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The influence of torque and velocity on erector spinae muscle fatigue and its relationship to changes of electromyogram spectrum density

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Abstract The influence of contraction force and velocity during isokinetic contractions on the development of fatigue in the erector spinae muscle was studied. Seven male subjects performed a series of 250 contractions at 25% and 50% of their isometric maximal voluntary contraction (MVC) at 40 and $80^{\circ} \cdot s^{-1}$. Fatigue defined as a decrease of the contractile capacity of the muscles was studied by means of a 15-s maximal test-contraction following the exercise. Both the initial force and the force decrement during the test-contraction were studied. Surface electromyogram (EMG) signals of the main tracts of the erector spinae muscle were recorded. The frequency content was studied by calculating the zero-crossing rate for the signals obtained during dynamic contractions and by means of fast Fourier transformation for the test contraction. After the 50% MVC dynamic contractions the initial force during the postexercise test-contraction was significantly lower than after the 25% MVC contractions. No significant influence of contraction velocity on fatigue development was found. The force decrement during the test-contraction did not depend on the experimental conditions. The EMG amplitude indicated that the subjects were better able to relax their muscles during the counter movement (flexion) at high forces and high velocities compared to the other experimental conditions. The frequency content of the EMG signals during the dynamic contractions and the postexercise test-contraction showed only very weak relationships with fatigue. Therefore, spectrum EMG parameters as determined in the present study do not seem suitable as indicators of muscle fatigue as a consequence of dynamic contractions of trunk extensor muscles.

Key words Localized muscle fatigue · Isokinetic contractions · Electromyography

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Introduction

Muscle fatigue of the trunk extensor muscles has been suggested as one of the factors contributing to the etiology of occupational low back pain (e.g. De Vries 1968). Epidemiological research has shown that repetitive and heavy lifting are associated with the occurrence of low back injuries (Hildebrandt 1987). In lifting activities in the sagittal plane relatively large torques have to be counterbalanced by the erector spinae muscle. Therefore considerable fatigue may develop in this muscle during repetitive lifting. The required muscle forces can be reduced by reducing the loads to be lifted. In practice, however, there has been shown to be a trade-off between the load to be lifted on the one hand and the preferred frequency of lifting on the other hand (Buseck et al. 1988) and thus between on the one hand contraction force, and on the other hand frequency of the contractions and consequently the contraction velocity. The effect of these factors on the development of fatigue of the erector spinae muscle is to a large extent unknown.

Muscle fatigue has been defined as the process leading to a decrease in the performance capacity of the muscle (De Luca 1984), and, in practice, is usually the process leading to a decrease of the force generating capacity of the muscle, as determined from an isometric maximal voluntary contraction (MVC, e.g. Bigland-Ritchie and Woods 1984). In ergonomic practice, however, the measurement of the force generating capacity is often not possible. Therefore, it has been advocated by some to use changes in the electromyogram (EMG) signal, especially of the EMG spectrum density, as an indicator of localized muscle fatigue (Lindström et al. 1977; Kadefors 1978). This approach has been supported by evidence of a close relationship between the rate of change of the EMG spectrum density and the rate of fatigue development in the trunk extensors during isometric exertion (Dieën et al. 1993a, Mannion and Dolan 1994). A similar relationship has been shown to exist in dynamic activities in other muscles (Komi and Tesch 1979; Hagberg 1981). However, it has been pointed out by various authors (Matthijsse et al. 1987; Dieën et al. 1992; 1993b) that the relationship between fatigue development and changes of EMG spectrum density is not consistent among subjects. Furthermore, evident development of fatigue is not in all types of exercise accompanied by changes of the EMG spectrum density (Petrofsky 1979). Clearly, fatigue development and changes of the EMG spectrum density cannot be considered as synonymous, for both processes are multi-facetted and only in part correlated. The validity of using spectrum changes of the EMG to indicate fatigue development, therefore, needs further study.

