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Extrapolation of time series of EMG power spectrum parameters in isometric endurance tests of trunk extensor muscles

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Abstract

The aim of the present study was to test the viability of using short isometric contractions of trunk extensor muscles to perform an assessment of their endurance capacity. To this aim two types of analysis were performed. First, electromyographic (EMG) mean power frequency (MPF) slopes with respect to time as estimated over shorter fixed periods were compared to slopes estimated over the full contraction period of a contraction sustained until the endurance time. Second, the relationship between MPF slope estimates as estimated over various periods and the endurance time of the muscle group was evaluated. Five subjects performed three isometric trunk endurance tests at 25%, 50% and 75% of their maximum voluntary contraction (MVC), respectively. EMG signals of the left and right multifidus, iliocostalis and longissimus muscles were continuously recorded and spectral parameters were calculated. The MPF appeared to decrease consistently during all endurance tests. The extrapolation from a MPF time series of half the estimated contraction period to the time series of the complete contraction period gave reasonable results at all force levels, when data from several electrode locations were incorporated in a single slope estimate (mean or steepst slope). The accuracy of the prediction of trunk extensor endurance on the basis of these parameters describing the MPF time series over half the estimated contraction period was satisfactory. Endurance time predictions from yet shorter periods were unreliable. © 1998 Elsevier Science Ltd. All rights reserved.

Keywords: Surface electromyography; Power spectrum; Trunk extensors; Muscle fatigue; Endurance; Low back pain

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1. Introduction

Fatigue of the trunk extensor muscles has been suggested to be an important factor in the etiology of low back pain (LBP) [5,34]. In line with this suggestion, LBP patients have been shown to have a reduced endurance capacity of the trunk extensor muscles as compared to healthy controls [33]. This would imply that endurance tests could be a helpful tool in the diagnosis of functional deficits associated with LBP. However, trunk extensor endurance tests especially in LBP patients may be influenced by factors unrelated to muscle fatigue, like fear, pain and motivation. Therefore, other more objective methods would be preferable above endurance tests.

An alternative method to study muscle fatigability might be the use of electromyography (EMG). During sustained submaximal contractions at levels above about 20% MVC the amplitude of the EMG signal usually increases, while the mean power frequency (MPF) and centre frequency (CF) decrease [12]. Several authors [38,41] have shown that the rate of decline of the spectral parameters during isometric trunk extensor activity is determined by the torque level. This suggests that the rate of decline may be related to the fatigue induced. In an extensive study Hagberg [18] demonstrated a relationship between the rate of change of spectral parameters and the endurance time or limit time (i.e. the time a submaximal contraction can be sustained at the target level) in the biceps brachii muscle. Dieën and co-workers [14] have shown in a limited number of well motivated and pain free subjects that the rate of change of the MPF during isometric trunk extension at moderate torque levels was closely related to the endurance time. Mannion and Dolan [26] obtained similar results for the rate of change of the CF during isometric trunk extensions.

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The viability of EMG based assessments of back muscles has been demonstrated by several authors who used changes in the EMG frequency content to discriminate between LBP patients and healthy subjects [6,13,36,37]. However, in these studies contractions were not sustained until the endurance time was reached, but for an arbitrary prespecified period, representing only a part of the expected endurance time. This approach has the obvious advantage of avoiding the very strenuous latter part of the endurance test. It can be questioned, however, whether such short recording periods allow extrapolation to the EMG changes during a whole endurance test, and more importantly whether they allow prediction of the endurance time.

Kondraske *et al*. [25] compared normalized slopes of the MPF as estimated from a 50% MVC trunk extension until the endurance time with slopes as estimated from the first 20 s of this contraction. They found a moderate correlation between the two estimates ($r = 0.78$). Relationships with endurance time were not reported. A more comprehensive analysis comparing slope estimates obtained from various record lengths and at various contraction intensities has to our knowledge not been reported.

Therefore, the present study was designed to compare MPF slope estimates from record lengths of 30 s, 60 s, half the estimated contraction period, and the full contraction period, obtained during contractions of 25%, 50% and 75% MVC sustained until the endurance time. Furthermore, the relationship of these slope estimates to the endurance time was investigated. It has been shown previously that slope estimates of EMG spectral parameters from single electrode loactions are not sufficiently reproducible for clinical usage [15,30]. In line with this, criteria based on multiple electrode locations allow far better discrimation between LBP patients and healthy subjects [36] and a better prediction of the endurance time [26]. Therefore, the main emphasis in the present study was placed on parameters derived from multiple electrode locations. To this end, the mean of all slope estimates, representing the status of the muscle group as a whole and the steepest slope among the electrode locations, thought to represent the status of the weakest link in the muscle group, were used.

2. Methods

2.1. Subjects

Five healthy male subjects (mean (S.D.) age: 27.2 (5.2) years; height: 186 (4.7) cm; mass: 83 (12) kg) participated in this experiment. Prior to the experiment the subjects were informed about the procedure and possible risks before they gave their informed consent. On the day of the test none of the subjects reported any low back pain.

2.2. Set-up

The experiment was performed with the subjects seated on a Biodex dynamometer (Biodex Corporation, New York, U.S.A.). The torque which the subjects exerted on the axis of the dynamometer was digitized at 50 Hz by a DAS8 AD converter (Keithly Metrabyte, Tauntan MA, U.S.A.) and fed-back to the subjects by means of a monitor that displayed both the target and the actual force.

