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Behaviour of a surface EMG based measure for motor control: Motor unit action potential rate in relation to force and muscle fatigue

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Abstract

Surface electromyography parameters such as root-mean-square value (RMS) and median power frequency (FMED) are commonly used to assess the input of the central nervous system (CNS) to a muscle. However, RMS and FMED are influenced not only by CNS input, but also by peripheral muscle properties. The number of motor unit action potentials (MUAPs) per second, or MUAP Rate (MR), being the sum of the firing rates of the active motor units, would reflect CNS input solely. This study explored MR behaviour in relation to force and during a fatiguing contraction in comparison to RMS and FMED.

In the first experiment ($n = 10$) a step contraction of shoulder elevation force (20–100 N) was performed while multi-channel array EMG was recorded from the upper trapezius muscle. The sensitivity of MR for changes in force (1.8%/N) was almost twice as high as that of RMS (0.97%/N), indicating that MR may be more suitable for monitoring muscle force. The second experiment ($n = 6$) consisted of a 15-min isometric contraction of the biceps brachii. MR increased considerably less than RMS (0.9% vs. 4.1%), suggesting that MR selectively reflects central motor control whereas RMS also reflects peripheral changes. These results support that, at relatively low force levels, MR is a suitable parameter for non-invasive assessment of the input of the CNS to the muscle.

Keywords: Multi-channel surface EMG; Motor control; Motor unit action potentials; Healthy subjects

1. Introduction

The electrical activity accompanying muscle contractions can be measured non-invasively by means of surface electrodes placed at the skin above a muscle (surface electromyography, SEMG). Variables related to the amplitude of the SEMG signal (e.g., root-mean-square, RMS, value) and to its frequency content (e.g., median frequency of the power spectrum, FMED) are commonly used in movement analysis to assess physiologically relevant aspects such as muscle activation level and development of muscle fatigue.

It is well-known that RMS increases with force, but different shapes of this relationship have been reported. Gerdle et al. (1991) reported a linear relation between normalized RMS and normalized torque for rectus femoris, vastus medialis and vastus lateralis over the complete range of force. In contrast, another study reported a lower slope of the RMS–force relation at force levels up to 40% MVC than for higher force levels (Hagberg and Hagberg, 1989). A similar pattern was found for the back muscles (Kumar and Narayan, 2001).

During isometric, isotonic sustained contractions, in general RMS increases and FMED decreases (see, e.g., De Luca, 1984; Merletti et al., 1990; Merletti et al., 2002; Madeleine et al., 2002; Oberg et al., 1992). These changes are considered to be myoelectric manifestations of muscle

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fatigue, related to additional motor unit recruitment and/or a decrease of muscle fiber conduction velocity.

When interpreting changes in RMS and FMED it should be considered that they are influenced by both central motor control properties (e.g., changes in recruitment) as well as peripheral muscle properties (e.g., changes in shape and duration of the intracellular action potential, motor unit size and muscle fiber conduction velocity). Interpretation of the changes in RMS and FMED in terms of specific physiological mechanisms is therefore not straightforward.

Recent research has shown that motor unit action potentials (MUAPs) can be isolated from the SEMG when recorded with electrode arrays (Gazzoni et al., 2004; Disselhorst-Klug et al., 2000; Stegeman et al., 2000; Merletti et al., 2003). This provides the opportunity to selectively assess the input from the central nervous system (CNS) to a muscle (Kallenberg and Hermens, 2006a). The CNS controls muscle activity by recruitment of MUs and by changing their firing rate. The number of MUs and their firing rate are reflected in the number of motor unit action potentials (MUAPs) per second (MUAP Rate, MR).

A previous simulation study showed that there is a strong, monotonously increasing relation between MR and both number of MUs and firing rate, while MR is not affected by MU size (a peripheral property). The objective of the present study was to explore the behaviour of MR in experimental conditions in comparison to two conventional SEMG parameters (RMS and FMED). This was performed during a step contraction with increasing force levels, and during a fatiguing contraction.

2. Methods

Two experiments were performed. In the first experiment, the relations of MR and RMS with force were investigated. Subjects had to perform a step contraction consisting of shoulder elevation force steps from 20 to 100 N.

The behaviour of MR, RMS and FMED during a fatiguing contraction was investigated in the second experiment, which consisted of a sustained isometric contraction of 15 min at 10% MVC of the biceps brachii.

2.1. Subjects

A Dutch questionnaire about work and health (Hildebrandt et al., 2001) was used to select subjects. This questionnaire comprised questions about work history and vocational satisfaction, as well as questions about health, i.e., history, duration and location of complaints and history of therapy. Only the health-related questions were used to select subjects. Subjects were included when they did not have any self-reported complaints in the neck, shoulders, arms or upper back during the last year. Ten subjects (5 male, 5 female, mean age 31.0, standard deviation (SD) 11.6 years, mean height 179.7, SD 10.9 cm, mean weight 69.6, SD 9.8 kg, mean body mass index 21.5, SD 1.6 kg/m^2) participated in the first experiment. Six subjects (3 male, 3 female, mean age 22.9, SD 2.2 years, mean height 179.5, SD 4.2 cm, mean weight 67.8, SD 6.1 kg, mean body mass index 21.1, SD 1.8 kg/m^2) participated in the second experiment. The study was approved by the local medical ethics committee and all subjects signed a written informed consent.

