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Myoelectric manifestations of fatigue at low contraction levels in subjects with and without chronic pain

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Abstract

The aim of the present study was to investigate differences in myoelectric responses to fatigue development between cases with chronic neck-shoulder pain ($n = 10$) and healthy controls ($n = 10$) during a low force level sustained contraction.

Subjects performed a 15-min isometric shoulder elevation at a force level of 40 N (sustained contraction), preceded and followed by a step contraction, consisting of five force levels from 20 to 100 N.

EMG recordings were made with a two-dimensional electrode array on the upper trapezius of the dominant side. Root-mean-square (RMS_G), median power frequency ($FMED_G$), conduction velocity (CV), number of motor unit action potentials per second (MUAP Rate) and MUAP shape properties were estimated. Changes over time and differences between the groups were statistically evaluated with a linear mixed model.

During the sustained contraction, cases showed less increase in RMS_G than controls (controls: 58.5%, cases: 33.0%). $FMED_G$ and CV decreased in controls ($FMED_G$: −6.3%, CV: −5.3%) and stayed constant ($FMED_G$) or slightly increased (CV, 3.15%) in cases. Overall, cases showed a less pronounced myoelectric response to the fatiguing task than controls, which may be related to additional recruitment of higher-threshold MUs. A possible explanation might be that cases were already (chronically) fatigued before the experiment started.

Keywords: Fatigue; Chronic pain; Multi-channel electrode arrays; Electromyography

1. Introduction

Chronic muscular pain is becoming an increasingly important problem in western countries. The Third European Survey on working conditions in acceding and candidate EU countries in 2000 (Paoli and Parent-Thirion, 2003) revealed that 23% of the workers in the EU countries report muscular pain in the neck-shoulder region. Muscular pain is the third important work-related health problem in the EU, only preceded by backache (34%) and stress (28%).

Chronic pain has been studied in relation to muscle activity patterns by surface electromyography (EMG). Several authors reported that EMG activity during work tasks was increased in patients with neck-shoulder complaints (Veiersted et al., 1990; Veiersted, 1994; Lundberg et al., 1999; Madeleine et al., 2003). Other studies demonstrated an inability to relax in trapezius myalgia patients (Elert et al., 1992). Furthermore, a decreased percentage of muscle rest, measured as short silent periods in the EMG ('gaps'), has been demonstrated in chronic pain patients (Veiersted, 1994; Hägg and Åström, 1997). In another

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study, Veiersted showed that lack of gaps was a weak but significant predictor for the development of pain (Veiersted et al., 1993).

Other authors have investigated the development of muscle fatigue in chronic pain patients. Fatigue is known to be reflected in the EMG signal as an increase of its amplitude and a decrease of its characteristic spectral frequencies (De Luca, 1984; Merletti et al., 1990, 2002; Madeleine et al., 2002). In a study on fatigue development in patients with unilateral myalgia was found that the affected side showed less myoelectric signs of fatigue than the healthy side (Öberg et al., 1992). Other studies of fatigue in relation to chronic pain did not reveal differences in EMG parameters, but showed that endurance time was shorter for chronic pain cases (Hagberg and Kvarnström, 1984; Hansson et al., 1992; Larsson et al., 2000).

Pain in the neck-shoulder region relatively often occurs in relation to computer use (Bongers et al., 2002; Jensen, 2003), where the force levels generated by the muscles are very low. Nevertheless, studies of the development of fatigue in pain patients usually are performed at relatively high force levels: for example, Öberg et al. used 0 kg, 1 kg and 2 kg hand loads while subjects held the arms straight at 90° of elevation in the scapular plane (Öberg et al., 1992). It is questionable whether the development of fatigue during such contractions represents the fatigue that occurs during computer work sufficiently. Therefore, in this study we focused on investigating the myoelectric manifestations of fatigue at low force levels.

A hypothesis that is often used to explain chronic pain development is the Cinderella hypothesis (Hägg, 1991). This hypothesis states that low-threshold motor units (MUs) are getting damaged because of too long activation and lack of muscle rest in chronic pain patients (Hägg, 1991). However, the Cinderella hypothesis does not explain how development of damaged MUs proceeds. A possible mechanism could be that healthy MUs first become chronically fatigued, i.e. their force production is reduced, before getting damaged. Assuming this, one would expect that part of the (low-threshold) MUs of cases with chronic pain is already fatigued before the start of a contraction, or that they would get fatigued very early during the contraction. To compensate for the loss of force, higher-threshold MUs would be recruited. This additional recruitment may mask the myoelectric manifestations of fatigue (Houtman et al., 2003). We therefore hypothesize that cases with chronic pain respond less pronounced than healthy controls to a low-level fatiguing contraction in terms of changes in EMG.

To investigate this hypothesis, computer workers with and without chronic pain performed a low force level fatiguing contraction. Multi-channel surface EMG was recorded from the upper trapezius before, during and after the fatiguing contraction. Besides commonly used EMG parameters, conduction velocity (CV) and the number of motor unit action potentials per second (MUAP Rate, MR) were assessed.

2. Methods

2.1. Subjects

Ten healthy subjects (control group, 5 male, 5 female, mean (SD) age 31.0 (11.7) years, weight 69.6 (9.8) kg, height 180 (11) cm, body-mass index (BMI) 21.5 (1.6) kg/m^2) and ten cases with chronic pain (case group, 6 male, 4 female, mean (SD) age 36.7 (9.3) years, weight 70.6 (6.5) kg, height 178 (10), BMI 22.5 (2.8) kg/m^2) took part in this study. The two subject groups were not different with respect to their demographic characteristics (Student's t-test for independent samples, $p > 0.24$). All subjects signed an informed consent. The study was approved by the local medical ethics committee.

A Dutch questionnaire about work and health (Hildebrandt et al., 2001) was used to select subjects. This questionnaire comprises questions about work history and vocational satisfaction, and questions about health, i.e. history, duration and location of complaints and history of therapy.

Subjects were included when they performed predominantly computer work for at least 20 hours per week. Subjects were excluded if they reported non-work related neuromuscular disorders. Subjects were included in the control group when they did not have any self-reported complaints in the neck, shoulders, arms or upper back during the last year. Subjects were included in the case group when they reported more than 30 days during the last year with pain in the neck and/or shoulders. Subjects were excluded if they had complaints in more than three body regions. Cases were recruited from the Dutch society for patients with Repetitive Strain Injury.