2.3. Electromyography

Electromyographic signals were obtained from surface recordings using Ag-AgCl electrodes (recessed circular Beckmann type electrodes, interelectrode distance 2 cm, active area 1 cm^2). The EMG signals were recorded from three different parts of both the left and right erector spinae muscles. Electrodes were placed longitudinally over the left and right multifidus, iliocostalis lumborum and longissimus thoracis muscles. Electrode locations for the multifidus and iliocostalis lumborum muscles were at the intersection of the line corresponding to the fibre directions and horizontal lines through the spinous process of L5 and L2, respectively. The muscle fibre directions were determined according to De Foa *et al*. [11] (multifidus muscle: parallel to the line between the posterior superior iliac spine (PSIS) and the L1–L2 interspinous space; iliocostalis muscle: parallel to the line between the caudal tip of the PSIS and the lateral border of the iliocostalis muscle at the 12th rib). The electrode location for the longissimus thoracis muscle was analogous to Roy *et al*. [36] (vertically placed 3 cm lateral to the spinous process of L1). The reference electrode was placed over the C7 spinous process. Prior to the experiment the skin was cleaned with alcohol. EMG signals, which were continuously recorded during the whole experiment, were amplified (Contact Precision Instruments, London, UK), band-pass filtered (1–300 Hz), digitized and stored on a PC at a sample frequency of 1,024 Hz.

2.4. Procedure

The experiment consisted of sustained isometric contractions of the trunk extensor muscles at 25, 50 and 75% MVC sustained until the endurance time. Each subject performed all three tests on three different days, interspersed by at least two days. Before the test the subjects performed a warming-up of 30 isometric contractions at about 100 Nm. After this period they rested for 5 min and then the MVC was determined according to the Caldwell protocol [8], involving three trials of 5 s each with a gradual increase in force during the first 2 s and a plateau-phase of 3 s. This procedure was repated on every test day and the contraction level was set at a percentage of the maximum obtained during that day. After again 5 min rest the endurance test was started. Towards the end of the test the subjects were verbally encouraged to maintain the torque. Whenever the torque dropped below 90% of the target level, the test was stopped and the endurance time was noted.

2.5. Data analysis

The mean power frequency (MPF) of all 2 s episodes of the EMG signals was determined by calculating and averaging the spectra of 80% overlapping Hanning windows of 1,024 samples using a fast Fourier transformation.

Two different types of fits were used to determine the slope of the MPF against time: A linear and a logarithmic fit. The linear fit yielded the best results in most cases, though differences were marginal. Therefore, linear regression analysis was used to determine the slope of the MPF against time. This slope was estimated over four different time periods: (1) over the first 30 s of the test, (2) over the first 60 s (3) over half the estimated contraction period and (4) over the full contraction period. The fit over 60 s was applied to the data of the 25% and 50% MVC tests only, since at 75% MVC 60 s was above the endurance time in some subjects. The endurance time estimate for the third type of fit (half contraction period) was obtained by taking half of the average endurance time of the present subjects.

The four slope estimates were compared and differences were evaluated by means of an ANOVA with repeated measurements. *Post hoc* testing was performed by means of paired *t*-tests. Furthermore, the coefficients of correlation between the 30 s, 60 s, and half contraction period fit on one hand and the full contraction period fit on the other hand were calculated. The relationships between the endurance time and the mean of all slopes across muscles and the steepest slope among the six muscles were studied per contraction level. According to several previous studies [14,18,26] a power fit should be used to describe the relationship of the slope estimates of the MPF with endurance. However, given the small domain over which the fit should be determined in these cases, a linear approximation of the relationship was used. The analysis was repeated on pooled data from the three contraction levels. In this case the strength of the relationship was determined by means of linear regression analysis on logarithmically-transformed data, yielding a power fit.

3. Results

The MVC values averaged across subjects were 295 (S.D. 125), 285 (S.D. 91) and 279 (S.D. 82) Nm and the endurance times were 606 (S.D. 217), 144 (S.D. 24) and 54 (S.D. 10) s, for the 25%, 50% and 75% MVC condition, respectively. No significant differences were seen between the MVC's of the three contraction levels, but the differences in endurance times were highly significant ($P < 0.01$).

Fig. 1 shows the range and mean of the slopes of the MPF time series at each contraction level averaged over muscles. The MPF decreased for all conditions, as was evidenced by the negative slopes estimated over the full contraction period. Note that, in contrast to the other contraction levels, at 75% MVC the fit over half the estimated contraction period (27 s) was made over a shorter period than the fit over 30 s. Fig. 2 gives a typical example of the 30 s and full contraction period regression lines at each of the three contraction levels. Table 1 gives the results of the ANOVA. The MPF slopes appeared to be significantly more steep at higher contraction levels. In addition, the slopes were significantly steeper in the multifidus and longissimus muscles as compared to the iliocostalis muscle. More important, the period over which the slope was estimated affected its mean value significantly. Also the interaction effects of force with the period over which the slope was estimated appeared to be significant. *Post hoc* testing revealed that at 25% MVC the slope estimated over the first 30 s and 60 s of the contraction were significantly less steep than the other two slope estimates. This is clearly illustrated in Fig. 1 and the top window of Fig. 2. At 50% MVC no significant differences were present, whereas at 75% MVC the full period estimate was significantly less steep than the other two estimates (see Fig. 1 and Fig. 2 bottom window).