Part of the data has been used for an earlier publication (Kallenberg et al., 2006).

2.2. General procedures

In the first experiment, shoulder elevation force and SEMG of the upper trapezius muscle were measured simultaneously. Subjects performed a step contraction consisting of five force levels (20, 40, 60, 80, 100 N, corresponding to approximately 5–30% MVC, Schulte et al., 2006). A repetition of the second step (40 N) was added as sixth step to investigate possible myoelectric manifestations of fatigue, reflected as differences in the SEMG parameters of the second and sixth steps. Each step had to be maintained for 10 s while one second was used for transition to the next step.

Subjects were seated on an adjustable chair that was high enough to prevent them from touching the floor with their feet. 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 shoulder elevation force on both sides. Subjects were instructed to perform the contractions symmetrically. They were not allowed to speak or to move the head during the recordings, they had to sit straight and keep their hands rested in their lap. Subjects were not allowed to cross their feet. The position of the force sensors was adjusted to body size, such that the sensor center was located slightly above the acromion. In rest, the force sensors were just not touching the subject.

In the second experiment, subjects had to maintain a force level of 10% of the maximum voluntary contraction (MVC) of the biceps brachii for 15 min. A low force level was chosen because in such conditions conventional SEMG parameters give inconsistent results; the decrease of muscle fiber conduction velocity of the fatiguing MUs might be counteracted by additional recruitment of fresh MUs. MR seems especially suitable in this condition because it would selectively reflect the input of the CNS to the muscle.

Before the electrodes were placed, MVC was determined according to the recommendations of Mathiassen et al. (1995). Subjects were asked to perform a maximal contraction three times with 2-min rest in between. Verbal encouragement as well as real-time feedback of the force level was provided. When the third measurement was more than 10% higher than the highest of the first two, a fourth and if necessary, fifth measurement was performed. The force signal was averaged with a 100 ms moving window. The maximum value was considered the MVC.

Subjects were seated on a chair with adjustable height. The lower arms were supported in a horizontal position with the hand in the neutral position (midway between pronation and supination). The elbow was kept in 90° flexion and the upper arm was slightly abducted while it remained in the plane of the trunk. A non-elastic strap was used to connect the wrist to the force transducer, which was fixed to the floor. During the isometric contractions of the biceps brachii, subjects were asked to pull on the strap without changing the elbow position. Subjects were instructed to relax their lower arms. Perceived fatigue was measured with a 10-point Likert scale before and after the 15 min-contraction.

In both experiments, force feedback was provided on a laptop screen in front of the subject. The gain of the force feedback was adapted such that deviations of 1 N could clearly be seen. The force signals were sampled with 1 kHz, digitized with a 16-bits A/D converter, and stored on a laptop.

2.3. SEMG recordings

SEMG of the dominant upper trapezius was recorded using a two-dimensional 16-channel array developed by the Helmholtz-Institute for Biomedical Engineering, Technical University Aachen, Aachen, Germany (Disselhorst-Klug et al., 2000), see Fig. 1. The array consisted of four columns 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.

Before electrode placement, the skin was cleaned using abrasive paste. Electrode placement was done in accordance with the SENIAM recommendations for SEMG recordings (Hermens et al., 2000). In the first experiment, electrodes were placed on the trapezius of the dominant side with the columns parallel to the line from the spinous process of the seventh cervical vertebra (C7) to the acromion with the centre of the electrode 2 cm lateral from the midpoint. In the second experiment, electrodes were placed on the biceps brachii with the columns parallel to the line from the acromion to the fossa cubit, with the center of the electrode placed at one third from the fossa cubit. For both muscles, this resulted in placement of the electrode between the innervation zone and the tendon, such that propagating MUAPs were recorded. This was checked by visual inspection, and if necessary electrodes were repositioned. The side on which the electrode was placed (dominant or non-dominant) was randomised. In both experiments, a ground electrode was placed at the wrist.

The monopolar signals were amplified with a gain of 1000 and band-pass filtered (10–500 Hz) with a custom made SEMG 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, digitized using a 16 bit A/D-converter (National Instruments) and stored on a portable PC. 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 columns in parallel to the muscle fibers.

2.4. Data analysis

All data was offline band-pass filtered with a second-order zero phase shift Butterworth filter (10–400 Hz). For calculation of MR and MUAP shape properties, single differential signals with an IED of 10 mm were constructed from the two middle rows of monopolarly recorded signals (see Fig. 2). This resulted in two sets of four unidirectionally propagating single differential signals. For both of these sets, cross-correlation between adjacent signals was calculated (resulting in three values from each set) and the set with the highest average correlation coefficient was selected for further processing.

MUAPs were detected with a method that used the Continuous Wavelet Transform to identify shapes that were similar to a mother wavelet (i.e., the first-order Hermite–Rodriguez function). The algorithm separated the MUAPs from the surrounding background activity. The algorithm searched for candidate MUAPs on all channels. A candidate had to occur in at least three channels before being called a MUAP. Outcome 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), Farina et al. (2000), Kallenberg and Hermens (2006a).

