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Muscular load characterization during isometric shoulder abductions with varying force

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Abstract

This study sought to characterize muscle loading and fatigue during static shoulder abductions with varying force. In a supine posture, participants maintained fixed shoulder abductions against a time-varying external resistance, generated by a dynamometer-spring mechanism. Patterns (cumulative distribution) of the external resistance were varied by selecting different 10th and 90th percentiles of the distribution. Dynamometer angular velocities were also varied, to reflect different rates of cyclic muscle contraction. The degree of local fatigue development was assessed by common measures, including endurance time, strength reduction, and perceived discomfort. Myoelectric (EMG) signals were continuously obtained from the middle deltoid muscle throughout experimental exercise (60 min max). Changes in EMG root-mean-square (RMS) and spectral measures (derived from 1-s windows at peaks in the cyclic contractions) were used as manifestations of muscle fatigue. For each minute, the RMS signal was further reduced using two methods, the cumulative probability distribution of EMG (CPDE) and exposure variation analysis (EVA). The former resulted in three percentile values (10th, 50th, and 90th), whereas the latter method resulted in 10 different measures (grouped by EMG activity level and duration). A main finding of the study was the applicability of several common fatigue indicators for these cyclic, repetitive exertions. Overall, the use of CPDE and EVA to characterize task differences and predict muscle fatigue was found to have limited value.

Keywords: Localized muscle fatigue; Electromyography; CPDE; EVA; Exposure assessment

1. Introduction

Muscle fatigue has been defined as any reduction of the force producing capacity of muscle as a consequence of muscle activity (Christensen, 1986). Fatigue is of particular importance in ergonomics, because of effects on performance as well as assumed effects on health. Muscle fatigue has mainly been studied under isometric continuous or intermittent constant force conditions. The external load is defined in these experiments as the force produced by the subject on the environment, usually in percentages of maximum voluntary contraction force (MVC). Production of this force requires muscle activity leading to a reduction in the force producing capacity and eventually a failure to produce the required force. The rate of fatigue development is reflected in the time to force failure or endurance time. Electromyographic (EMG) based fatigue indicators, reflecting the signal's frequency content, can be used to predict the point of force failure, where such failure is determined as an inability to continue a task or based on a fixed level of force decrement (Hagberg, 1981; Dieën et al., 1993; Dieën et al., 1998; Merletti and Roy, 1996). It is believed that, during fatiguing contractions, shifts in EMG frequency spectra are partly attributable to a decrease in muscle fiber conduction velocity (Luttmann et al., 2000). It should also be noted that muscle fatigue in turn affects muscle activity. In isometric constant force contractions, the central drive increases to compensate for loss of force producing capacity. This is usually

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Nomenclature

AV	dynamometer angular velocity
CPDE	cumulative probability distribution of EMG
$CPDE_{10/50/90}$	10th, 50th, or 90th percentile of CPDE
CPDF	dynamometer cumulative probability distribution of (external) force
$CPDF_{10/90}$	10th or 90th percentile of CPDF
D	EMG activity duration
EMG	electromyography
ET	endurance time
EVA	exposure variation analysis
EVA1–9	cumulative duration of EVA relative to total recording period
FFT	Fast Fourier transform
L	EMG activity level
MnPF/MdPF	mean/median power frequency
MVC	maximum voluntary contraction
R_{CPDE}	rates of CPDE change
R_{EVA}	rates of EVA change
R_{MnPF}/R_{MdPF}	rates of Mn/MdPF change
RMS	root-mean-square of EMG signals
R_{MVC}	rates of MVC change
RPD	ratings of perceived discomfort
R_{RMS}	rates of RMS increase
R_{RPD}	rates of RPD change
R_{SumEVA}	rates of SumEVA change
SumEVA	summation across EVA 'cells'

reflected in an increase of the EMG amplitude during the contraction, although the EMG amplitude is also affected by other factors such as the change in frequency content. Such increases in activity do not, however, occur consistently. In some subjects, muscle activity will decrease (temporarily), presumably because load sharing between muscle changes (Dieën et al., 1993; Zijdewind et al., 1995; McLean and Goudy, 2004). Such changes in load sharing in sustained activities have been shown to slow down fatigue development (Dieën et al., 1993) and to disturb the relationship between the endurance time and EMG based fatigue indicators (McLean and Goudy, 2004).