Table 2 shows the coefficients of correlation between the slopes estimated over the full contraction period and those obtained using shorter fixed periods, i.e. the estimated half contraction period, the first 60 s, and the first 30 s. Figures are given for each muscle separately. In addition, the coefficients of correlation for the mean slopes across muscles and for the steepest slopes among the six muscles are given. In spite of the systematic differences between slopes obtained over different periods as revealed by the ANOVA, in general a clear relationship between the slope estimates was present. Overall the strength of this relationship was quite satisfatcory at 50% MVC. At 75% MVC results were good for some but clearly not for all muscles. At 25% MVC only the estimated half contraction period yielded satisfactory results. In general the results appeared to be better for the multifidus and longissimus thoracis muscles as compared to the iliocostalis muscles, but still better for the parameters derived from multiple electrode locations

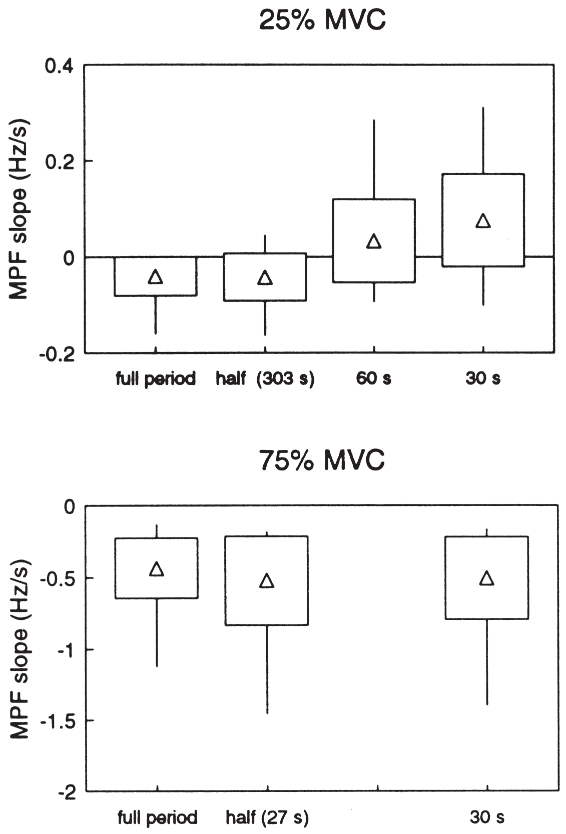

Fig. 1. Slopes of the normalized MPF time series averaged across muscles as estimated over the full endurance time, half endurance time, the first 60, and the first 30 s. The triangles indicate the mean values, the boxes the mean plus and minus 1 S.D. and the vertical bars indicate the range. Note that at 75% MVC half the endurance time is less than 30 s.

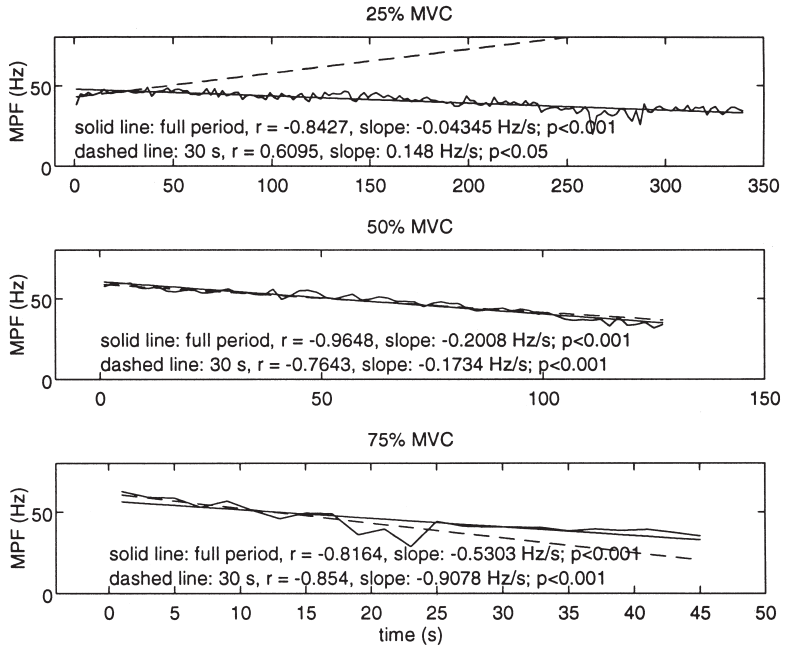

Fig. 2. Typical examples of the regression lines at the three different contraction levels. Dotted line, fitted over the first 30 s; solid line, fitted over the full contraction period.

Table 1
Results of analysis of variance on the slope estimates for the mean power frequency time series

Treatment	*F*	*P*
force	457.54	< 0.001
period	3.40	0.018
left/right	0.44	0.506
muscle	88.33	< 0.001
force * period	3.68	0.003
force * left/right	0.09	0.911
force * muscle	18.80	< 0.001
period * left/right	0.18	0.907
period * muscle	0.18	0.983
left/right * muscle	1.05	0.351
force * period * left/right	0.16	0.976
force * period * muscle	0.10	1
force * left/right * muscle	0.42	0.794
period * left/right * muscle	0.08	0.998

(i.e. the mean and steepest slopes). Fig. 3 gives scatterplots of these slopes obtained over the full contraction period against those obtained over half the estimated contraction period. It can be seen that in spite of the non-significant correlations for the mean slope at 75% MVC and for the steepest slope at 50% MVC quite a consistent trend towards a linear relationship between the slopes is present.