MR was calculated for adjacent, non-overlapping epochs of one second throughout the duration of the contractions as the

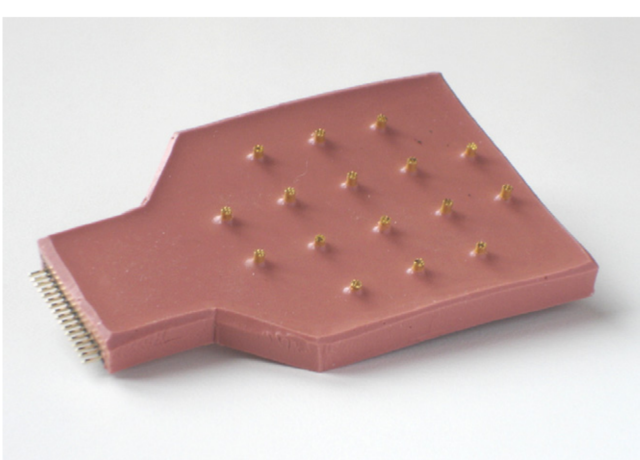

Fig. 1. Two-dimensional 16-channel electrode array used for EMG recordings. The inter-electrode distance is 10 mm. The array was placed parallel to the line from C7 to the acromion.

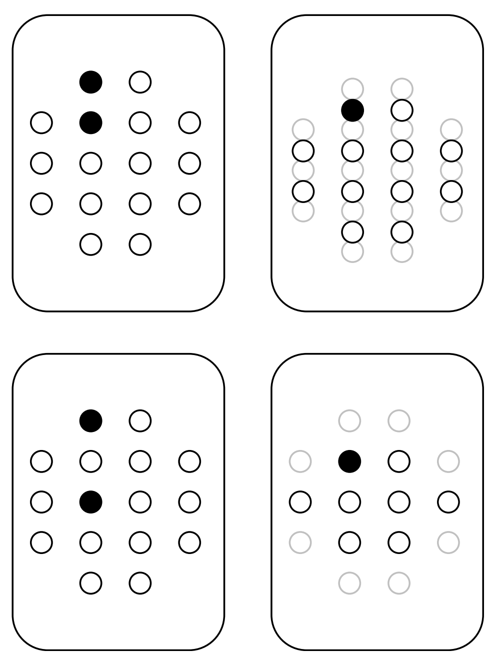

Fig. 2. Construction of single differential signals with 10 mm (upper) and 20 mm (lower) inter-electrode distance from the monopolar recordings. The locations of the monopolar recordings are shown on the left part. The grey circles on the right part indicate the location of the original monopolar recordings. The black-lined circles indicate the (virtual) location of the constructed signals. The filled circles show an example of the construction.

number of detected MUAPs per second. Thus, MR reflects the sum of the firing rates of all contributing MUs.

In addition, the RMS value (RMS_{MUAP}) and the median frequency of the power spectrum ($FMED_{MUAP}$) of each detected MUAP was calculated (Kallenberg and Hermens, 2006a). Histograms of these parameters were used to examine peripheral properties of the MU population. RMS_{MUAP}, related to the depth under the electrode and to the size of the MUs, was calculated by taking the square root of the sum of all squared data samples of the MUAP, divided by the number of samples. $FMED_{MUAP}$ reflects the frequency content of the MUAPs, which is mainly related to the MUAP duration (Hermens et al., 1992) and muscle fiber conduction velocity (Lindstrom and Magnusson, 1977; Dumitru et al., 1999; Arendt-Nielsen and Mills, 1985). $FMED_{MUAP}$ was calculated as the median value of the power spectrum, obtained using the fast Fourier-transform with a rectangular window. The MUAP shapes were zero-padded to obtain a frequency resolution of 1 Hz. $FMED_{MUAP}$ and RMS_{MUAP} were calculated from each of the four single differential channels, and the values were averaged across the channels afterwards.

For analysis of global SEMG parameters, three single differential signals with an IED of 2 cm were constructed from the monopolar signals by subtracting signals with 2 cm in between in the direction parallel to the muscle fibers, in accordance with the SENIAM guidelines for conventional SEMG (Hermens et al., 2000); see Fig. 2. The signals were constructed from the same set of monopolar signals as used for MR estimation. 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 1 s for each of the three signals. Average values across the three signals were calculated.

For the step contraction of the trapezius muscle, the SEMG parameters (RMS_G, $FMED_G$, MR, RMS_{MUAP}, $FMED_{MUAP}$) were averaged for each force step (across the 10-s duration). For the sustained contraction of the biceps brachii, averages were calculated per 30 s.

2.5. Statistical analysis

Linear regression analysis was applied for each individual subject for investigating relations between SEMG parameters and force (excluding the sixth step of the step contraction that was a repetition of the second step). The slopes of the regression lines (normalized to the value at the third force step) are reported as measure for the sensitivity of the SEMG parameters for changes in force.

For comparison of the sixth and second step of the step contraction, normality of the distributions of RMS_G and MR was checked with the Kolmogorov–Smirnov test prior to statistical testing. This data was normally distributed, therefore a Student's t-test for paired samples was used to test for differences.