2.2. General procedures

Shoulder elevation force and EMG of the m. trapezius were measured simultaneously during two step contractions and an isometric sustained contraction of 15 min at a force level of 40 N (further referred to as sustained contraction). The step contractions were performed before and after the sustained contraction to examine possible changes in motor control. The step contractions consisted of five force levels (20, 40, 60, 80, 100 N) with a duration of 10 seconds each. In between the levels, one second for transition to the next level was given. The first step contraction was followed by 15 min rest (see Fig. 1). After that, the sustained contraction was performed, directly followed by the second step contraction.

When the exerted force tended to decrease, subjects were encouraged to maintain the required force level.

Subjects were seated on an adjustable chair that was high enough to prevent them from touching the floor with their feet. Subjects were instructed not to speak or move the head during the recordings, to sit straight, and to keep their hands rested in their lap. Subjects were not allowed to cross their feet.

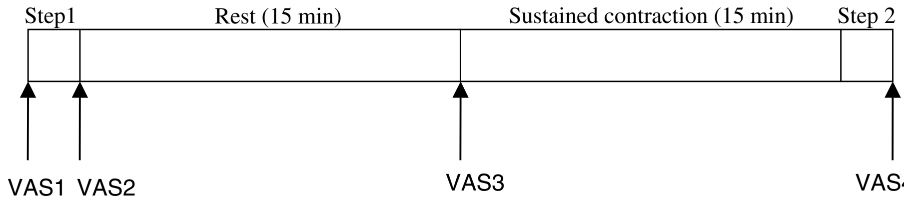

Fig. 1. Experimental protocol. Subjects had to perform a step contraction (levels of 20–100 N in steps of 20 N, duration of each level 10 s), followed by 15 min rest. After that, a sustained contraction with a duration of 15 min at a level of 40 N was performed, immediately followed by a second step contraction. VAS: Visual Analogue Scales of perceived fatigue. Arrows indicate time instances when the VAS scores were recorded.

The chair was attached to a frame that was fixed to the wall. Two force transducers (Thermonobel, Karlskoga, Sweden) were attached to the frame for measuring the shoulder elevation force. The position of the force sensors was adjusted to body size, such that the sensor center was located slightly above the acromion (see Fig. 2). In rest, the force sensors were just not touching the subject. The distance between the spinous process of the seventh cervical vertebra (C7) and the acromion was measured. Force feedback was provided on a laptop screen in front of the subject. The force signals were sampled with 1 kHz, digitized with a 16-bits A/D converter, and stored on a laptop.

Perceived fatigue in the neck-shoulder region was measured with Visual Analogue Scales (VAS, Gift, 1989) before the experiment started, after the first step contraction, after the rest period and after the second step contraction (Fig. 1). The minimal VAS score (0) corresponded to "not fatigued at all", the maximum VAS score (10) corresponded to "as much fatigued as possible".

2.3. EMG recordings

EMG of the dominant upper trapezius was recorded using a two-dimensional 16-channel electrode array developed by the Helmholtz-Institute for Biomedical Engineering, Technical University Aachen, Aachen, Germany (Schulte et al., 2006a). The array consisted of four rows of gold-coated pin electrodes with a diameter of 1.5 mm, the first and fourth containing three contact points and the middle two containing five contact points. The inter-electrode distance (IED) was 10 mm in both directions (see Fig. 3).

Fig. 2. Measurement set-up with the two force transducers located slightly above the acromion at both sides. Force and EMG (not visible) were measured simultaneously from the trapezius muscle. Force feedback was provided on a laptop in front of the subject.

Before placement of the electrode array, the skin was cleaned using abrasive paste. The electrode array was placed with the rows parallel to the line from C7 to the acromion with the centre of the electrode array 2 cm distally from the midpoint, in accordance with the SENIAM recommendations (Hermens et al., 2000). The signals were visually inspected online. Propagation of signals and minimal shape differences between subsequent signals were used as criteria for correct alignment of the electrode array rows in parallel to the muscle fibers. A ground electrode was placed on the wrist of the dominant side.

The monopolar signals were amplified with a gain of 1000 and band-pass filtered (10–500 Hz, Butterworth filter) with a custom made EMG amplifier (Helmholtz-Institute for Biomedical Engineering, Technical University Aachen, Aachen, Germany, input resistance 10^{12} Ω, common mode rejection ratio 78 dB, signal to noise ratio 84 dB). The signals were sampled at 4000 Hz, digitised using a 16 bit A/D-converter (National Instruments) and stored on a laptop.

2.4. Data analysis

For analysis of global EMG parameters, six single differential signals with an IED of 2 cm were constructed offline from the monopolar signals of the middle two rows of the array by subtracting signals from electrodes with 2 cm in between in the direction parallel to the muscle fibers, in accordance with the SENIAM guidelines for conventional surface EMG (Hermens et al., 2000); see Fig. 3. The signals were inspected visually for the presence of artefacts and noise. Epochs containing artefacts were removed and channels with noise were discarded.

Global RMS (RMS_G) and median power frequency ($FMED_G$) were calculated from adjacent, non-overlapping signal epochs of one second for each of the six signals. Since averaging across multiple electrodes increases the stability of the RMS_G estimates (Staudenmann et al., 2005), average values across the six signals were calculated for RMS_G as well as $FMED_G$.