In view of the satisfactory results for the mean and steepest slopes, their relationship to endurance was further analysed by means of regression analysis. Table 3 gives an overview of the results. As can be seen from this table, generally strong relationships between the MPF slopes and the endurance time were found. Relationships were also quite strong for the slopes obtained over fixed periods, with the exception of 60 s and 30 s at 25% MVC. In view of the low scatter of the data at each contraction level, an analysis of the data pooled over the three contraction levels was added. This is indicative of the strength of the relation in a less homogeneous group of subjects. Fig. 4 and Fig. 5 compare the endurance time predictions of full contraction period slopes and estimated half contraction period slopes. As can be seen virtually identical relationships become apparent. Thus clinical tests could just as well employ a contraction for a fixed period of about half the estimated endurance time as a contraction sustained until the endurance time, provided that data from several electrode locations is used. The feasability of classifying subjects into at least a dichotomous distribution of high and low endurance on the basis of the MPF slope estimates is evaluated in Table 4. The top row in each cell gives the rank order of the subjects based on endurance time. The following rows present the rank-ordering based on the MPF slopes. The right hand column indicates whether the rank-ordering is entirely correct (+ +) when compared to the endurance time rank order, whether all subjects are correctly classified as having either below or above average endurance (+) or whether neither of these criteria is met (−). As can be seen the second criterion is met for all full and half contraction period mean slope estimates. In only one case the half contraction period steepest slope estimate did meet neither of the two criteria.

Table 2
Coefficients of correlation between the mean power frequency slope as estimated over the full contraction period and over the shorter fixed periods

		right			left		multiple	
	multifidus	iliocostalis	longissimus	multifidus	iliocostalis	longissimus	steepest	mean
25% MVC								
half (303 s)	0.87	0.90	0.88	0.90	0.97	0.87	0.91	0.90
60 s	NS	NS	NS	NS	NS	NS	NS	NS
30 s	NS	NS	NS	NS	NS	NS	NS	NS
50% MVC								
half (72 s)	NS	0.81	0.94	NS	0.95	0.99	NS	0.95
60 s	NS	NS	0.93	NS	0.90	1.00	NS	0.97
30 s	0.92	0.96	0.89	0.89	0.87	0.99	0.94	1.00
75% MVC								
half (27 s)	0.92	NS	NS	NS	NS	0.92	0.91	NS
30 s	0.94	NS	NS	NS	NS	0.94	0.91	NS

NS, not significant.

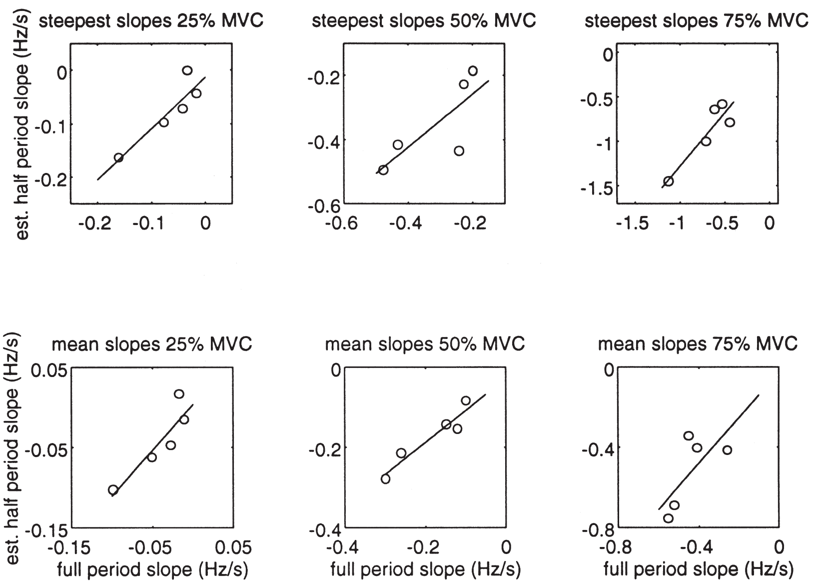

Fig. 3. Scatterplot of the steepest and of the six MPF slopes as obtained from the full contraction period against the steepest and mean of the six slopes as obtained from half the estimated contraction period.

4. Discussion

The aim of the present study was to test the viability of using short isometric contractions of trunk extensor muscles to perform an assessment of their endurance capacity. To this aim two types of analysis were performed. First, EMG mean power frequency slopes with respect to time as estimated over shorter fixed periods were compared to slopes estimated over the full contraction period of a contraction sustained until the endurance time. Second, the relationship between MPF slope estimates as estimated over various periods and the endurance time of the muscle group was evaluated. The second type of analysis can be considered the most directly aimed at the question at hand. However, even in the highly motivated and pain-free group of subjects used in the present study the determination of endurance time is not without problems. It has been argued therefore, that the type of EMG parameters used in the present study may provide a more accurate description of the functional capacity of the muscles [13]. For this reason, the first analysis was added. This analysis is essentially an evaluation of the extent to which the change of the MPF over time can be described by a single (in this case linear) model. Deviations from a time course described by a single model might occur due to changes in the activation level of the muscle or due to the fact that the change in MPF is determined by several physiological processes with different time constants (e.g., changes in intramuscular pH, muscle blood flow, muscle temperature).