A mixed linear model was chosen to analyse the SEMG parameters during the sustained contraction. Mixed linear models are designed to handle correlated data, e.g., including multiple observations within each subject. For the 15 min sustained contraction, parameters were extracted for 30 s windows, resulting in 30 observations per subject. Besides factors that describe the average behaviour of the group of subjects (fixed factors), the model allows the inclusion of separate noise terms for each subject and for each measurement within a subject (random factors). Restricted maximum likelihood estimation was used for estimation of the parameters. Time was included as a covariate because it is a continuous factor, arbitrarily quantified in 30 s periods. Because of its influence on RMS (Nordander et al., 2003), body mass index was included as covariate as well. A random intercept for each subject was included. Furthermore, since the increase or decrease in SEMG parameters may be different for each subject, a random term for the slope in time for each subject was included. The model can mathematically be described as

$$k = \beta_0 + \beta_1 \mathrm{time} + \beta_2 \mathrm{BMI} + \mu_0 + \mu_1 \mathrm{time} + \varepsilon$$

With k the measurement data, β_i the coefficients for the fixed factors, μ_0 the random intercept per subject, μ_1 the random term for the slope per subject and ε the remaining noise.

The model was applied to calculate the significance of the factors. Increases or decreases in parameters were estimated by applying the model again with only the significant factors included (manual backwards elimination). Means and confidence intervals (CI) of the parameter estimates are reported.

3. Results

The cross-correlation coefficient between the channels was 0.87 (SD 0.13) for the data of the first experiment. It was only in one case necessary to exclude a signal epoch (3 s) from the analysis due to artefacts in the SEMG signals. All subjects were able to follow the force steps of the step contraction to a reasonable extent: the mean force levels were within 2% of the required force levels. Standard deviations were less than 5% of the mean force levels. See Fig. 3 for an example of a force curve.

In Fig. 4, RMS_G and MR during the step contraction are shown. In the first five steps, the increase of RMS_G is approximately linear. Remarkably, RMS_G of the sixth step (i.e., the repetition of the second step) was much higher than that of the second step (Student's t-test for paired samples, $p < 0.001$). Individual linear regression models including force and an intercept explained on average 96% (range 79–99%, $0.001 < p < 0.11$) of the variance in

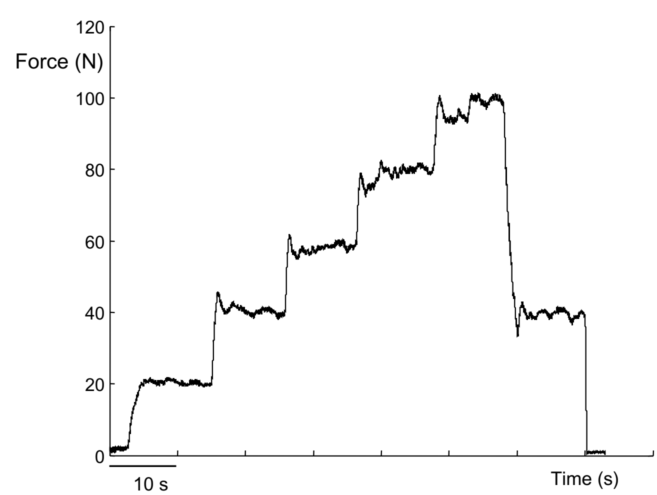

Fig. 3. Typical example of force curve. Force steps of 20, 40, 60, 80 and 100 N were maintained for 10 s each, with 1 s in between for transition to the next step. The sixth step was a repetition of the second step (40 N).

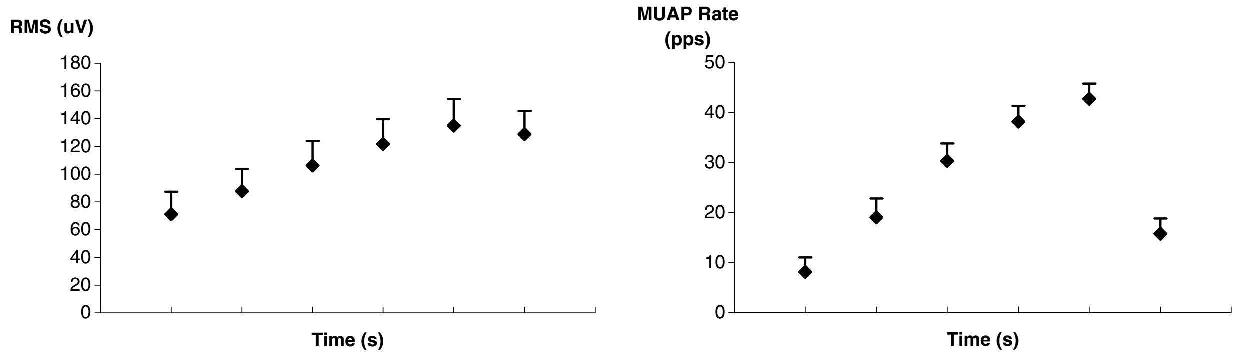

Fig. 4. Global RMS (left) and MR (right) calculated from EMG recordings of the upper trapezius during the step contraction of shoulder elevation force. Force increased from 20 to 100 N in steps of 20 N. The sixth step was a repetition of the second step (40 N). Black diamonds show the average RMS and MR values for each force step, bars show standard errors of the mean.

RMS_G. The sensitivity of RMS_G for changes in force, expressed as the normalized slope of the regression line, was on average 0.97%/N (range 0.27–1.8%/N).