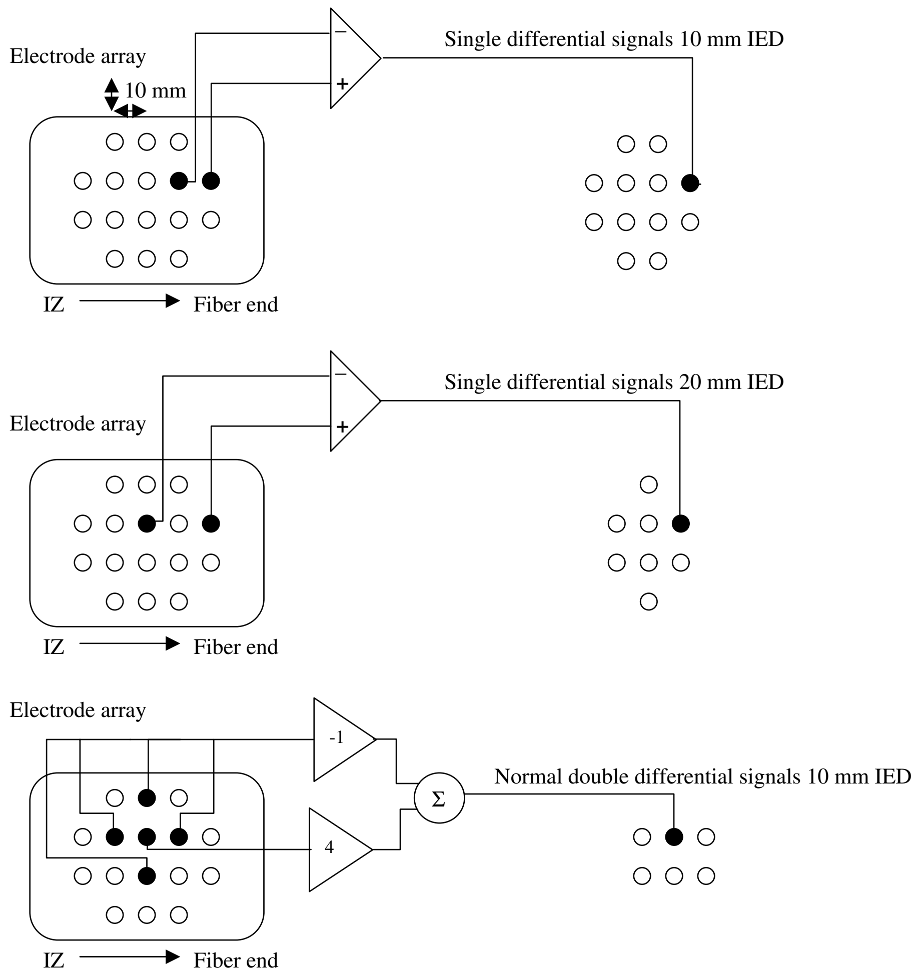

Fig. 3. Schematic representation of the electrode and the construction of several spatially filtered signals from the monopolar recordings.

Estimates of conduction velocity (CV) have been shown to be dependent on the applied spatial filter. Longitudinal (parallel to the muscle fibers) and 2D-filters are supposed to reduce end-of-fiber components (that bias CV estimation) to a larger extent than transverse filters (Schulte et al., 2003). Furthermore, the 2D normal double differential filter (NDD) has been shown to result in a high spatial selectivity in both longitudinal and transverse direction (Farina et al., 2003). Therefore, NDD-filtered signals were used for CV estimation. The number of MUs that contribute to the NDD signals is limited. The CV estimate that results from the interference pattern of the contributing MUs is a weighted average value of the single MU CVs, with weights depending on the location and size of each MU.

Two sets of NDD signals were constructed from the monopolarly recorded signals (see Fig. 3). From each set, the first and last channel was used for CV calculation. CV was estimated using a cross-correlation based algorithm that calculated the time delay corresponding to the maximum of the cross-correlation function, using its time derivative. The two CV estimates were averaged. Furthermore, for each subject at least 50% of the CV values had to be between 2 and 8 m/s; otherwise, the CV data of the subject were excluded from the analysis.

For calculation of MUAP Rate, eight single differential signals with an IED of 10 mm were constructed from the two middle rows of monopolarly recorded signals (see Fig. 3). This resulted in two sets of four unidirectionally propagating single differential signals. For both of these sets, cross-correlation between adjacent signals (in the direction parallel to the fibers) was calculated, resulting in three values from each set. The set with the highest average correlation coefficient was selected for further processing.

MUAPs were detected with a method that uses the Continuous Wavelet Transform to identify shapes that were similar to a mother wavelet (i.e. the first order Hermite-Rodriguez function). The algorithm searched for candidate MUAPs on all channels. A candidate had to occur in at least three channels before being called a MUAP. The outcomes of the detection algorithm were the times of occurrence of the MUAPs detected, and the MUAP shapes on all channels. For more details, see (Gazzoni et al., 2004) and (Farina et al., 2000).

MUAP rate (MR) was calculated for adjacent, non-overlapping epochs of one second throughout the duration of the tasks as the total number of detected MUAPs per second. MR reflects the sum of the firing rates of the active

MUs and was shown to be strongly related to both the number of active MUs and their firing rate, especially for moderate force levels (Kallenberg and Hermens, submitted for publication). For higher force levels, MR is affected by overlappings of MUAPs, that are either detected as single MUAPs or not recognized. The applied force levels in the present study were moderate: the highest force level in the step contraction was 100 N, which corresponds to about 25–30% of maximal voluntary contraction (MVC), that was 357 N for healthy subjects in the same experimental setup (Schulte et al., 2006b).

In addition, the RMS value (RMS_{MUAP}) and median power frequency ($FMED_{MUAP}$) were calculated from the MUAP shapes identified by the algorithm used for MR estimation (Kallenberg and Hermens, 2006). The power density spectrum, from which $FMED_{MUAP}$ was extracted, was calculated using Fast Fourier Transformation and a rectangular window. The extracted MUAP shapes were zero-padded to obtain a frequency resolution of 1 Hz. The MUAP shape parameters RMS_{MUAP} and $FMED_{MUAP}$ were calculated separately from each of the four selected channels, and averaged across the channels afterwards. For each MUAP, RMS_{MUAP} was calculated from a window equal to the duration of the MUAP (on average 11.9 ms), which was determined by the detection algorithm based on the scale coefficient corresponding to the maximum of the scalogram.

For the step contractions, the outcome parameters (RMS_G, $FMED_G$, CV, MR, RMS_{MUAP} and $FMED_{MUAP}$) were averaged for each force level (resulting in five values per subject for each parameter). For the sustained contraction, averages were calculated per minute (resulting in 15 values per subject for each parameter).

2.5. Statistics

The VAS scores were statistically analysed with a two-way ANOVA with factors VAS number (1–4, referring to before and after the first step contraction, after the rest period and after the second step contraction) and group (case or control group). For statistical modeling of the EMG parameters, a mixed linear model was chosen. Mixed linear models are designed to handle correlated data, e.g. including multiple observations of each subject. The model enables the inclusion of two sources of noise: one noise term for each subject, and one noise term for each measurement within a subject. Restricted maximum likelihood estimation was used for extraction of the significant factors. Quantitative changes in the EMG parameters were estimated by applying the mixed model again with only the significant factors included.

The two step contractions were compared with a linear mixed model with fixed factors step contraction number (1 or 2, corresponding to before and after the sustained contraction), level (1–5) and group (case or control group). Level was included as fixed factor because it was restricted to five values (20–100 N). The interaction between group and level was included in the model to examine differences between the groups in response to an increasing force level. The interaction between group and step contraction number was included to examine differences in the effect of the sustained contraction between the two groups. Since it is known that BMI can have an effect on EMG parameters (Nordander et al., 2003), BMI was included as covariate. Furthermore, a random intercept for each subject was taken into account.

To compare the data of the sustained contraction between the two groups, a mixed model with fixed factor group and covariates minute and BMI was used. Minute was included as a covariate since it represents a continuous parameter (time), that was arbitrarily quantified in minutes. The interaction between group and minute was included to examine differences in fatigue development between the two groups.