It is conceivable that fatigue-related changes in muscle activation over time are less predictable under less constraining external loads than the typical constant force contractions commonly studied. The aim of the present study, therefore, was to describe changes in muscle activation as estimated from the EMG amplitude over time and the development of fatigue in force varying isometric contractions. Also of interest were the potentials to describe muscle activity (and hence characterize complex tasks in ergonomic applications) using two relatively recent EMG data processing methods, specifically the cumulative probability distribution of EMG (CPDE) and exposure variation analysis (EVA). The former employed three percentile values (10th, 50th, and 90th) to characterize the EMG amplitude, while the latter classified the EMG amplitude into 10 groups according to magnitude and duration. Further, it was of interest to determine if task descriptors yielded by these methods were related to common indicators of local fatigue.

2. Methods

2.1. Participants

Eight college-age students (4 males, 4 females) participated in this study, with mean (SD) age, stature, and body mass of 21.4 (1.6) years, 170.1 (5.0) cm, and 62.7 (1.0) kg, respectively. All reported no recent (12 months) history of musculoskeletal injuries and performing physical exercise 2–4 times a week. Participants provided informed consent prior to their participation, using procedures approved by the Virginia Tech Institutional Review Board. An initial (practice) session was given that allowed for familiarization with the experimental procedures.

2.2. Experimental procedures and apparatus

In a supine posture (Fig. 1), the participants abducted their right (dominant) arm to 90°, and placed the arm on a metal platform supported by ball-bearings. This setup minimized frictional and gravitational effects. A padded strap was worn proximally to the elbow, and was attached to a mechanism generating an external force. The mechanism consisted of a dynamometer

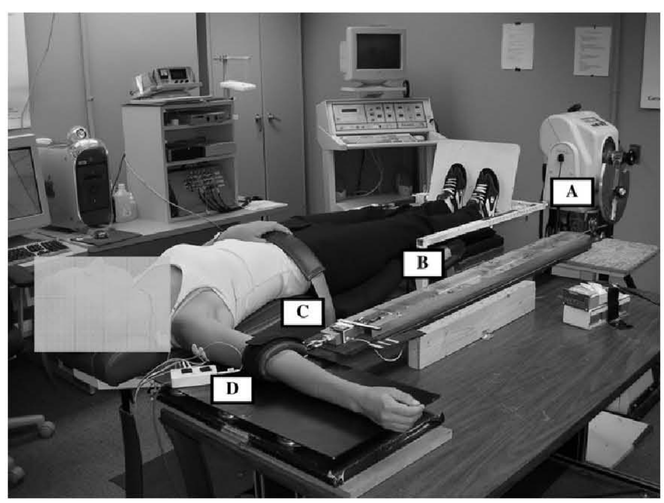

Fig. 1. Experimental setup. Dynamometer (A) moves back and forth with angular velocity and range of motion that are varied according to the experimental condition. A light metal wire connects the dynamometer, a set of springs (B), a force transducer (C), and a padded strap (D) worn at the elbow. With the arm maintained in a fixed position, movements of the dynamometer result in variable (cyclic) loading.

(Biodex System 3 Pro, Biodex Medical System, Inc., Shirley, NY, USA) with a custom-made metal arm attached to a set of metal springs that were subsequently connected to the elbow strap via a thin metal cable. In this experimental configuration, and with the participant's arm kept voluntarily in a fixed (abducted) position, cyclic rotations of the dynamometer arm produced force (resistance) that varied over time (with a roughly sinusoidal pattern), resulting in exertions that were isometric but with varying force. The 10th and 90th percentiles of the cumulative probability distribution of this external force ($CPDF_{10}$ and $CPDF_{90}$) were used to describe the workload, and were varied by changing the dynamometer initial position and its range of motion. Dynamometer angular velocity was also varied to obtain different patterns of cyclic external force.