MR increased linearly with force up to 80 N, while the increase in the last step was smaller. The values of MR for the second and the sixth step (both 40 N) were not different (Student's *t*-test for paired samples, $p = 0.39$). Individual linear regression models including force and an intercept explained on average 94% (range 88–97%, $0.002 < p < 0.019$) of the variance in MR. The sensitivity of MR for force, expressed as the normalized slope of the regression line, was almost twice as high as the sensitivity of RMS_G: on average 1.8%/N (range 0.73–3.1%/N).

Note that for both RMS_G and MR the standard errors of the mean are rather constant across the whole force range. The standard errors of MR are relatively small compared to its dynamic range.

In Fig. 5, histograms of RMS_{MUAP} (left) and $FMED_{MUAP}$ (right) of the detected MUAPs are shown for each step. The distribution of RMS_{MUAP} expands towards

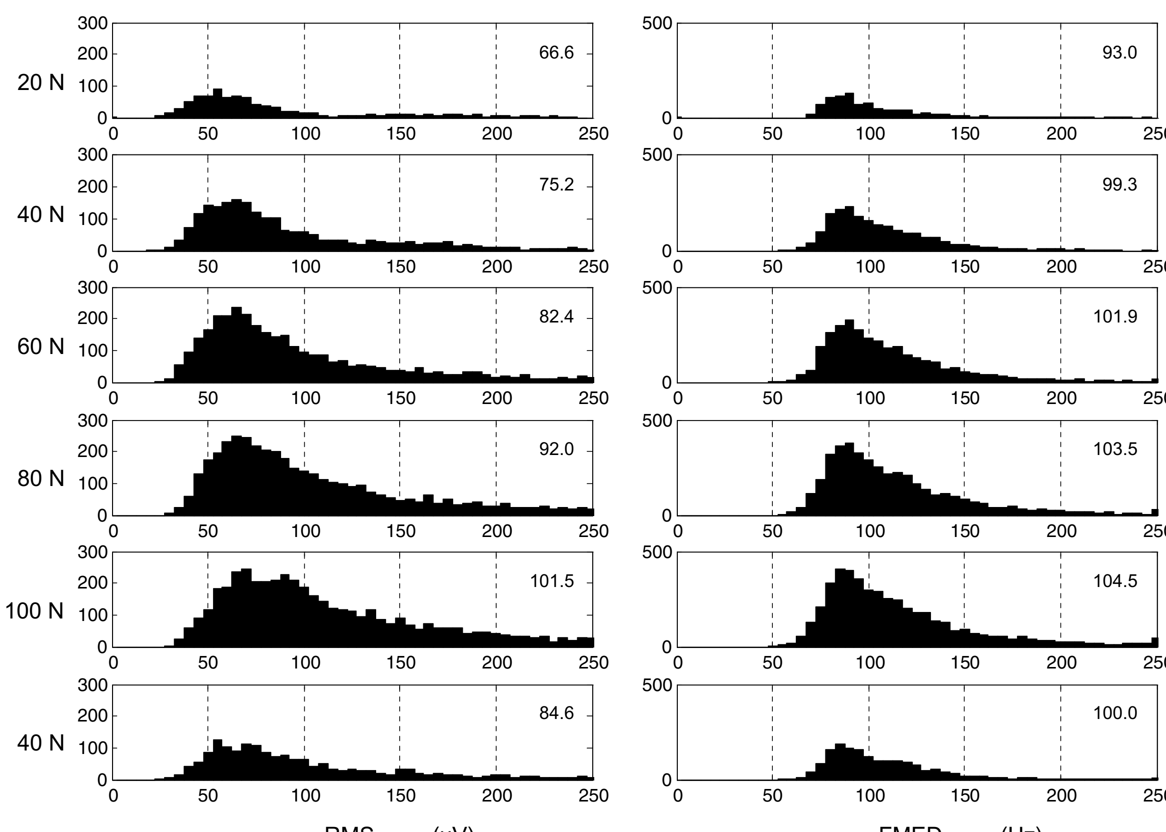

Fig. 5. Distributions of RMS_{MUAP} and $FMED_{MUAP}$ of the MUAPs detected in EMG recordings of the trapezius muscle for the five different force steps (20–100 N). The sixth step was a repetition of the second step (40 N). Median values of the histograms are reported in the upper right corners.