For both the step contractions and the sustained contraction, the residuals that the model generated were checked for normality with the one-sample Kolmogorov–Smirnov test. The only parameter for which the residual was not normally distributed was $FMED_G$ during the step contractions. Inspection of the data revealed one outlier. After removal of this data point, the residual was normally distributed.

The statistical tests were performed on six parameters, deduced from one EMG. The significance level should be corrected for this. For data with multivariate correlations, the modified Bonferroni correction can be applied (Simes, 1986). The *p*-values corresponding to the six parameters should be ordered such that $p(1) \leqslant p(2) \leqslant \cdots \leqslant p(6)$. The modified Bonferroni correction states that the significance level of the *k*th parameter is equal to $k * 0.05/6$. In the Results section is indicated whether differences were significant or not according to this criterion.

3. Results

The average distance between C7 and the acromion was 22.6, SD 1.62 cm.

The VAS scores of perceived fatigue are reported in Table 1. The VAS scores in the control group before the sustained contraction were low. The VAS scores after the sustained contraction are moderate in the control group and high in the case group. VAS scores were on average about two points higher in the case group than in the con-

Table 1
Visual analogue scales (mean and SD) of perceived fatigue

	Controls	Cases
VAS1	0.24 (0.28)	1.24 (1.77)
VAS2	0.95 (0.94)	3.39 (2.20)
VAS3	0.34 (0.40)	2.13 (2.13)
VAS4	4.32 (2.09)	6.17 (2.20)

VAS1 and VAS2 were measured before and after the first step contraction, VAS3 was measured before the sustained contraction and VAS4 was measured after the second step contraction.

trol group, before the first contraction. An ANOVA with factors group (control or case) and VAS number (1–4) revealed a significant dependency on both factors ($p < 0.01$).

In Fig. 4, the mean force curves for both groups are shown. Linear regression analysis resulted in a slope of 0.020 N per minute in the control group (not significantly different from zero, $p > 0.18$) and -0.14 N per minute (significantly different from zero, $p < 0.001$) in the case group. Inspection of individual force curves revealed that the slight decrease in the case group is due to one subject that was not able to maintain the force level.

In Fig. 5, RMS_G during both step contractions and the sustained contraction is shown. During the step contractions, RMS_G increased significantly with level ($p < 0.001$, $F(4,154) = 24.18$). RMS_G was significantly higher for the second than for the first step contraction ($p < 0.001$, $F(1,154) = 101.8$). No interaction between step contraction number and group was found.

During the sustained contraction, RMS_G increased significantly with time ($p < 0.001$, $F(1,257) = 207.4$) and there was a significant interaction between group and time ($p < 0.001$, $F(1,257) = 15.5$) with a stronger increase in the control group (control group: 3.9% per minute, case group: 2.2% per minute). RMS_{MUAP} showed a similar behaviour. The correlation between RMS_{MUAP} and RMS_G was very high (step contractions: $R^2 = 0.91$, sustained contraction $R^2 = 0.90$).

In Fig. 6, $FMED_G$ during both step contractions and the sustained contraction is shown. For the step contractions, there was a significant interaction between group and step contraction number ($p < 0.001$, $F(1,155) = 11.86$) with the case group showing an increase (4.7%) and the control group a decrease (-1.7%) from the first to the second step contraction. There was a trend for an increase of $FMED_G$ with level ($p < 0.08$, corrected significance level 0.04, $F(4,154) = 2.13$). There was a significant dependence of $FMED_G$ on BMI (2.0% per kg/m^2). For the sustained contraction, there was a significant dependence of $FMED_G$ on time ($p < 0.001$, $F(1,257) = 11.19$) and a significant interaction between group and time ($p < 0.001$, $F(1,257) = 21.31$). $FMED_G$ decreased with time in the control group and stayed constant in the case group (control group: -0.42% per minute, case group: 0.069% per minute, not significantly different from zero). For $FMED_{MUAP}$, results were similar. The correlation between $FMED_G$ and $FMED_{MUAP}$ was moderate (step contractions: $R^2 = 0.66$, sustained contraction: $R^2 = 0.72$).

In Fig. 7, CV during both step contractions and the sustained contraction is shown. For the step contractions, two subjects showed extreme values (higher or lower than the mean value ± three times the interquartile range). For the sustained contraction, three subjects showed extreme values and for two subjects more than 50% of the CV values were lower than 2 m/s. Data of these subjects were excluded from the analysis. CV was significantly higher during the second

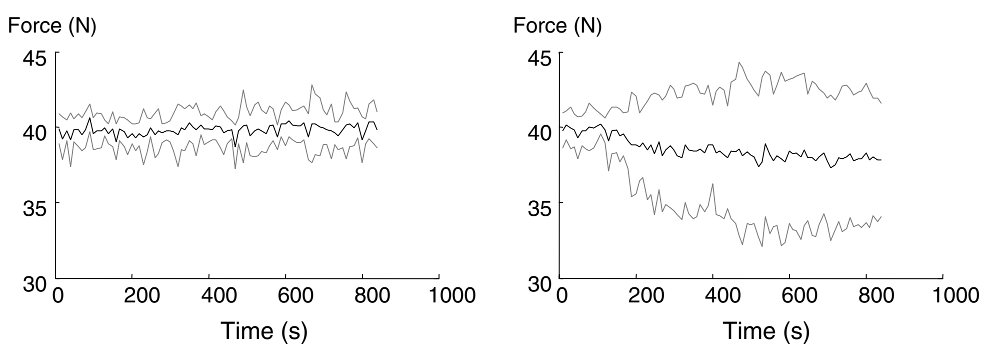

Fig. 4. Average force curves during the sustained contraction (force values were averaged across ten second periods and across subjects). Right side: control group, left side: case group. Black lines: group mean, grey lines: standard deviation.