Prior to testing, isometric maximum voluntary contraction (MVC) was determined based on the largest force obtained during four to five brief (5 s) strength test trials (gradual ramp-up, hold, and gradual ramp-down). Two minutes of rest breaks were provided between trials. This MVC was subsequently used as a basis for determining the exercise workload. The exercise involved maintaining shoulder abduction (as described above) against a cyclic resistance until exhaustion (or a maximum of one hour, whichever came earlier). A mirror was hung from the ceiling and frequent verbal feedback was given to help the participant maintain the arm in the required position.

2.3. Exercise conditions and experimental design

Three factors were investigated in the present study (Table 1), each at two levels, resulting in eight different exercise conditions. The 10th percentile of the external force distribution ($CPDF_{10}$) was set at 2.5% (Low) or 10% (High) of the individual's MVC (determined at the beginning of a session), while the 90th percentile of the distribution ($CPDF_{90}$) was set at 20% (Low) or 30% MVC (High). To reflect different rates of muscle contraction, the dynamometer angular velocity (AV) was set at 20°/s (Slow) or 45°/s (Fast). Initials of factor levels were used to label an exercise condition. For instance, 'LHF' denotes a condition comprised of a low-level of $CPDF_{10}$, a high-level of $CPDF_{90}$, and the faster AV. Each participant performed all eight exercise conditions, which were intended to represent a considerable portion of the wide variability in external loads commonly found in occupational settings. A single condition was tested in each experimental session, and a minimum of 48-h of recovery was provided between sessions to minimize possible residual fatigue. The order of exercise conditions for each individual participant was specified using a balanced Latin Square design.

Table 1
Naming of exercise conditions

	$CPDF_{10}$			
	2.5% MVC		10% MVC	
	$CPDF_{90}$		$CPDF_{90}$	
	20% MVC	30% MVC	20% MVC	30% MVC
Velocity				
20°/s	LLS	LHS	HLS	HHS
45°/s	LLF	LHF	HLF	HHF

L: Low; H: High; S: Slow; F: Fast.

2.4. Data acquisition and processing

Positions of the dynamometer arm and force signals were sampled continuously at 128 Hz, and subsequently low-pass filtered (Butterworth, second order, 6 Hz cut-off frequency). EMG signals were obtained from the middle deltoid muscle using pre-gelled bipolar Ag/AgCl electrodes (1 cm diameter) with a 2.5 cm inter-electrode distance. Selection of the muscle was based on its accessibility for surface EMG, and due to a pronounced decline in strength observed in this muscle during overhead work (Nussbaum et al., 2001).

To ensure good skin-electrode contact, the skin was shaved, lightly abraded, and cleaned with 70% rubbing alcohol. A 20-min period was provided in order to stabilize the electrodes on the skin, with an inter-electrode resistance of less than 10 kΩ considered acceptable. Electrodes were placed over the middle deltoid muscle according to Hermens et al. (2000). Bony landmarks and a tape measure were used to ensure consistent electrode locations across experimental sessions.

Continuous EMG signals were recorded over the entire exercise period and obtained using an EMG amplifier (Measurement Systems Inc., Ann Arbor, MI, USA). A pre-amplifier strengthened the signals by 100, and further amplification was done to achieve signals of $\sim\pm 5$ V. Both raw (hardware-filtered at 10–500 Hz, sampled at 2048 Hz) and root-mean-square (RMS) signals (100 ms time-constant, sampled at 128 Hz) were collected. EMG RMS was subsequently low-pass filtered (Butterworth, second order, 6-Hz cut-off frequency). In addition, levels of EMG RMS at maximum dynamometer-arm positions were estimated based on the means of 1-s data window. These maximum positions represented the greatest external forces applied to the participant's arm. Using these data windows, further processing in the frequency domain included subdividing raw EMG into three 0.5-s samples (50% overlap). Each of these samples was multiplied by a Hanning-weighted window, and later subjected to Fast Fourier Transform (FFT). The resulting frequency spectra were averaged, from which the mean (MnPF) and the median power frequency (MdPF) of the 1-s data windows were determined (Merletti and Lo Conte, 1997).