Changes of EMG spectrum density are most often evaluated by means of the mean power frequency (MPF) or the median frequency as determined by means of fast Fourier transformation (FFT) of the signal. In dynamic activities the location of the muscle fibres with respect to the electrodes, the force, and the muscle length vary, thereby influencing the spectrum density of the recorded EMG (Hagberg and Ericson 1982; Bazzy et al. 1986; Matthijsse et al. 1987; Bilodeau et al. 1990, 1991). Two approaches have been used to avoid these problems in the evaluation of EMG changes during such activities. The EMG can either be sampled during a predetermined part of the range of motion or during isometric test-contractions performed preceding and following the dynamic activity.

Obviously, the first approach can only be used with standardized movements. Furthermore, due to the nonstationary nature of the EMG in dynamic activities, FFT should not be used to evaluate spectrum changes (Bendat and Piersol 1971). The latter problem can be solved by determining the number of zero-crossings of the EMG signal normalized to the signal window length, which is linearly related to the MPF and median frequency and does not require a stationary input signal (Hägg 1981). A drawback of this parameter is its high variability in comparison to the FFT based parameters (Broman et al. 1985). The second method may have a limited validity because the recruitment pattern during the test-contraction may differ from the pattern during the actual activity. In the present study the outcome of both approaches was related to fatigue due to dynamic activity of the trunk extensors to test their validity.

To summarize, in the present study trunk extensor contraction force and velocity in combination with frequency were varied. The aims were to determine the influence of these variations on erector spinae muscle fatigue and to see if a relationship exists between the fatigue development and the outcomes of two ways of quantifying changes of the EMG spectrum density.

Methods

Subjects

Seven male subjects with no history of low back pain and with a normal physical history volunteered to participate in the experiments. Previous to the experiments they were informed of the aims and procedures of the study and signed a statement of informed consent. Their mean age was 24.3 years (range 20–27), mean height was 1.81 m (range 1.73–1.89), and mean body mass was 71.6 kg (range 65–79).

Materials

The experiments were performed on a Kinetic Communicator (Farrel and Richards 1986). With this KinCom dynamometer it was possible to measure and control isometric and isokinetic contractions in a full range of motion. A seat and a brace attached to the arm of the dynamometer, which have been previously described (Dieën et al. 1993a), made it suitable for performing sagittal plane trunk movements.

The subjects were placed in a semi-sitting position (hips flexed 55°, knees flexed 90°), with the pelvis, the lower legs, and knees supported. Straps across the proximal parts of the lower legs, upper legs, and pelvis were used to fix the subject's position in such a way that only the trunk muscles were responsible for the torque exerted. The pivot point of the dynamometer was lined up with the L5/S1 joint in the sagittal plane. The brace was attached to the lever arm of the dynamometer and placed around the thorax of the subject. Direct feedback of the force level was provided on a computer screen. Angle and angular velocity were measured and controlled during the experiments.

Prior to the experiments, after cleaning and gentle abrasion of the skin, surface EMG-electrodes (Medi-Trace ECG, silver-silver chloride) were bilaterally attached over the subject's erector spinae muscle. Electrode locations were marked with ink to ensure exact electrode replacement. Six electrode pairs were used for the three functional parts of the erector spinae muscle on the left and right side (longissimus muscle: 3 cm lateral to L1; iliocostalis lumborum muscle; 6 cm lateral to L2; and multifidus muscle: 3 cm lateral to L5). Inter-electrode distance was approximately 2 cm. The EMG signals were amplified with purpose build pre-amplifiers and telemetrically transported to a Biomes-80 receiver. All signals (EMG, force, angle, angular velocity) were stored on tape (TEAC SR-70, bandwidth 0-625 Hz) and printed using a Gould ES 1000 for an immediate indication of signal quality.

Procedure

The subjects paid a preliminary visit to the laboratory to get used to the equipment and to train in the performance of a symmetric trunk extension at a constant torque.