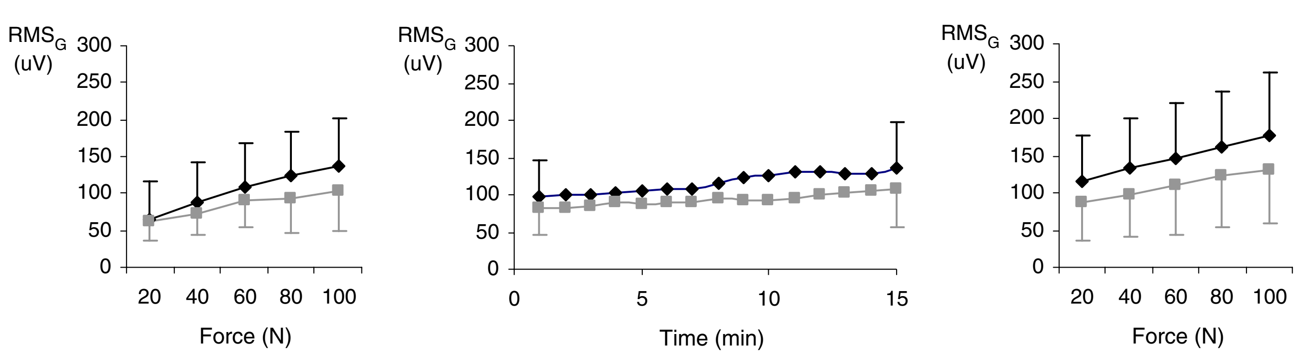

Fig. 5. RMS_G during both step contractions (left and right) and sustained contraction (middle). Black diamonds: controls ($N = 10$), grey squares: cases ($N = 10$). Note that the x-axis is different for the step contractions and the sustained contraction. Bars indicate standard deviations.

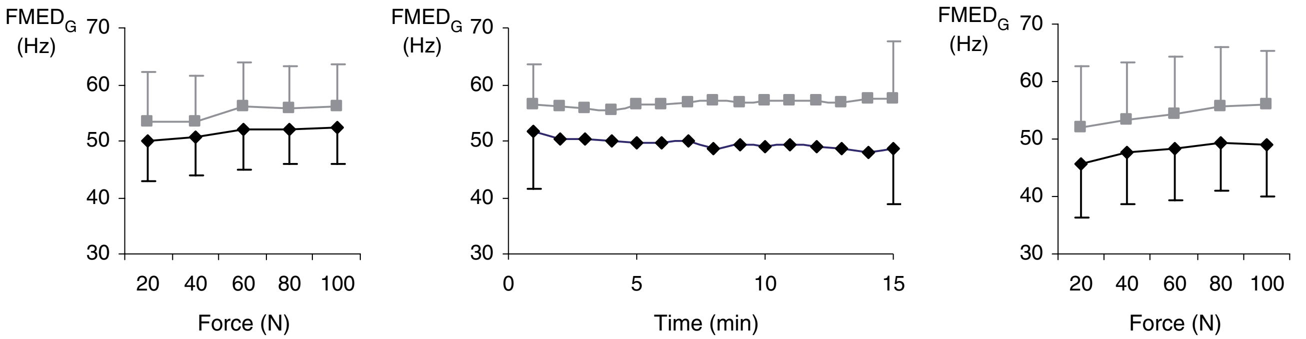

Fig. 6. $FMED_G$ during both step contractions (left and right) and sustained contraction (middle). Black diamonds: controls ($N = 10$), grey squares: cases ($N = 10$). Note that the x-axis is different for the step contractions and the sustained contraction. Bars indicate standard deviations.

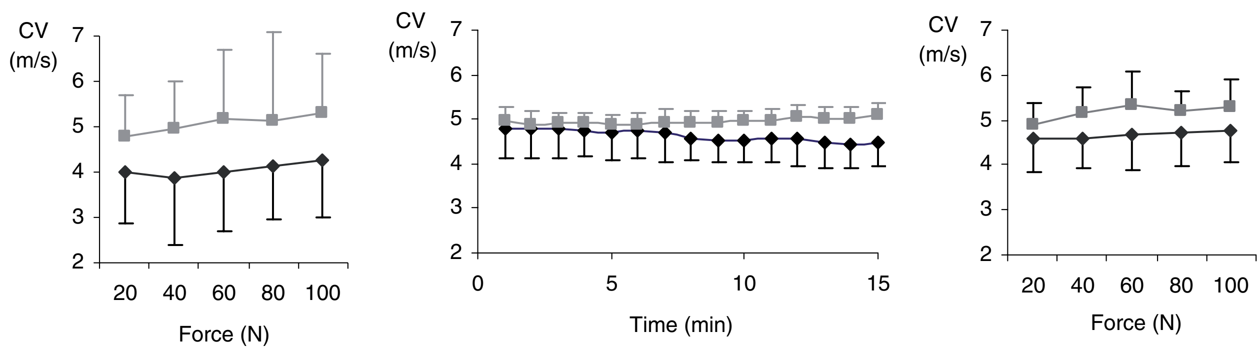

Fig. 7. CV during both step contractions (left and right) and sustained contraction (middle). Black diamonds: controls ($N = 8$), grey squares: cases ($N = 8$). Note that the x-axis is different for the step contractions and the sustained contraction. Bars indicate standard deviations.

step contraction than during the first step contraction ($p < 0.001$, $F(1,120) = 31.96$). No interaction between step contraction number and group was found. There was a dependence on BMI ($p < 0.001$, $F(1,27) = 18.7$). A trend for a dependence of CV on level was found ($p < 0.10$, corrected level 0.05, $F(4,115) = 1.99$). During the sustained contraction, there was a significant interaction between group and time ($p < 0.001$, $F(1,216) = 47.5$). CV decreased in controls and increased slightly in cases (control group: −0.35% per minute, case group: 0.21% per minute).

In Fig. 8, MR during both step contractions and the sustained contraction is shown. For the step contractions, MR increased significantly with level ($p < 0.001$, $F(4,153) = 86.5$). There was a significant relation between MR and step contraction number ($p < 0.026$, corrected significance level 0.033, $F(1,155) = 5.04$) and a trend for dependence on the interaction of group and step contraction number ($p < 0.039$, corrected significance level 0.025, $F(1,155)=4.33$). MR increased in the control group (12.6%) while in the case group it stayed constant (0.27%, not significantly different from zero) from the first to the second step contraction. Fig. 8 shows that this is largely caused by differences at the lower force levels. A significant negative association of MR with BMI was found ($p < 0.014$, corrected significance level 0.025, $F(1,23) = 7.15$).

For the sustained contraction, a significant negative association with BMI was found (2.9% per kg/m^2, $p < 0.002$, corrected significance level 0.008, $F(1,15) = 13.3$).