Table 3
Coefficients of correlation between the endurance time and the mean power frequency slope as estimated over various periods

		steepest slope	mean slope
25% MVC	full period	0.81	0.83
	half (303 s)	0.85	0.82
	60 s	NS	NS
	30 s	NS	NS
50% MVC	full period	0.94	0.95
	half (72 s)	0.84	0.98
	60 s	0.85	0.98
	30 s	0.95	0.92
75% MVC	full period	0.84	NS
	half (27 s)	0.93	0.95
	30 s	0.90	0.97
pooled data	full period	0.95	0.95
	half	0.86	0.95
	60 s (n = 10)	0.91	0.91
	30 s	0.91	0.86

For the pooled data set the correlation was calculated on logarithmically transformed data. NS, not significant.

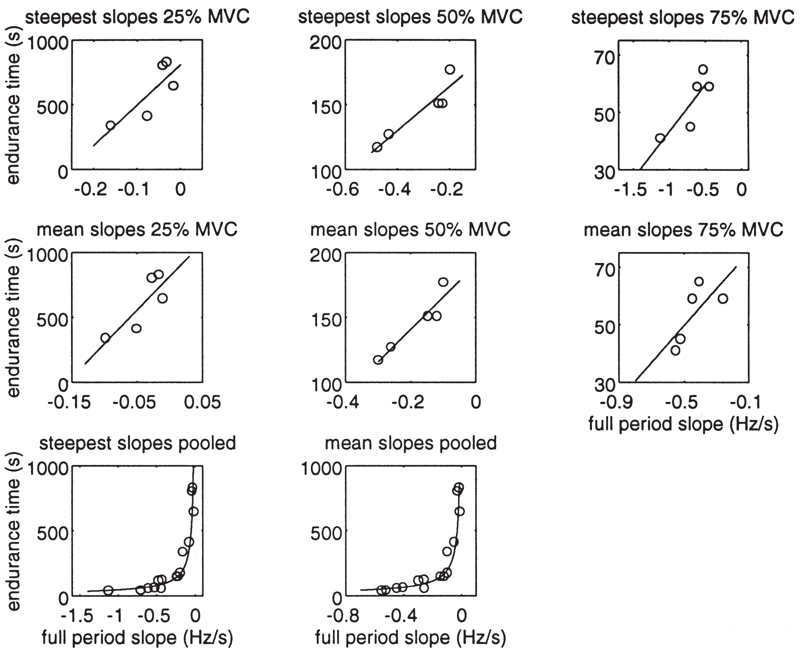

Fig. 4. Scatterplot of the steepest and of the six MPF slopes as obtained from the full contraction period against the endurance time.

The actual time courses of the MPF strongly depended on the force level at which the exercise was performed. As can be seen in Fig. 1 and Fig. 2 the MPF increased during the first part of the 25% MVC endurance test. In contrast the slopes at 75% MVC tended to more negative in the beginning of the contraction. At 50% MVC the MPF did appear to follow a linear course. The behaviour of the MPF time series at 25% MVC might be related to additional recruitment of motor units. Additional recruitment of motor units has been shown to occur throughout sustained contractions at low contraction levels [9,17,28,31,45]. This process appears to follow the recruitment order generally seen in force gradation [10,45]. In the lumbar part of the erector spinae the EMG frequency content increases with forces increasing from 0 to about 40% MVC [27]. This would imply that additional recruitment would initially retard or mask the MPF decline caused by fatigue-related processes at the 25% MVC level.

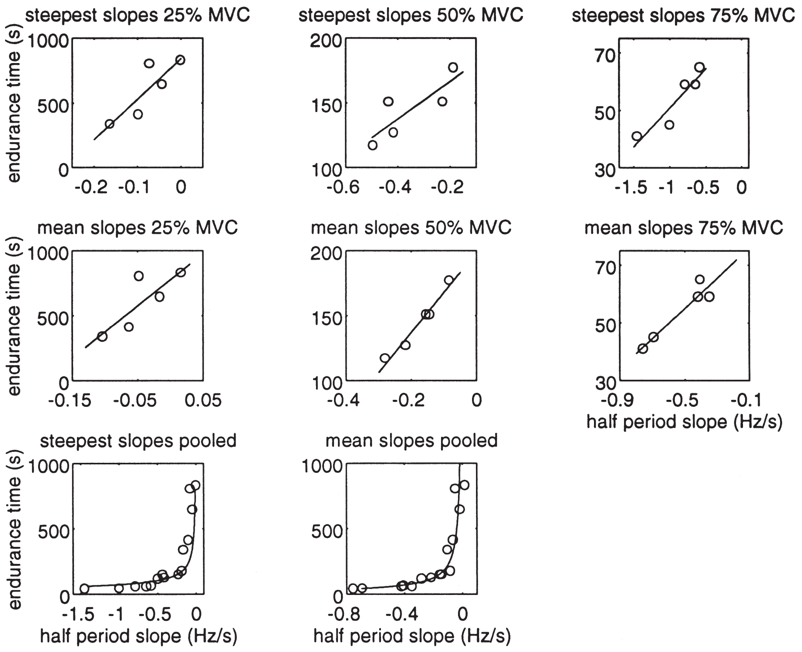

Fig. 5. Scatterplot of the steepest and mean of the six slopes as obtained from half the estimated contraction period against the endurance time.