On 4 separate days the subjects carried out four experimental conditions. Following a standardized warming-up, the MVC was determined in a 40° flexed position of the trunk, using the protocol proposed by Caldwell et al. (1974). The EMG signals were stored to serve as a reference for the signals obtained in the postexercise test-contraction. In each experimental condition the subjects performed 250 extensions of the trunk in a range of 15°–60° trunk flexion. The conditions consisted of combinations of a contraction force of 25%, 50% MVC and a contraction velocity of $40°\cdot s^{-1}$, $80°\cdot s^{-1}$ resulting in contraction frequencies of $19\cdot min^{-1}$ and $38\cdot min^{-1}$ and exercise durations of 13.1 and 6.6 min, respectively. Thus total work differed only between the conditions at different force levels, whereas power output at equal work varied within the force conditions. The order of the experimental conditions was randomized. After each experimental condition, the subjects were allowed a rest period of at least 1 full day. The signals belonging to the first 3 contractions of each series of 25 contractions and of contractions 248–250 were stored on tape, yielding 11 samples per subject for further analysis. Immediately after the 250 fatiguing contractions a sustained maximal contraction was performed for 15 s, the trunk being flexed 40°. Force and EMG signals were stored for the full 15 s.

Data analysis

Force signals were low-pass filtered (cut-off frequency 100 Hz). The EMG-signals were band-pass filtered (high-pass 10 Hz, low-pass 400 Hz). All signals were A-D converted at a frequency of 1000 Hz.

Force data were corrected for the weight of the brace and subsequently converted to torques. The MVC postexercise was determined over a 3-s period starting at the peak of the 15-s force curve (MVC_{post}). The slope of the force decline was determined by linear regression. Thus the force generating capacity was described by its relative decrease due to the exercise and the capacity to sustain a certain force level by the slope of the decline during the test-contraction.

The ability of the subjects to relax their muscles during the counter movement (flexion) was quantified by the root mean square (rms) ratio, following Elert and Gerdle (1989). This parameter is defined as the rms of the EMG signal during negative angular velocity (flexion) divided by the rms of the EMG signal obtained when velocity was positive (extension). The rms-ratio was determined for the 1st, 6th and 11th (final) sample and averaged.

The number of zero-crossings normalized to the number of samples in the EMG signal (ZCR) was determined for all 11 samples taken from the series of dynamic contractions. The slope of the decline of the ZCR over the 250 contractions was determined by linear regression. The EMG signals from the postexercise test-contraction were analysed by means of FFT. Both the median frequency and the MPF were determined over windows of 1024 samples, starting at each full 100 ms of the 15-s contraction. The initial median and mean power frequency (MPF) were determined by averaging the data obtained from the first 3 s.

Statistics

The influence of contraction force and velocity on the force generating capacity was investigated by means of multiple analysis of variance with repeated measures. To see if a relationship existed between the parameters describing the force generating capacity and the EMG parameters at the group level a similar analysis was performed for the EMG parameters. Furthermore, coefficients of correlation between the force parameters and the EMG parameters were determined for each experimental condition. In all tests a significance level of 5% was used.

Results

The pre-exercise MVC ranged from 307 to 369 Nm, these values and the concomitant EMG parameters were used to normalize all results. Median frequency and MPF were overall highly correlated ($r > 0.85$). Therefore, only results concerning MPF are presented.

The torque produced during the dynamic contractions at a target level of 25% MVC appeared to be on average 27.6% at the low angular velocity and 29.6% at the high angular velocity. During the contractions at a 50% target level the mean torques were 46.5% and 46.4%, respectively.

The decrease of the maximal capacity to generate force was reflected by the initial torque during the test-contraction (MVC_{post}). The MVC_{post} appeared to be 77.74% (SD 21.08%) of the pre-exercise value in the low torque, low velocity condition and 73.74% (SD 12.81%) in the low torque, high velocity condition. In the high torque condition it was 66.13% (SD 18.80%) and 64.01% (SD 11.54%) after the low and high velocity contractions, respectively. The effect of the torque of the dynamic contractions was significant ($P < 0.01$), whereas no effect of velocity was found. For the slope of the torque decline during the test-contraction [mean $-1.8\%\cdot s^{-1}$ (SD $0.4\%\cdot s^{-1}$)], reflecting the capacity to sustain the torque, no influence of the factors torque and velocity was found. Therefore, this parameter has not been further analysed.