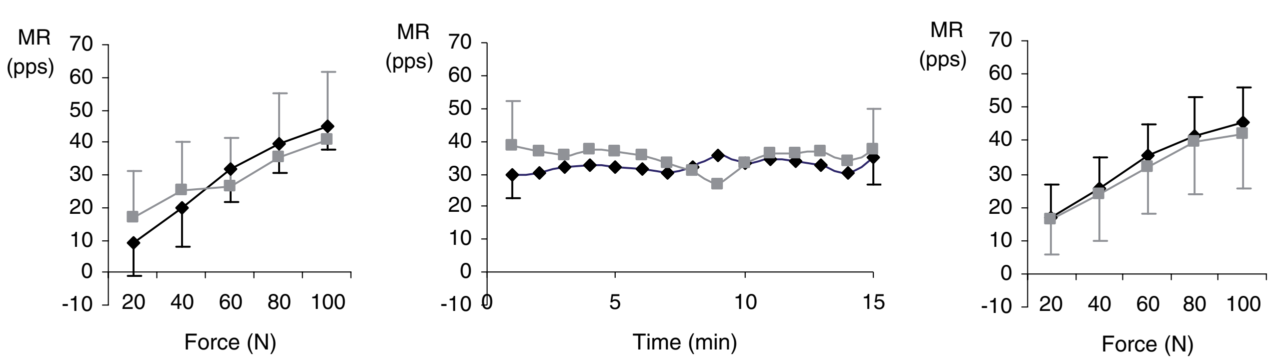

Fig. 8. MR during both step contractions (left and right) and sustained contraction (middle). Black diamonds: controls ($N = 10$), grey squares: cases ($N = 10$). Note that the x-axis is different for the step contractions and the sustained contraction. Bars indicate standard deviations.

Furthermore, there was a trend for interaction between time and group ($p < 0.06$, $F(1,244) = 3.54$, corrected significance level 0.05), with increasing MR in the control group and decreasing MR in the case group (control group: 0.40% per minute, case group: −0.25% per minute).

4. Discussion

The objective of this study was to investigate differences in myoelectric manifestations of fatigue development at low force levels between healthy subjects (controls) and subjects with chronic neck-shoulder pain (cases). It was hypothesized that part of the MUs would already be chronically fatigued before the start of the experiment in the case group. The early decrease in force production of these MUs would be compensated for by recruitment of higher-threshold MUs, which in turn would result in a less pronounced response in terms of myoelectric changes to a low-level fatiguing contraction in the case group. To generate fatigue, subjects were asked to perform a 15 min sustained contraction at a force level of 40 N.

Myoelectric signs of fatigue consist of an increase of RMS_G and a decrease of global CV (conventionally measured with two surface electrodes) and $FMED_G$ (Merletti et al., 1990, 2002; Madeleine et al., 2002). For a single MU, a decrease in muscle fiber CV induces an increase in MUAP duration (Lindstrom and Magnusson, 1977) and a decrease in $FMED_{MUAP}$ (Seki and Narusawa, 1998). Fatigue also leads to additional MU recruitment (Olsen et al., 2001; Jensen et al., 2000; Enoka et al., 1989; Maton and Gamet, 1989; Moritani et al., 1986), which would induce an increase in MR with fatigue. Furthermore, according to the size principle (Henneman et al., 1965), additionally recruited MUs are higher-threshold MUs with in general a higher muscle fiber CV (Andreassen and Arendt-Nielsen, 1987) and a larger size. This would result in an increase in RMS_{MUAP} (Roeleveld et al., 1997).

Several results indicate that the sustained contraction was indeed fatiguing. Firstly, the VAS scores after the sustained contraction were much higher than the VAS scores before. Secondly, in the control group myoelectric manifestations of fatigue were found: during the sustained contraction, RMS_G, RMS_{MUAP} and MR increased significantly, while CV and $FMED_G$ decreased. It was hypothesized that cases would show less myoelectric signs of fatigue than controls. Indeed, the results indicate that cases do not respond as much as controls to a fatiguing task. During the sustained contraction, myoelectric manifestations of fatigue were smaller in the case group. Cases showed a less steep increase in RMS_G and RMS_{MUAP}. Additionally, controls showed a decrease in $FMED_G$, $FMED_{MUAP}$ and CV whereas in the case group these variables stayed constant or increased slightly with time. In addition, there was a trend for a difference in MR slope: in the control group MR increased with time and in the case group MR decreased with time. Further evidence to support the hypothesis that cases show less signs of fatigue is demonstrated when comparing the two step contractions. Similar to the results of the sustained contraction, controls showed a decrease in $FMED_G$ and $FMED_{MUAP}$ from the first to the second step contraction, whereas these variables increased in cases.

These results are in line with the hypothesis that part of the MUs is chronically fatigued in the case group. The higher VAS scores in the case group, already before the contractions started are in agreement with the hypothesis as well. Furthermore, the VAS scores did not return to baseline during a 15 min rest period in the case group, in contrast to the control group.

A confounding factor might be the slight decrease in force level in the case group, although visual inspection of individual data revealed that this is due to only one subject. Another confounding factor might be the development of pain during the contraction. This might cause a decrease of the activation level of the trapezius that is compensated for by synergistic muscles. However, no evidence for such a mechanism was found; global RMS did not decrease in the case group.

The finding that cases show less signs of fatigue is in line with a study that compared secretaries with work-related bilateral chronic muscle pain with healthy controls (Schulte et al., 2006a). In this study, changes in CV during an isometric endurance task were more pronounced in the control group than in the case group. The present results are also in line with previous findings reported by Öberg and colleagues (Öberg et al., 1992). They demonstrated a smaller increase in RMS and a smaller decrease in mean power frequency on the affected side than on the healthy side in cases with unilateral myalgia. A lower response to a fatiguing task in cases has been demonstrated as well for cases with low back pain (Oddsson and De Luca, 2003; Lariviere et al., 2003).

Underlying the hypothesis that cases would show less signs of fatigue than controls, it was assumed that the painful muscles of cases are chronically fatigued. This would imply that peripheral muscle fiber properties (e.g. muscle fiber CV, cell membrane properties etc.) are changed in cases. This should then be reflected in the EMG parameters already at the beginning of the tasks. For the case group, MR seems to be somewhat higher for the low force levels of the first step contraction (20 and 40 N). Although not significant, this increase might give some indication that the force output of the low-threshold MUs is decreased in cases. To investigate this further, a study with a larger sample size should be performed.

During the sustained contraction, there was a significant but slight increase of CV in the case group. Since CV of higher threshold MUs is generally higher (Masuda and De Luca, 1991; Andreassen and Arendt-Nielsen, 1987; Hogrel, 2003), this may indicate additional recruitment of higher-threshold MUs. Such additional recruitment would lead to an increase in MR. However, the results also show that MR does not increase in the case group, which seems to be in contrast. A possible explanation is that part of the low-threshold MUs are de-recruited due to fatigue. To compensate for this, less high-threshold MUs are needed because of their larger force output (Feiereisen et al., 1997), which results in a lower MR.