Processed EMG RMS was also used as a basis for determining CPDE and EVA measures. The 10th, 50th, and 90th percentile values of the CPDE ($CPDE_{10}$, $CPDE_{50}$, and $CPDE_{90}$) were derived from 1-min of RMS data windows, a procedure that was applied throughout exercise durations. Similarly, the same data were also used to determine nine EVA measures (Table 2), which were categorized based on combinations of levels of EMG activity (less than 10% max, between 10% and 30% max, and greater than 30% max), and the durations associated with each of the levels (less than 1 s, between 1 and 3 s, and greater than 3 s).

Table 2
EVA matrix showing the nine EVA measures (EVA1–EVA9)

	EMG activity level (*L*)		
	<10% max	10–30% max	>30% max
EMG activity duration (D)			
<1 s	EVA1	EVA4	EVA7
1–3 s	EVA2	EVA5	EVA8
>3 s	EVA3	EVA6	EVA9

2.5. Dependent measures and data analysis

Several common objective and subjective measures were used to assess the degree of exercise-induced fatigue. Endurance time (exercise duration) was noted at the end of an exercise, whereas rates of MVC reduction (R_{MVC}) were calculated according to ($MVC_{end\ of\ exercise} - MVC_{start\ of\ exercise}$)/endurance time. Ratings of perceived discomfort (RPD), using a common 10-point scale (Borg, 1990), were obtained every four minutes throughout an exercise, and were later expressed as rates of RPD increase (R_{RPD}) using simple linear regressions.

Changes in EMG (obtained at peak dynamometer-arm positions, see explanations above) were used as manifestations of local fatigue, and were defined as rates of RMS increase (R_{RMS}) and rates of decrease in mean and median power frequency (R_{MnPF} and R_{MdPF}). These rates were obtained based on slopes of linear regressions. CPDEs and EVAs derived from the initial minute of exercise characterized (non-fatigued) muscle loading, whereas rates of change for these measures (R_{CPDE} and R_{EVA}) were expected to indicate fatigue. Note that an additional EVA-based measure was determined as the absolute sum of all the nine EVAs (Sum EVA), and its changes overtime was expressed as R_{SumEVA}.

Due to the relatively small number of participants and potential departures from normality, non-parametric (Rank *F*) ANOVA was used to test the effects of the independent measures. Stepwise multiple linear regression was employed to establish specific relationships between local fatigue measures and the initial CPDEs and EVAs. Mellow's Cp criterion was used in order to determine the best subset of predictors. Significance of all statistical tests was based on $P < 0.05$. Omega squared (ω^2), a measure of effect size, was further used to determine the relative sensitivity of the different measures of local fatigue (Keppel, 1991). Presentation orders of the exercise conditions could potentially affect the response variables, but initial analysis did not indicate any significant effects ($P > 0.10$).

3. Results

3.1. Endurance and common indicators of muscle fatigue

The exercise conditions resulted in considerably different degrees of muscle fatigue. Endurance times ranged from 20 to 60 min, while mean rates of strength reduction (R_{MVC}) varied between 0.3% and 2%/min. Similar variability across conditions was also found for changes in subjective perceptions of discomfort (R_{RPD}). The light exercise conditions produced R_{RPD} of about 0.1 units/min, whereas the most fatiguing conditions yielded R_{RPD} of 6–8 times greater. Differences in external workload and rates of muscle contraction were generally associated with changes in these measures of fatigue ($P < 0.05$), though the effect of AV on R_{MVC} only approached significance ($P = 0.06$).