The rms-ratios averaged over subjects are presented in Fig. 1. A significant influence of torque condition ($P < 0.01$), and velocity condition ($P < 0.05$) is present.

Figs. 2 and 3 give a summary of the effects of the experimental conditions on the two EMG-parameters used as indicators of fatigue. The EMG parameters derived from the test-contraction are based on data from six subjects, since the data of one subject had to be discarded due to technical problems. Hardly any influence of the experimental conditions on the EMG parameters is discernible. Only the slope of the ZCR appeared to depend on the velocity of the contractions ($P < 0.01$). In addition, a significant interaction of torque and velocity on this EMG parameter was present ($P < 0.01$). The MPF differed significantly among muscles ($P < 0.01$), the decrease being greater for the

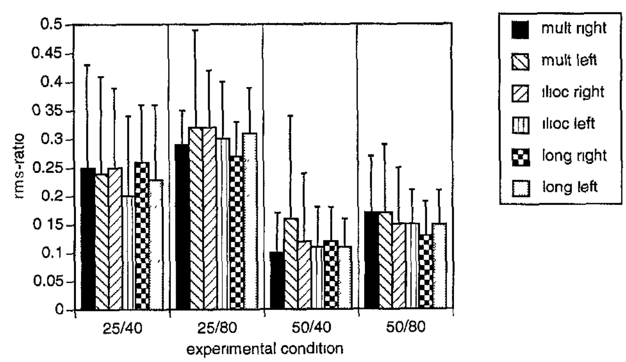

Fig. 1 The rms-ratios in the four experimental conditions. The *error bars* indicate the standard deviation. *Mult* multifidus muscle, *ilioc* iliocostalis lumborum muscle, *long* longissimus muscle

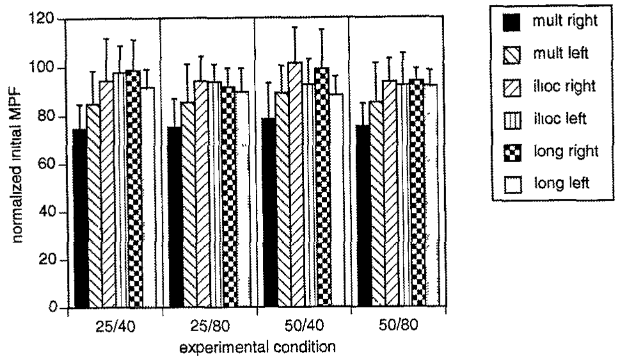

Fig. 2 The initial mean power frequency (*MPF*) obtained during the postexercise test-contraction expressed as a percentage of the *MPF* during the pre-exercise maximal voluntary contraction. The *error bars* indicate the standard deviation. Definitions as for Fig. 1

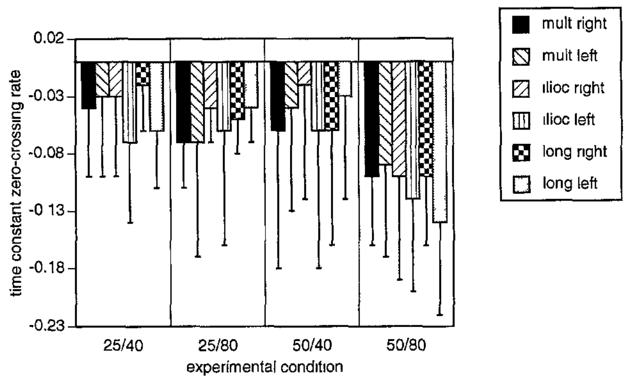

Fig. 3 The time constant of the proportion of zero-crossings during the dynamic contractions. The *error bars* indicate the standard deviation. Definitions as for Fig. 1

multifidus muscle (17%) than for the other two muscles (6% and 8%).