Table 4
Rank ordering of the subjects on the basis of endurance time and on the basis of the mean power frequency slopes derived over various periods

steepest slopes							
25% MVC		2	5	1	4	3	
	full period	2	5	4	3	1	+
	303 s	2	5	4	1	3	+
	60 s	2	4	5	1	3	–
	30 s	4	5	2	3	1	–
50% MVC		4	2	3	5	1	–
	full period	4	2	3	5	1	+ +
	72 s	4	3	2	5	1	–
	60 s	4	2	3	5	1	+ +
	30 s	2	4	3	5	1	+
75% MVC		4	2	1	5	3	
	full period	4	2	1	3	5	+
	27 s	4	2	5	1	3	+
	30 s	4	2	5	1	3	+
mean slopes							
25% MVC		2	5	1	4	3	
	full period	2	5	4	3	1	+
	303 s	2	5	4	1	3	+
	60 s	4	2	5	1	3	
	30 s	4	5	2	3	1	
50% MVC		4	2	3	5	1	
	full period	4	2	5	3	1	+
	72 s	4	2	3	5	1	+ +
	60 s	4	2	5	3	1	+
	30 s	4	2	5	3	1	+
75% MVC		4	2	1	5	3	
	full period	4	2	1	3	5	+
	27 s	4	2	5	3	1	+
	30 s	4	2	5	3	1	+

+ + indicates that the MPF rank order is identical to the endurance time rank order. + indicates that the MPF correctly identifies subjects as having an above or below average endurance time.

At 75% MVC the MPF declines less rapidly in the second part of the contraction period, leading to a less steep slope of the MPF when estimated over the full contraction period. This result could be explained by the fact that at higher contraction levels, the activity of a part of the back musculature decreases in the latter part of the contraction [7,10]. It has been suggested that this is caused by derecruitment of type II fibres [10]. This would result in a slowing down of the changes in MPF in the latter part of the contraction, as more fatigue resistant fibres remain active. Possibly even some recovery of the MPF may occur, due to the fact that in back muscles type II have a smaller diameter [2,23,35,42,46] and thus a lower action potential conduction velocity [19], as compared to type I fibres.

In a recent study the rate of decline of the CF in the first half of an endurance test at 60% MVC was compared to the total endurance test [16]. In line with our findings at 50% MVC, slopes that described the CF changes during the first half of the endurance test were found not to differ from the slopes obtained over the full endurance time. In a study employing a fatigue test according to Biering-Sorensen [5] a steeper initial slope was found at the lumbar level (L3), while no difference was found at the thoracic level (T10) [26]. In this type of test the torque generated is not standardized at a percentage of the individual MVC. However, in view of the mean endurance time obtained (116 s), the contraction level can be estimated to be about 50% MVC. As mentioned in the introduction, Kondrakse *et al*. [25] at the same percentage of MVC found a moderate correlation between slopes obtained over the full contraction period as compared to slopes over the first 20 s.

From previous studies on moderate isometric trunk extension exercise, it was concluded that the time constant of the spectrum shift of the multifidus and longissimus muscles could be used to predict the endurance time [14,26]. In the present study similar results were found. During moderate isometric activity the mean and steepest MPF slopes as estimated from the full contraction period proved to be closely related to the endurance time. The same held for the slopes as obtained from half the estimated contraction period in this study and a previous study by Mannion and Dolan [26]. The mechanism underlying the MPF changes is most likely the slowing of action potential conduction velocity [1,3,32,38,40]. Action potential conduction velocity is influenced by the accumulation of metabolic products, specifically hydrogen ions in the muscle tissue [24,29] an increase of the cytoplasmic calcium concentration [4], and by the accumulation of potassium outside the muscle fibres [20,24,43]. These physiological changes in the muscle not only influence the conduction velocity but are also important features in the development of muscle fatigue. This could underlie the relationship between the changes of the MPF and fatigue induced.

The results of the present study show that EMG tests over a fixed period can be used to predict back extensor endurance time and hence to assess the functional capacity of the muscle group in this respect. However, the test period should be long enough to allow reliable predictions and hence be adjusted to the contraction level at which the test is performed. A slope estimate of the MPF obtained over half of what can be estimated to be the endurance time at a specific contrraction level appeared to suffice. Shorter tests may yield inaccurate results.