In the control group, $FMED_G$ decreased from the first to the second step contraction, whereas CV did not. Although the frequency content of the signal is largely determined by CV (Lindstrom and Magnusson, 1977; Dumitru et al., 1999) Merletti et al. (1990) showed that changes in spectral variables due to fatigue are larger than those in CV. Furthermore, they concluded that CV is not the only physiological factor affecting spectral variables. Another study showed that $FMED_G$ is affected by shape and duration of the MUAPs (Hermens et al., 1992).

In the present study, the placement of the force sensors was adjusted to body size, such that the sensors were located slightly above the acromion. Thus, the subjects applied the force in the same way; i.e. the contact with the sensor occurred at a bony point. Since the required force was constant, differences in the distance between C7 and acromion might lead to differences in applied torque. However, since the differences in distance between C7 and acromion were rather small (SD was 7% of their mean value) the effect on the results can be expected to be limited.

Many authors have used relative force levels, related to MVC for investigating the relation between EMG parameters and force (e.g. Conwit et al., 1999; Bilodeau et al., 2003; Queisser et al., 1994). One of the reasons for using relative force levels is that inter-subject variability is assumed to be decreased. We used absolute force levels (in N) rather than relative force levels for several reasons. First, since MVC is probably different between the two groups, relative force levels might mask the differences between the groups. Second, the assessment of MVC is difficult, especially in people with pain or fear of pain, since the measured MVC might not be the maximal force, but the force that is acceptable in terms of pain. Third, the force levels that are needed in daily life activities are also absolute levels independent of a subject's muscle capacity. Schulte et al. (2006b) reported lower MVC values for cases than for controls. The force level during the sustained contraction would thus be relatively higher for cases than for controls. Thus, it would be expected that cases encounter more fatigue than controls. However, the results of our study indicate that cases show *less* myoelectric signs of fatigue. When relative force levels would have been used, this effect might have been even stronger.

5. Conclusion

The aim of the present work was to investigate differences in myoelectric manifestations of fatigue development between cases and controls during a low force level sustained contraction. Cases showed a less pronounced myoelectric response to the fatiguing task than controls, suggesting that additional higher-threshold MUs are recruited. The results are in line with the hypothesis that part of the MUs in cases was already (chronically) fatigued at the start of the experiment.

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References

Andreassen S, Arendt-Nielsen L. Muscle fibre conduction velocity in motor units of the human anterior tibial muscle: a new size principle parameter. J Physiol 1987;391(1):561–71.

Bilodeau M, Schindler-Ivens S, Williams DM, Chandran R, Sharma SS. EMG frequency content changes with increasing force and during fatigue in the quadriceps femoris muscle of men and women. J Electromyogr Kinesiol 2003;13(1):83–92.

Bongers PM, deVet HC, Blatter BM. Repetitive strain injury (RSI): occurrence, etiology, therapy and prevention. Nederlands tijdschrift voor geneeskunde 2002;146(42):1971–6.

Conwit RA, Stashuk D, Tracy B, McHugh M, Brown WF, Metter EJ. The relationship of motor unit size, firing rate and force. Clin Neurophysiol 1999;110(7):1270–5.

De Luca CJ. Myoelectrical manifestations of localized muscular fatigue in humans. Crit Rev Biomed Eng 1984;11(4):251–79.

Dumitru D, King JC, Zwarts MJ. Determinants of motor unit action potential duration. Clin Neurophysiol 1999;110:1876–82.

Elert JE, Rantapää-Dahlqvist SB, Henriksson-Larsen K, Lorentzon R, Gerdle BU. Muscle performance, electromyography and fibre type composition in fibromyalgia and work-related myalgia. Scand J Heumatol 1992;21(1):28–34.

Enoka RM, Robinson GA, Kossev AR. Task and fatigue effects on low-threshold motor units in human hand muscle. J Neurophysiol 1989;62(6):1344–59.

Farina D, Fortunato E, Merletti R. Noninvasive estimation of motor unit conduction velocity distribution using linear electrode arrays. IEEE Trans Bio-med Eng 2000;47(3):380–8.

Farina D, Schulte E, Merletti R, Rau G, Disselhorst-Klug C. Single motor unit analysis from spatially filtered surface electromyogram signals. Part I: spatial selectivity. Med Biol Eng Comput 2003;41(3):330–7.

Feiereisen P, Duchateau J, Hainaut K. Motor unit recruitment order during voluntary and electrically induced contractions in the tibialis anterior. Exp Brain Res 1997;114(1):117–23.

Gazzoni M, Farina D, Merletti R. A new method for the extraction and classification of single motor unit action potentials from surface EMG signals. J Neurosci Meth 2004;136(2):165–77.

Gift AG. Visual analogue scales: measurement of subjective phenomena. Nurs Res 1989;38:286–8.

Hagberg M, Kvarnström S. Muscular endurance and electromyographic fatigue in myofascial shoulder pain. Arch Phys Med Rehab 1984;65(9):522–5.

Hägg GM. Static work loads and occupational myalgia – a new explanation model. Amsterdam, Netherlands: Elsevier Science; 1991, 141–143.

Hägg GM, Åström A. Load pattern and pressure pain threshold in the upper trapezius muscle and psychosocial factors in medical secretaries with and without shoulder/neck disorders. Int Arch Occup Env Health 1997;69:423–32.

Hansson GA, Stromberg U, Larsson B, Ohlsson K, Balogh I, Moritz U. Electromyographic fatigue in neck/shoulder muscles and endurance in women with repetitive work. Ergonomics 1992;35(11):1341–52.

Henneman E, Somjen G, Carpenter DO. Functional significance of cell size in spinal motoneurons. J Neurophysiol 1965;28:560–80.

Hermens HJ, van Bruggen TAM, Baten CTM, Rutten WLC, Boom HBK. The median frequency of the surface EMG power spectrum in relation to motor unit firing and action potential properties. J Electromyogr Kinesiol 1992;2:15–25.

Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G. Development of recommendations for SEMG sensors and sensor placement procedures. J Electromyogr Kinesiol 2000;10(5):361–74.

Hildebrandt VH, Bongers PM, van Dijk FJ, Kemper HC, Dul J. Dutch Musculoskeletal Questionnaire: description and basic qualities. Ergonomics 2001;44(12):1038–55.

Hogrel JY. Use of surface EMG for studying motor unit recruitment during isometric linear force ramp. J Electromyogr Kinesiol 2003;13(5):417–23.

Houtman CJ, Stegeman DF, Van Dijk JP, Zwarts MJ. Changes in muscle fiber conduction velocity indicate recruitment of distinct motor unit populations. J Appl Physiol 2003;95:1045–54.