Linear correlations of MVC_{post} with all EMG parameters were calculated. Only the relationship with the MPF of the left and right iliocostalis and longissimus muscles in the low torque, high velocity condition and of the right iliocostalis muscles in the high torque, high velocity condition was significant. In addition, the correlations between the EMG parameters were calculated. No significant relationship appeared to exist between the MPF and the slope of the ZCR.

Discussion

In the present study two indicators of localized muscle fatigue were used. Firstly, a maximal test-contraction was used to assess the capacity of the muscle to generate force. Secondly, the slope of the decline of the torque during this postexercise test-contraction was used to assess the capacity of the muscles to sustain a required force level. The latter parameter is related to the endurance time, which has previously been used as a fatigue indicator by several authors (e.g. Sahlin and Ren 1989). The maximal torque and the endurance are likely to yield independent information as has been demonstrated by the difference in recovery times (Sahlin and Ren 1989). However, related to endurance, the slope of the contraction as used in the present study is not a full alternative for measurement of endurance. The slope of the force can be expected to be related to the maximal force. When the type II fibre pool in the muscle is exhausted during the exercise, the MVC_{post} will be low. In this case the slope would be expected to be low as well, since the torque would now be produced by fatigue resistant type I fibres only. This might explain the absence of any discrimating power of this parameter.

The former force parameter, the MVC_{post}, demonstrated an effect of the dynamic torque on the development of fatigue. No effect on fatigue development of contraction velocity and frequency was found.

The effect of force on fatigue can be masked in part by the difference in relaxation during the counter movement as indicated by the rms-ratios This ratio appeared to be significantly larger in the low torque condition. In a study by Elert and Gerdle (1989) a relationship has been demonstrated between torque decrement and the rms-ratio during maximal torque, high velocity arm anteflexion. Our data revealed the same tendency in the experimental condition most similar to theirs, the high torque, high velocity condition ($r = 0.58$, $P < 0.10$). In contrast to Elert and Gerdle (1989) we did not find any influence of the sequence of the experimental conditions on the rms-ratio. This finding indicates the importance of co-ordination for the fatigability of the trunk extensors as has been previously shown in isometric exercise (Dieën et al. 1993a).

The effect of contraction velocity on fatigue development in skeletal muscle has received surprisingly little attention. In our study the total work was equal within force conditions, whereas the power output varied. Elert and Gerdle (1989) have found a more rapid power decrease during low velocity contractions. However, work per cycle and therefore total work varied between velocity conditions (being higher in the low velocity condition). Jones et al. (1985) have investigated these effects in cycling at maximal power output. In their study, as in the present, total work in both conditions was constant. In contrast with our results, these authors have found a more rapid fatigue development at higher pedalling rates. An explanation for this disparity may be found in the different methods of determining fatigue. The former study has used power output during the dynamic contractions to quantify fatigue. In the present study the torque during an isometric contraction was used. Since it has been shown that recruitment in dynamic contractions may differ from recruitment in isometric contractions (Henneman and Mendel 1981; Kuo and Clamann 1981), isometric testing might be less sensitive to fatigue development as a result of dynamic exercise.

Regarding the dependence of fatigue development on contraction force and velocity it can be concluded that force has a predominant effect. Therefore ergonomic interventions limiting the required forces could be a means of protecting the trunk extensor muscles from over-exertion, even if this results in higher contraction velocities. The present study is however by no means conclusive with respect to this question.