In view of the clinical applicability of EMG tests the question may be posed if the method suffices to discriminate healthy individuals from LBP patients. In a study by Jorgensen and Nicolaisen [22] the mean endurance time at 60% MVC in a patient group was 35 (S.D. 22.8) s and in a control group it was 54 (S.D. 22.1) s. From these figures it is immediately clear that a single endurance estimate can never differentiate at the individual level. The role of endurance or EMG tests would more likely be to assess whether in a LBP patient a deficit in endurance capacity is present. EMG based endurance indicators are probably sufficiently reliable to do so. In the present study a relatively low between-subject variance in endurance time was present, with coefficients of variation of 18, 16 and 35 at 25%, 50% and 75% MVC, respectively. The mean coefficient of variation of several studies in the literature was 38% (range 18–58%) [14–16,21,25]. Nevertheless it proved possible to correctly classify subjects as having a below or above average endurance on the basis of the MPF slopes obtained over the estimated half contraction period, as was shown in Table 4. In a study by Mofforid *et al*. [30] an increase in back muscle endurance time during the Biering-Sörensen test [5] of 22% was found after six weeks of training two times per day. If such an increase in endurance is reflected in the MPF time series, as seems likely on the basis of the close correlations between endurance and the MPF slope estimates, the method would be sensitive enough to detect changes of this magnitude. In our data set the ranges of endurance times at a comparable contraction level was 41% of the mean value. In line with this, Roy *et al*. [39] demonstrated significant changes in EMG changes during isometric contractions after a rehabilitation programme for low back pain patients.

In the results of the present study some differences can be seen between muscles. The frequency content of the iliocostalis muscle appeared to decrease less consistently as compared to the other muscles (e.g., Table 2). In line with this, the slopes of the MPF of this muscle were less steep and less often related to the endurance time (relationships for single muscles were not presented). An explanation for this phenomenon might be the functional difference between the muscles. Several studies on the function of the parts of the erector spinae muscle during isometric extension have demonstrated the medial muscles (multifidus and longissimus) to be more active during submaximal isometric extensions as compared to the lateral muscle (iliocostalis) [7,14,44]. These findings indicate that for the purpose of testing endurance it is probably sufficient to use EMG data from the medial muscle groups. This would mean a reduction of time needed for data acquisition and analysis, which can be important in the clinical context.

5. Conclusion

In conclusion, the method to fit the MPF time series over a recording period representing approximately half the total endurance time seems to be applicable to assess endurance capacity of the trunk extensor muscles. This method seems useful in clinical situations as an alternative for endurance tests, as it minimizes the influence of psychologic and motivational factors on the test results.

References

[1] Arendt-Nielsen L, Mills KR. The relationship between mean power frequency of the EMG spectrum and muscle fibre conduction velocity. Electroencephalogr Clin Neurophysiol 1985;60:130–5.

[2] Bagnall KM, Ford DM, McFadden KD, Greenhill BJ, Raso VJ. The histochemical composition of human vertebral muscle. Spine 1984;9:470–3.

[3] Barnes WS, Williams JH. Effects of ischemia on myo-electric signal characteristics during rest and recovery from static work. Am J Phys Med 1987;66:249–63.

[4] Bianchi CP, Narayan S. Muscle fatigue and the role of transverse tubules. Science 1982;215:295–6.

[5] Biering-Sörensen F. Physical measurements as risk indicators for low back trouble over a one-year period. Spine 1984;9:106–19.

[6] Biedermann HJ. A method for assessing the equivalence of repeated measures of muscle fatigue rate estimated from EMG power spectrum analysis. J Electromyogr Kinesiol 1991;1:288–92.

[7] Bunkens, H. Back muscle fatigue and chronic low back pain. A quantitative analysis of back muscle responses in presons with and without chronic back disorders. PhD thesis. Katholieke Universiteit Leuven, Leuven, 1996.

[8] Caldwell LS, Chaffin DB, Dukes-Dobos FN, Kroemer KHE, Laubach LL, Snook SH, Wasserman DE. A proposed standard procedure for static muscle strength testing. Am Ind Hyg Assoc J 1974;35:201–6.

[9] Christensen H, Fuglsang-Frederiksen A. Quantitative surface EMG dusring sustained and intermittent submaximal contractions. Electroencephalogr Clin Neurophysiol 1988;70:239–47.