Jensen C. Development of neck and hand-wrist symptoms in relation to duration of computer use at work. Scand J Work, Environ Health 2003;29(3):197–205.

Jensen BR, Pilegaard M, Sjøgaard G. Motor unit recruitment and rate coding in response to fatiguing shoulder abductions and subsequent recovery. Eur J Appl Physiol 2000;83(2–3):190–9.

Kallenberg LAC, Hermens HJ. MUAP shape properties and MUAP Rate in cases with work-related chronic pain. Eur J Appl Phys 2006;96:203–8.

Kallenberg LAC, Hermens HJ. Behaviour of motor unit action potential rate, estimated from multi-channel surface EMG, as a measure for motor control. J Neuroeng Rehab, submitted for publication.

Lariviere C, Arsenault AB, Gravel D, Gagnon D, Loisel P. Surface electromyography assessment of back muscle intrinsic properties. J Electromyogr Kinesiol 2003;13(4):305–18.

Larsson B, Bjork J, Elert J, Gerdle B. Mechanical performance and electromyography during repeated maximal isokinetic shoulder forward flexions in female cleaners with and without myalgia of the trapezius muscle and in healthy controls. Eur J Appl Physiol 2000;83(4–5):257–67.

Lindstrom LH, Magnusson RI. Interpretation of myoelectric power spectra: a model and its applications. IEEE Trans Biomed Eng 1977;65(5):653–62.

Lundberg U, Dohns IE, Melin B, Sandsjö L, Palmerud G, Kadefors R, Ekstrom M, Parr D. Psychophysiological stress responses, muscle tension, and neck and shoulder pain among supermarket cashiers. J Occupat Health Psychol 1999;4(3):245–55.

Madeleine P, Farina D, Merletti R, Arendt-Nielsen L. Upper trapezius muscle mechanomyographic and electromyographic activity in humans during low force fatiguing and non-fatiguing contractions. Eur J Appl Physiol 2002;87(4–5):327–36.

Madeleine P, Lundager B, Voigt M, Arendt-Nielsen L. The effects of neck-shoulder pain development on sensory-motor interactions among female workers in the poultry and fish industries. A prospective study. International archives of occupational and environmental health 2003;76(1):39–49.

Masuda T, De Luca CJ. Recruitment threshold and muscle fiber conduction velocity of single motor units. J Electromyogr Kinesiol 1991;1:116–23.

Maton B, Gamet D. The fatigability of two agonistic muscles in human isometric voluntary submaximal contraction: an EMG study. II. Motor unit firing rate and recruitment. Eur J Appl Physiol Occupat Physiol 1989;58(4):369–74.

Merletti R, Knaflitz M, De Luca CJ. Myoelectric manifestations of fatigue in voluntary and electrically elicited contractions. J Appl Physiol 1990;69(5):1810–20.

Merletti R, Farina D, Gazzoni M, Schieroni MP. Effect of age on muscle functions investigated with surface electromyography. Muscle Nerve 2002;25(1):65–76.

Moritani T, Muro M, Nagata A. Intramuscular and surface electromyogram changes during muscle fatigue. J Appl Physiol 1986;60(4):1179–85.

Nordander C, Willner J, Hansson GA, Larsson B, Unge J, Granquist L, Skerfving S. Influence of the subcutaneous fat layer, as measured by ultrasound, skinfold calipers and BMI, on the EMG amplitude. Eur J Appl Physiol 2003;89(6):514–9.

Öberg T, Sandsjo L, Kadefors R, Larsson SE. Electromyographic changes in work-related myalgia of the trapezius muscle. Eur J Appl Physiol Occupat Physiol 1992;65(3):251–7.

Oddsson LI, De Luca CJ. Activation imbalances in lumbar spine muscles in the presence of chronic low back pain. J Appl Physiol 2003;94(4):1410–20.

Olsen HB, Christensen H, Sogaard K. An analysis of motor unit firing pattern during sustained low force contraction in fatigued muscle. Acta Physiol Pharmacol Bulgarica 2001;26(1–2):73–8.

Paoli P, Parent-Thirion A. Working conditions in the acceding and candidate countries, European foundation for the improvement of working and living conditions. Office for official publications of the European Community, Luxembourg: Dublin, Ireland, Publisher; 2003.

Queisser F, Bluthner R, Brauer D, Seidel H. The relationship between the electromyogram-amplitude and isometric extension torques of neck muscles at different positions of the cervical spine. Eur J Appl Physiol Occupat Physiol 1994;68(1):92–101.

Roeleveld K, Stegeman DF, Falck B, Stalberg EV. Motor unit size estimation: confrontation of surface EMG with macro EMG. Electroencephalogr Clin Neurophysiol 1997;105(3):181–8.

Schulte E, Farina D, Rau G, Merletti R, Disselhorst-Klug C. Single motor unit analysis from spatially filtered surface electromyogram signals. Part 2: conduction velocity estimation. Med Biol Eng Comput 2003;41:338–45.

Schulte E, Kallenberg LA, Christensen H, Disselhorst-Klug C, Hermens HJ, Rau G, Sogaard K. Comparison of the electromyographic activity in the upper trapezius and biceps brachii muscle in subjects with muscular disorders: a pilot study. Eur J Appl Phys 2006b;96:185–93.

Schulte E, Miltner O, Junker E, Rau G, Disselhorst-Klug C. Upper trapezius muscle conduction velocity during fatigue in subjects with and without work-related muscular disorders: a non-invasive high spatial resolution approach. Eur J Appl Physiol 2006a;96:194–202.

Seki K, Narusawa M. Relation between the size of motor units and the spectral characteristics of their action potentials. Electroencephalogr Clin Neurophysiol 1998;109(5):436–43.

Simes RJ. An improved Bonferroni procedure for multiple tests of significance. Biometrika 1986;73:751–4.

Staudenmann D, Kingma I, Stegeman DF, van Dieen JH. Towards optimal multi-channel EMG electrode configurations in muscle force estimation: a high density EMG study. J Electromyogr Kinesiol 2005;15(1):1–11.

Veiersted KB. Sustained muscle tension as a risk factor for trapezius myalgia. Int J Ind Ergon 1994;14:333–9.

Veiersted KB, Westgaard RH, Andersen P. Pattern of muscle activity during stereotyped work and its relation to muscle pain. Int Arch Occupat Environment Health 1990;62(1):31–41.

Veiersted KB, Westgaard RH, Andersen P. Electromyographic evaluation of muscular work pattern as a predictor of trapezius myalgia. Scand J Work, Environ Health 1993;19(4):284–90.

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