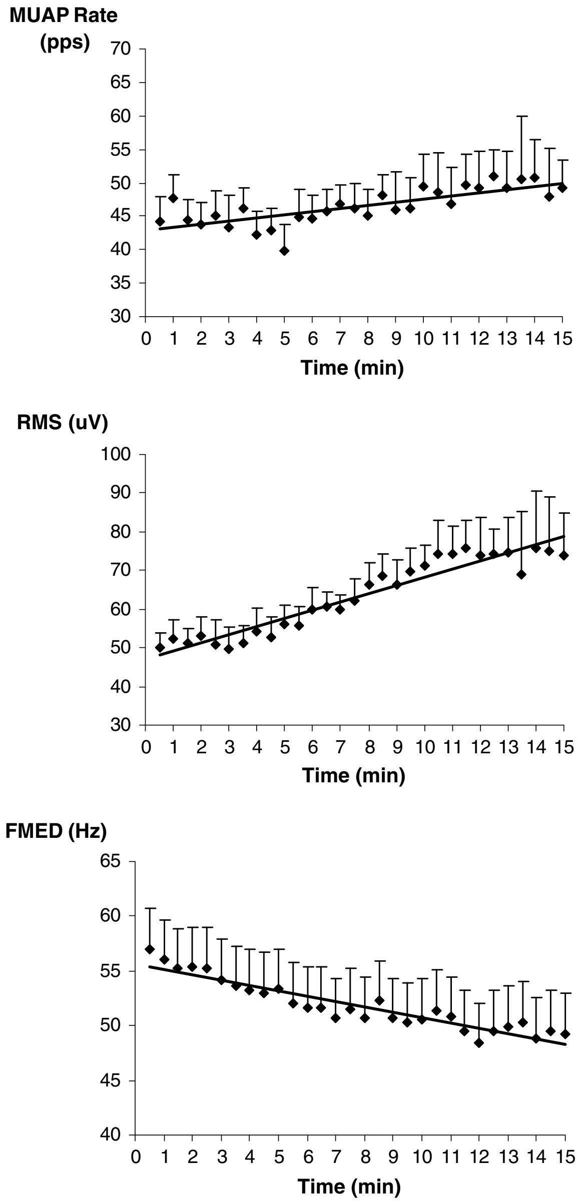

Fig. 6. Changes in time of MR, global RMS and global FMED during a 15-min sustained contraction of the biceps muscle at 10% MVC. Diamonds show mean values, bars show standard errors of the mean.

higher values when the force increases, as is also reflected in the median values of the RMS_{MUAP} histograms. The low RMS_{MUAP} values are still present at higher force steps while higher RMS_{MUAP} values are added to the distribution when the force increases. The distribution of $FMED_{MUAP}$ changes less with force than that of RMS_{MUAP}. The median values of the histograms show that there is a slight shift towards higher $FMED_{MUAP}$ values with force. The shapes of the histograms of $FMED_{MUAP}$ for the second and sixth step are rather similar, while there are relatively more high RMS_{MUAP} values at the sixth than at the second step.

The behaviour of MR during a fatiguing contraction was investigated in the second experiment. The Likert scale scores (range 0–10) of perceived fatigue were 1.0 (SD 1.55) before, and 6.0 (SD 2.37) after the contraction. The mean MVC of the biceps brachii was 184 N (SD 45.6).

For the data of the second experiment, the cross-correlation coefficient between the channels was 0.73 (SD 0.14). For two subjects, there were a few (1–3) short periods of the signal that contained artefacts. These periods were left out from the data analysis.

MR and RMS_G both increased and $FMED_G$ decreased with time (Fig. 6). MR increased significantly with time with 0.91%/min (0.39 pps/min, CI 0.0076–0.78, $p < 0.043$). The intercept at the beginning of the contraction was 43.0. The mean SD across subjects averaged over time was 9.91 pps. RMS also increased significantly with time with 4.1%/min (2.0 μV/min, CI 0.42–3.5, $p < 0.022$) and its intercept was 47.6 μV. The mean SD was 15.5 μV. FMED decreased significantly with time with 0.88%/min (0.49 Hz/min, CI −0.70 to −0.28, $p < 0.002$) with an intercept of 55.7 Hz. The mean SD was 9.04 Hz.

The distributions of RMS_{MUAP} and $FMED_{MUAP}$ of the detected MUAPs during the first 30 s of the contraction are shown in Fig. 7 (upper graphs). The RMS_{MUAP} values are lower than the RMS_{MUAP} values extracted from the trapezius signals at 20 N (lower graphs), while the force level is similar: the biceps brachii contraction was performed at 10% MVC, which on average corresponds to 18 N. The median RMS_{MUAP} value of the biceps (37.9 μV) is also much lower than that of the trapezius (66.6 μV), while in

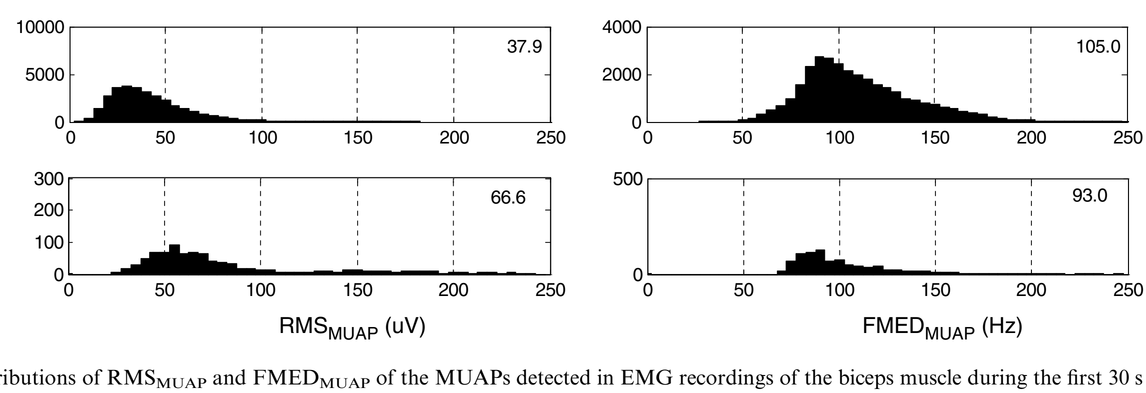

Fig. 7. Distributions of RMS_{MUAP} and $FMED_{MUAP}$ of the MUAPs detected in EMG recordings of the biceps muscle during the first 30 s of a sustained contraction at 10% MVC (upper graphs). The distributions of RMS_{MUAP} and $FMED_{MUAP}$ from the trapezius muscle at the force step of 20 N are shown below for comparison (lower graphs). Median values of the histograms are reported in the upper right corners.