[10] Cooper RG, Stokes MJ. Load-induced inflexion of the surface

electromyographic signal during isometric fatiguing activity of normal human paraspinal muscle. Electromyogr Clin Neurophysiol 1994;34:177–84.
[11] De Foa JL, Forrest WJ, Biedermann HJ. Muscle fiber direction of longissimus, iliocostalis and multifidus: landmarks derived from reference lines. J Anat 1989;163:243–7.
[12] De Luca CJ. Myoelectric manifestations of localized fatigue in humans. CRC Crit Rev Bioeng 1984;11:251–79.
[13] De Luca CJ. Use of the surface EMG signal for performance evaluation of back muscles. Muscle Nerve 1993;16:210–6.
[14] Van Dieën JH, Oude Vrielink HHE, Housheer AF, Lötters FBJ, Toussaint HM. Trunk extensor endurance and its relationship to electromyogram parameters. Eur J Appl Physiol 1993;66:388–96.
[15] Van Dieën JH, Heijblom P. Reproducibility of isometric trunk extension torque, trunk extensor endurance and related electromyogram parameters in the context of their clinical applicability. J Orth Res 1996;14:139–43.
[16] Dolan P, Mannion AF, Adams MA. Fatigue of the erector spinae muscles. A quantitative assessment using frequency banding of the surface electromyographic signal. Spine 1995;20:149–59.
[17] Fallentin N, Jørgensen K, Simonsen EB. Motor unit recruitment during prolonged isometric contractions. Eur J Appl Physiol 1993;67:335–41.
[18] Hagberg M. Muscular endurance and surface electromyogram in isometric and dynamic exercise. J Appl Physiol Respir Environ Exerc Physiol 1981;51:1–7.
[19] Håkansson CH. The propagation velocity of isolated frog muscle fibres. Acta Phsyiol Scand 1954;31:22–3.
[20] Jones DA: Muscle fatigue due to changes beyond the neuromuscular junction. In: Human Muscle Fatigue: Physiological Mechanisms. ed by Porter R. Whelan J. Pitman Medical, London: 178-196, 1981.
[21] Jörgensen K, Nicolaisen T. Two methods for determining trunk extensor endurance. A comparative study. Eur J Appl Physiol 1986;55:639–44.
[22] Jörgensen K, Nicolaisen T. Trunk extensor endurance: determination and relation to low-back trouble. Ergonomics 1987;30:259–67.
[23] Jörgensen K, Nicolaisen T, Kato M. Muscle fiber distribution, capillary density, and enzymatic activities in the paravertebral muscles of young men. Significance for isometric endurance. Spine 1993;18:1439–50.
[24] Juel C. Muscle action potential propagation velocity changes during activity. Muscle Nerve 1988;11:714–9.
[25] Kondraske GV, Deivanayagam S, Carmichael T, Mayer TG, Mooney V. Myoelectric spectral analysis and strategies for quantifying muscular fatigue. Arch Phys Med Rehabil 1987;68:103–10.
[26] Mannion AF, Dolan P. Electromyographic median frequency changes during isometric contraction of the back extensors to fatigue. Spine 1994;19(11):1223–9.
[27] Mannion AF and Dolan P: The effects of muscle length and force output on the EMG power spectrum of the erector spinae. *J Electromyogr Kinesiol*. (In press).
[28] Maton B. Human motor unit activity during the onset of muscle fatigue in submaximal isometric isotonic contractions. Eur J Appl Physiol 1981;46:271–81.
[29] Miller RG, Giannini D, Milner-Brown HS, Layzer RB, Koretsky AP, Hooper D, Weiner MW. Effects of fatiguing exercise on high-energy phospates, force and EMG: evidence for three phases of recovery. Muscle Nerve 1987;10:810–21.
[30] Moffroid MT, Haugh LD, Haig AJ, Henry SH, Pope MH. Endurance training of trunk extensor muscles. Phys Ther 1993;73:3–10.
[31] Moritani T, Muro M, Nagata A. Intramuscular and surface electromyogram changes during muscle fatigue. J Appl Physiol 1986;60:1179–85.
[32] Mortimer JT, Magnusson R, Petersén I. Conduction velocity in ischemic muscle: effect on EMG frequency spectrum. Am J Physiol 1970;219:1324–9.
[33] Nicolaisen T, Jörgensen K. Trunk strength, back muscle endurance and low-back trouble. Scand J Rehabil Med 1985;17:121–7.
[34] Parnianpour M, Nordin M, Kahanovitz N, Frankel V. The triaxial coupling of torque generation of trunk muscles during isometric exertions and the effect of fatiguing isoinertial movements on the motor output and movement patterns. Spine 1988;13:982–92.
[35] Polgar J, Johnson MA, Weightman D, Appleton D. Data on fibre size in thirty-six human muscles. J Neurol Sci 1982;19:307–18.
[36] Roy SH, De Luca CJ, Casavant DA. Lumbar muscle fatigue and chronic lower back pain. Spine 1989;14:992–1001.
[37] Roy SH, De Luca CJ, Snyder-Mackler L, Emley MS, Crenshaw RL, Lyons JP. Fatigue, recovery, and low back pain in varsity rowers. Med Sci Sports Exerc 1990;22:463–9.
[38] Roy SH, Schneider J. Effects of electrode location on myoelectric conduction velocity and median frequency estimates. J Appl Physiol 1987;61:495–7.
[39] Roy SH, DeLuca CJ, Emley M, Buijs RJC. Spectral EMG assessment of back muscles in patients with LBP undergoing rehabilitation. Spine 1995;20:38–48.
[40] Sadoyama T, Masuda T, Miyano H. Relationships between muscle fibre conduction velocity and frequency parameters of surface EMG during sustained contraction. Eur J Appl Physiol 1983;51:247–56.
[41] Seidel H, Beyer H, Brauer D. Electromyographic evaluation of back muscle fatigue with repeated sustained contractions of different strengths. Eur J Appl Physiol 1987;56:592–602.
[42] Sirca A, Kostevc V. The fibre type composition of thoracic and lumbar paravertebral muscles in man. J Anat 1985;141:131–7.
[43] Sjögaard G. Water and electrolyte fluxes during exercise and their relation to muscle fatigue. Acta Physiol Scand 1986;128(suppl 556):129–36.
[44] Vink P, Van Der Velde EA, Verbout AJ. A functional subdivision of the lumbar extensor musculature. Recruitment patterns and force-RA-EMG relationships under isometric conditions. Electromyogr Clin Neurophysiol 1988;28:517–25.
[45] Völlestad NK, Vaage O, Hermansen L. Muscle glycogen depletion patterns in type I and subgroups of type II fibres during prolonged severe exercise in man. Acta Physiol Scand 1984;122:433–41.
[46] Zhu X, Parnianpour M, Nordin M, Kahanovitz N. Histochemistry and morphology of erector spinae muscle in lumbar disc herniation. Spine 1989;14:391–7.

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