone at and around the injured site (51,52). Brügger (53) suggested that injured muscles become inhibited and muscles protecting the injured area would become hypertonic (54). This is supported by reports that type II fiber atrophy is larger when disuse is caused by pain due to specific inhibition of the more rapid type II motor units still allowing slower type I units to be activated (55,56). Others have suggested that each motoneuron pool is actually supplied by both excitatory and inhibitory nociceptive interneurons (57). This would be an obvious mechanism for the CNS to directly modify the activation balance between muscles around an injury site.

In spite of the speculative nature of the relationship between characteristics of the surface EMG signal and the presence of muscular pain, it is obvious that there is strong support in the literature that the physiological processes associated with pain may have specific effects on the activation balance of muscles at and around an injured site. In chronic LBP subjects, the acute inflammatory reaction is no longer present, so the changes seen in the imbalance parameters cannot be explained by direct stimulation of pain receptors in the previously injured tissue. Instead, the changes we see in this category of subjects may reflect long-term effects of subtle postural adjustments resulting from a strategy to avoid the sensation of pain in the acute phase of the injury. These postural changes may gradually cause voluntary and/or involuntary changes in activation of the muscles at the site of pain, which become established over time as a

“normal” behavior for that individual. This may also explain why persons appear to display specific “signatures” in their activation balance of the lumbar back muscles. We are currently in the process of further investigating the properties of these parameters in a larger population of subjects with pain.

Most persons with LBP will self-recover within 6–8 weeks. However, the approximately 5–10 percent of cases that become chronically disabled account for about 90 percent of the costs (58–60). It is still not known why certain subjects develop a chronic disability. There are some data to suggest that certain behavioral factors may be of importance (61). However, physiological data on this topic are very scarce. To acquire such data there is a need for a good tool to assess precise and objective information regarding the function of the back muscles early in the acute injury phase and to then monitor their behavior through different stages of the injury. We predict that in the future the concept of surface EMG-based imbalance or load-sharing parameters may provide the clinician with important person-specific information already in the acute phase of the injury, to help prevent the development of a chronic disability. Surface EMG provides us with a powerful noninvasive tool to investigate the status and function of muscles. However, we are only at the beginning of understanding the characteristics of the surface EMG signal and its relationship to impairment. With further research using new advanced signal processing methods, we will gradually build up a better understanding of how to interpret the surface EMG signal. The knowledge gained from this process will eventually be used in a clinical environment to benefit the individual persons as better and more cost-effective treatment methods.

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