The use of EMG parameters in the present study warrants a discussion of some methodological points. Firstly, due to possible cross-talk, the specificity of the signals obtained for the muscles mentioned in this study cannot be guaranteed. However, using similar electrode locations and inter-electrode distances, Vink et al. (1989) have shown that the signals show only limited cross-talk and are related to functionally different tracts of the erector spinae muscle. So, though it could not be ascertained whether the signals could really be attributed to the anatomical entities multifidus, longissimus, and iliocostalis muscle, each signal did contain meaningful and partially unique information. Secondly, replacement of the electrodes between days seriously limited the reproducibility of EMG parameters. Though much care was taken to replace the electrodes as exactly as possible by means of ink marks, it was decided to use only relative parameters (values normalized to a pre-exercise value obtained on the same day or slopes). Thirdly, an increase of muscle temperature during exercise was likely to have occurred and will have affected the spectrum density of the signals. This increase may have masked fatigue related changes of the EMG. Whether or not this was a systematic effect cannot be ascertained.

In the present study the EMG signals were recorded both during dynamic exercise and during an isometric test-contraction. The spectrum density of the EMG recorded during the dynamic contractions may have been contaminated by shifts of the electrodes with respect to the active muscle fibres and to each other. This effect was avoided by using isometric test contractions. However, during test-contractions the recruited motor unit pool may have differed from that during the actual exercise, probably limiting the sensitivity of this method. Usually test-contractions are performed at a fixed absolute torque level, whereas in the present study the absolute torque pre- and postexercise was different. A disadvantage of the normal approach may be that postexercise the activity of a different pool of motor units is recorded due to additional fatigue related recruitment. By using an MVC both times this is avoided. It does however limit comparability with previous studies.

In the present study, the MPF decreased on average by about 0% to 25%, with considerable variation among muscles and subjects. The ZCR during the dynamic contractions similarly decreased with considerable variations in slope between muscles and subjects. In some muscles in some subjects even an increase of the spectrum density was found. Previous studies dealing with changes of the EMG spectrum density due to dynamic activity likewise have shown inconsistent findings. Petrofsky (1979) has reported both increasing and decreasing median frequencies in quadriceps muscle during cycling, depending on the intensity of the exercise. Komi and Tesch (1979) have found a significant decrease of MPF following 100 maximal concentric contractions only in vastus lateralis muscles with more than 50% fast twitch fibre area. Kuorinka (1988) has reported large differences among subjects in response to dynamic elbow extensions until exhaustion. In some subjects a clear decrease of MPF occurred, whereas in other subjects it remained constant. Only a few studies have dealt specifically with dynamic exercise of the errector spinae muscle. A small (about 5%–10%) decrease of the median frequency EMG during lifting has been found by Petrofsky and Lind (1978). Potvin and Norman (1993) have found decreases of MPF of 20% and 12% in a test-contraction after 20 min of continuous lifting and 120 min of intermittent lifting respectively. The concomitant MVC decrements were 17% and 21%, while the endurance time at 40% MVC decreased by as much as 60% and 62%, respectively. So also in their study the decrease of the force producing capacity was not related to the decrease of MPF.

Komi and Tesch (1979) and Hagberg (1981) have shown that a relationship between fatigue development and spectrum changes of EMG can be found in dynamic exercise of arm and leg muscles. Previously, we have demonstrated such a relationship in isometric trunk extension (Dieën et al. 1993b). Though a weak relationship appeared to exist between the MVC_{post} and the MPF, it is clear from the present study that EMG parameters do not accurately reflect the factors limiting performance in dynamic contractions of the trunk extensors. This could be due to central fatigue, i.e. a performance decrement due to a central nervous factors. However, earlier studies (see Bigland-Ritchie and Woods 1984) have shown fatigue during short-lasting exercise as in the present study (maximum 13 min) to be primarily caused by peripheral factors. Nevertheless, subject motivation may have limited MVC_{post}. In addition, some peripheral changes possibly limiting performance, such as a diminished Ca^{++} release, are not directly reflected in the spectrum density of EMG. This has been corroborated by the complete recovery of MPF soon after concentric exercise, when the force and endurance of the muscle are still affected (Kroon and Naeije 1991). In conclusion, though they probably reflect relevant information with respect to membrane changes coinciding with muscle fatigue, spectrum changes of the EMG of the erector spinae muscle during dynamic activity cannot be considered accurate indicators of fatigue as defined in terms of performance capacity.

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