3.2. EMG amplitude and spectral measures

Typical manifestations of local fatigue were not found for the RMS data. For example, an increasing RMS (Table 3) was observed for one of the highest force conditions (HHS), but an opposing (decreasing) trend was found for a similar condition (HHF). Mixed patterns of RMS change were also found for several other conditions. None of the experimental factors had any significant effect ($P > 0.20$) on R_{RMS}. Relatively more consistent patterns were observed with respect to changes in MnPF and MdPF (see Table 3). Declining patterns were observed across exercise conditions, with maximum changes of up to nearly −0.9 Hz/min. Of the three factors manipulated, only $CPDF_{90}$ had significant effects on R_{MnPF} and R_{MdPF} ($P < 0.05$).

Table 3
Mean (SD) rates of change in EMG-based measures

Cond.	R_{RMS} (% max/min)	R_{MnPF} (Hz/min)	R_{MdPF} (Hz/min)
HHS	0.167 (1.25)	−0.687 (1.25)	−0.874 (1.42)
HHF	−0.125 (1.09)	−0.425 (0.61)	−0.522 (0.79)
LHS	−0.002 (0.00)	−0.139 (0.44)	−0.108 (0.34)
LHF	−0.154 (0.00)	−0.133 (0.29)	−0.147 (0.27)
HLS	0.098 (0.00)	−0.016 (0.18)	−0.033 (0.23)
HLF	0.189 (0.01)	−0.167 (0.16)	−0.133 (0.10)
LLS	0.011 (0.00)	−0.080 (0.14)	−0.073 (0.14)
LLF	−0.049 (0.00)	−0.001 (0.17)	−0.041 (0.09)

R_{RMS}: rates of RMS change.
$R_{MnPF/MdPF}$: rates of MnPF/MdPF change.

3.3. CPDEs and EVAs

Initial $CPDE_{10}$ was affected only by differences in $CPDF_{10}$ (Table 4); similarly, initial $CPDE_{90}$ was affected only by differences in $CPDF_{90}$. The 50th percentile ($CPDE_{50}$) was the only measure affected by all three independent variables. The effects of external workload and rates of muscle contraction were significant only for EVA2 (L < 10% MVC; 1 s ⩽ *D* ⩽ 3 s) and EVA6 (10% MVC ⩽ L ⩽ 30% MVC; *D* > 3 s), while the remaining initial EVAs were only affected by one or two independent variables (Table 4).

No general patterns of CPDE rates of change (R_{CPDE}) due to workload differences could be found. For example,

Table 4
Levels of significance for effects of exercise conditions on initial CPDE and EVA measures

CPDE and EVA measure	Exercise conditions		
	$CPDF_{10}$	$CPDF_{90}$	AV
$CPDE_{10}$	++	−	−
$CPDE_{50}$	++	++	++
$CPDE_{90}$	−	++	−
EVA1 (*L* < 10% MVC; *D* < 1 s)	−	++	+
EVA2 (*L* < 10% MVC; 1 s ⩽ *D* ⩽ 3 s)	++	+	++
EVA3 (*L* < 10% MVC; *D* > 3 s)	++	+	−
EVA4 (10% MVC ⩽ *L* ⩽ 30% MVC; *D* < 1 s)	−	++	−
EVA5 (10% MVC ⩽ *L* ⩽ 30% MVC; 1 s ⩽ *D* ⩽ 3 s)	−	−	++
EVA6 (10% MVC ⩽ *L* ⩽ 30% MVC; *D* > 3 s)	+	++	++
EVA7 (*L* > 30% MVC; *D* < 1 s)	−	++	−
EVA8 (*L* > 30% MVC; 1 s ⩽ *D* < 3 s)	−	++	−
EVA9 (*L* > 30% MVC; *D* > 3 s)	−	−	+

−, not significant; +, significant at $P < 0.05$; ++, significant at $P < 0.01$.