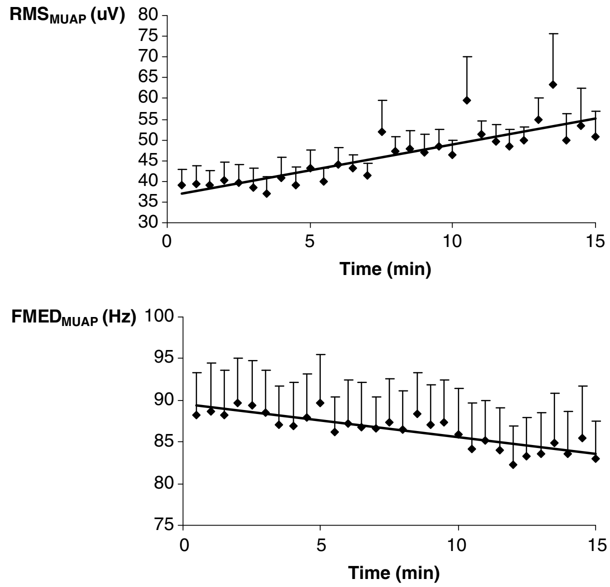

Fig. 8. Changes in time of RMS_{MUAP} and $FMED_{MUAP}$ during a 15-min sustained contraction of the biceps muscle at 10% MVC. Diamonds show mean values, bars show standard errors of the mean.

contrast the median $FMED_{MUAP}$ value is somewhat higher in biceps (105.0 Hz) than in trapezius (93.0 Hz).

In Fig. 8, the changes over time in MUAP properties are reported. RMS_{MUAP} increases over time while $FMED_{MUAP}$ decreases. RMS_{MUAP} increased significantly with time with 3.5%/min (1.26 μV/min, CI 0.148–2.38, $p < 0.034$). The intercept at the beginning of the contraction was 36.3 μV. The mean SD across subjects, averaged over time was 12.1 μV. There was trend for a decrease of $FMED_{MUAP}$ with time with 0.45%/min (0.40 Hz/min, CI −0.83 to 0.020, $p < 0.058$) and its intercept was 89.6 Hz. The mean SD was 12.5 Hz.

4. Discussion

The objective of this study was to explore the behaviour of MR in comparison with RMS_G and $FMED_G$ in experimental conditions.

MR was found to increase linearly with force up to a level of 80 N, probably due to a combined effect of recruitment of new motor units and an increase of firing rates of already recruited units. For higher force levels, the curve starts to flatten. This flattening may be related to at least two phenomena. Firstly, when the number of MUAPs in the signal increases, the MUAPs will start to overlap each other, causing underestimation of the number of MUAPs (see also Kallenberg and Hermens, 2006b). Secondly, when a higher force is required, generally bigger MUs are recruited (Henneman et al., 1965). These bigger MUs have a higher force output per firing, and thus, for a same increase in force, less additional MUs and/or firings are needed.

In contrast, the relation between RMS_G and force was linear over the whole range of investigated force levels, which is in agreement with findings for the upper leg muscles (Gerdle et al., 1991; Karlsson and Gerdle, 2001). Hagberg and Hagberg (1989) reported that the slope of the RMS–force curve was steeper for high force levels than for low force levels in the upper trapezius. We only measured a force range of low to moderate levels: a force of 100 N corresponds to about 25–30% of MVC, that was 357 N for healthy subjects in the same experimental setup (Schulte et al., 2006).

The sensitivity of MR to force was about twice as high as that of RMS_G. The inter-subject variability was smaller for MR than for RMS_G, as indicated by the standard errors of the mean, relative to the dynamic range. Furthermore, MR reflected the repetition of the second step better than RMS_G. These findings suggest that for low to moderate shoulder elevation force levels, MR is a better force estimator than RMS_G in the present setup.

The distribution of RMS_{MUAP} expands to higher values with increasing force. The amplitude of the MUAP depends on the size of the MU (Roeleveld et al., 1997) and on the distance between MU and electrode. Assuming a homogeneous distribution of the MUs in the muscle, the increase of RMS_{MUAP} may indicate recruitment of larger MUs. This is further supported by the increase of $FMED_{MUAP}$ with force. It was shown that $FMED_{MUAP}$ is mainly determined by muscle fiber conduction velocity (Lindstrom and Magnusson, 1977; Dumitru et al., 1999; Arendt-Nielsen and Mills, 1985) and by the duration of the MUAPs (Hermens et al., 1992). Since conduction velocity of a motor unit is correlated with its recruitment threshold (Andreassen and Arendt-Nielsen, 1987), the higher $FMED_{MUAP}$ values indicate the contribution of higher threshold MUs.