a comparison between conditions with the highest force levels differing only in AV (HHS vs. HHF) showed positive values for HHS condition, but positive as well as negative values for the other. Differences in force level ($CPDF_{10}$ and $CPDF_{90}$) did not produce changes in CPDE measures ($P > 0.45$), with the exception of a significant effect of $CPDF_{90}$ on $CPDE_{10}$ ($P < 0.05$) (Table 5). Changes in AV did not result in significantly different CPDEs ($P > 0.08$). A similar result was found with respect to the effects of external muscle loading on rates of EVA change. Of the 10 EVAs, R_{SumEVA} was the only measure that changed as a result of differences in $CPDF_{10}$ and $CPDF_{90}$ ($P < 0.01$). This EVA measure, however, was not influenced by changes in AV.

Table 5
Levels of significance for effects of exercise conditions on CPDE and EVA rates of change

CPDE and EVA measure	Exercise conditions		
	$CPDF_{10}$	$CPDF_{90}$	AV
R_{CPDE10}	–	+	–
R_{CPDE50}	–	–	–
R_{CPDE90}	–	–	–
R_{EVA1} ($L < 10\%$ MVC; $D < 1$ s)	–	–	–
R_{EVA2} ($L < 10\%$ MVC;$1 \text{ s} \leqslant D \leqslant 3$ s)	–	–	–
R_{EVA3} ($L < 10\%$ MVC; $D > 3$ s)	–	–	+
R_{EVA4} ($10\% \text{ MVC} \leqslant L \leqslant 30\%$ MVC; $D < 1$ s)	+	–	–
R_{EVA5} ($10\% \text{ MVC} \leqslant L \leqslant 30\%$ MVC; $1 \text{ s} \leqslant D \leqslant 3$ s)	–	–	+
R_{EVA6} ($10\% \text{ MVC} \leqslant L \leqslant 30\%$ MVC; $D > 3$ s)	–	–	–
R_{EVA7} ($L > 30\%$ MVC; $D < 1$ s)	–	–	–
R_{EVA8} ($L > 30\%$ MVC; $1 \text{ s} \leqslant D \leqslant 3$ s)	–	–	+
R_{EVA9} ($L > 30\%$ MVC; $D > 3$ s)	–	+	–
R_{SumEVA}	++	++	–

SumEVA: sum of absolute EVA.
–, not significant; +, significant at $P < 0.05$; ++, significant at $P < 0.01$.

3.4. Relationships between fatigue measures and initial CPDEs or EVAs

Several regression models (Table 6) were generated to estimate measures of muscle fatigue as a function of initial CPDEs or EVAs. All CPDE measures were included in the models, whereas up to four EVA measures were included as predictors. The coefficients of determination (adjusted R^2) were generally low, ranging from 0.10 to 0.45, and were generally higher for R_{MVC} and R_{RPD} than for other fatigue measures.

3.5. Sensitivity of measures to exercise conditions

A broad range of effect sizes (ω^2) was observed across the different indicators of muscle fatigue (Table 7), with moderate to large sensitivity ($\omega^2 > 0.30$) typically shown for non-EMG-based measures. Relatively low sensitivity was found for EMG RMS and most CPDE measures. Across EVA measures, the effect sizes were generally low, though R_{SumEVA} seemed to have relatively high sensitivity to changes in $CPDF_{10}$ and $CPDF_{90}$ ($\omega^2 > 0.50$).

4. Discussion

This study aimed to describe changes in muscle activation during isometric shoulder abduction with varying force. Such efforts were utilized as a logical next step in

Table 6
Fatigue measures as a function of initial CPDEs and EVAs (parameters and performance of multiple regression models)

Measure	Intercept	$CPDE_{10}$	$CPDE_{50}$	$CPDE_{90}$	Adj. R^2	SE
ET	67.51	−1.87	0.66	−0.83	0.18	6.58
R_{MVC}	−0.430	0.181	−0.187	0.097	0.31	0.30
R_{RPD}	−0.089	0.045	−0.030	0.020	0.25	0.10
R_{RMS}	0.0053	−0.0001	0.0004	0.0004	0.10	0.002
R_{MnPF}	0.204	−0.059	0.048	−0.024	0.15	0.13
R_{MdPF}	0.307	−0.081	0.078	−0.039	0.24	0.14