A remarkable difference was found between the second level of the step contraction and its repetition (sixth step). At the sixth step, MR returned to the value for the second step, whereas RMS_G remained much higher. This discrepancy could be related to peripheral changes caused by muscular fatigue, which affect RMS_G (Merletti et al., 1990; De Luca, 1984) but not MR. In addition, the RMS_{MUAP} histograms also show a higher median value at the sixth than at the second step. Although the explanation of these findings is not straightforward, one possibility might be that the lowering of the force step during the contraction is largely caused by a decrease in firing rate of the contributing MUs, while derecruitment of the large MUs only occurs to a limited extent. Another explanation might be that the duration of the MUAPs at the sixth level is elongated because of muscle fatigue, which would lead to an increased RMS_G, while it does not affect MR.

The second objective was to explore the behaviour of MR in comparison with RMS_G and $FMED_G$ in relation to fatigue. Subjects had to perform a 15-min isometric contraction of the biceps brachii at 10% of their MVC. Although this is a rather low force level, the results of the Likert scale (mean score 6.0 after the contraction) show that the contraction was indeed perceived as fatiguing. Further indications for muscle fatigue development were found in the SEMG variables: RMS_G, RMS_{MUAP} and MR increased while $FMED_G$ and $FMED_{MUAP}$ decreased with time. The increase of RMS_G, RMS_{MUAP} and MR is most likely related to recruitment of fresh MUs. Particularly at low force levels, additional recruitment may compensate the loss of force of already active MUs.

RMS_G increased with 4.1% per minute while there was only a 0.9% per minute increase of MR. The stronger increase of RMS_G is in line with the hypothesis that RMS_G, in contrast to MR, is also influenced by peripheral properties. Balog et al. (1994) reported changes in shape, size and duration of the intra-cellular action potential in relation to fatigue. These peripheral changes are probably reflected in the surface recorded MUAPs (which is shown by the increase in RMS_{MUAP}) and thereby also in RMS_G.

When the biceps brachii contraction is compared to the trapezius contraction (first step), a remarkable finding is that RMS_G is lower for the biceps than for the trapezius, while MR is higher in biceps. This may be explained by considering the substantially lower RMS_{MUAP} values in the biceps, because RMS_G is directly related to the size of the MUAPs, while MR is not affected by MUAP size. Lower RMS_{MUAP} values may point at more deeply located or smaller MUs. In agreement, an autopsy study showed a smaller average MU diameter in biceps than in trapezius (Polgar et al., 1973). This may be related to the function of the biceps that is involved with precise control of movements, while trapezius has a more postural function.

Another method for MUAP counting, using an amplitude-based detection algorithm was developed by Zhou et al. (2003). They illustrated the relation between MR and force with recordings from three subjects, indicating an increase of MR with increasing force as well. However, they found that MUAPs with small peak amplitude that were detected at relatively low force steps (20% MVC) were not detected at higher force steps (40% MVC). They attributed this finding to the intrinsic properties of the measurement system. Contrary to this, in our study the low values of RMS_{MUAP} remain present in the distributions even at higher force steps, indicating that the algorithm is able to detect small MUAPs, even in the presence of larger MUAPs. This might be caused by differences between the algorithms. In the study of Zhou et al., in the first stage, candidate MUAPs (peaks) are detected. A threshold, derived from the amplitude distribution of the candidates is used for deciding which candidates are real MUAPs. If the candidates generally are bigger, this threshold will become higher, thus excluding small MUAPs from being selected. In the present method, the threshold is not based on amplitude, but on similarity of the MUAP with a scalable mother wavelet.

A limitation of the algorithm is its inability to resolve superimpositions of MUAPs. Superimposed MUAPs are either not recognized or detected as single MUAP. For application to higher force steps, where superimpositions occur more often, the algorithm needs to be extended with a module that can handle superimpositions as is applied in software packages for decomposition of intramuscular EMG recordings (Zennaro et al., 2003; McGill et al., 2005).

Assessment of muscle activity with electrode arrays has several advantages when compared to conventional bipolar SEMG. Recordings with an array in general can provide a more complete view of muscle activity because not only temporal, but also spatial information becomes available. The combination of array recordings with decomposition algorithms allows the extraction of information at the level of MUAPs, or even at the level of individual MUs (Holobar and Zazula, 2004; Gazzoni et al., 2004; Kleine et al., 2000). Array SEMG recordings are particularly suitable for low to moderate force levels. For higher force levels, superimposed MUAPs complicate the extraction of information. To overcome this, higher order spatial filters may be used to limit the number of MUs that contribute to the signal. However, this also limits the view of the electrode, which might make the recording less representative of the whole muscle.

5. Conclusion

In this study the behaviour of MR and MUAP shape properties in relation to force and fatigue development was explored in comparison with the commonly used SEMG measures such as RMS_G and $FMED_G$. The results suggest that MR is more sensitive to changes in shoulder elevation force than RMS_G. In a sustained contraction of the biceps brachii, MR showed a slight increase while RMS_G increased strongly. This is in line with the hypothesis that MR is largely determined by motor control properties whereas RMS_G is also strongly influenced by MU size. A limitation of the present implementation of MR estimation is that it can only be used for contractions at low to moderate force steps. Further research to explore the applicability of MR for other muscles is underway.

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