Measure	Intercept	EVA2	EVA3	EVA4	EVA5	EVA6	EVA7	EVA8	EVA9	Adj. R^2	SE
ET	50.96						−203.94			0.17	6.62
R_{MVC}	2.53		−2.88	−11.13		−1.33		8.03		0.45	0.27
R_{RPD}	0.35	−1.07						2.20	0.53	0.30	0.10
R_{RMS}	0.00						0.05	−0.05	−0.02	0.21	0.00
R_{MnPF}	−0.43		0.90	1.96				−2.43		0.21	0.13
R_{MdPF}	−0.21	1.46		2.38	−1.34			−3.30		0.33	0.14

ET = Endurance time (min)
R_{MVC} = Rates of MVC change (%/min).
R_{RPD} = Rates of RPD change (unit/min).
R_{RMS} = Rates of RMS change (% MVC/min).
$R_{MnPF/MdPF}$ = Rates of MnPF/MdPF change (Hz/min).
$CPDE_{10/50/90}$ = 10th/50th/90th percentile of CPDE (% MVC).
EVA1–9 = Cumulative duration of EVA1–9 relative to total recording period (%).

Table 7
Sensitivity of dependent measures (ω^2)

Measure	Exercise conditions		
	$CPDF_{10}$	$CPDF_{90}$	AV
ET	0.82	0.85	0.42
R_{MVC}	0.72	0.64	0.11
R_{RPD}	0.88	0.92	0.30
R_{RMS}	*	0.07	0.08
R_{MnPF}	0.55	0.40	*
R_{MdPF}	0.57	0.41	*
R_{CPDE10}	*	0.30	0.16
R_{CPDE50}	*	*	0.10
R_{CPDE90}	*	*	0.21
R_{EVA1}	*	*	0.03
R_{EVA2}	*	*	*
R_{EVA3}	0.09	*	0.31
R_{EVA4}	0.31	*	0.18
R_{EVA5}	*	*	0.23
R_{EVA6}	*	*	*
R_{EVA7}	*	*	0.04
R_{EVA8}	*	0.14	0.27
R_{EVA9}	*	0.25	*
R_{SumEVA}	0.51	0.70	*

All CPDEs and EVAs represent rates of change.
* Negative ω^2 due to F-values <1.

understanding short-term responses beyond pure static contractions. In addition to conventional EMG-based indicators, more recent EMG signal processing methods were used, to determine their ability in describing different patterns of external load and as indicators of fatigue development.

A broad range of local fatigue was induced using the current protocols, and was manifested by changes in several commonly used measures. Endurance times varied markedly, ranging from less than 20 min for the most fatiguing conditions to 60 min for the lightest conditions. As a 'gold standard' (Gerdle et al., 2000), substantial declines (14–26%) in muscle strength were present in all conditions, equaling rates of −0.3%/min to more than −2%/min. Data on perceived local discomfort also suggested different rates of fatigue, varying from roughly 0.1 to 0.4 unit/min.

While muscle fatigue did occur, and developed at substantially different rates, such a phenomenon was not indicated by concomitant changes in EMG amplitude. In contrast, this discriminating characteristic was demonstrated by using spectral measures. Employing EMG based measures for evaluating dynamic activities has been criticized on methodological grounds (Duchene and Goubel, 1993; Roy et al., 1998), though several studies employing such a method have suggested its feasibility (e.g., Masuda et al., 2001; Morlock et al., 1997; Nussbaum, 2001; Potvin and Bent, 1997). This study, however, showed that the choice of fatigue assessment method based on EMG mean or median power frequency is worth considering, even for muscle loadings that are "quasi" static.

Comparisons among fatigue measures noted above showed greater sensitivity for endurance time, strength, and discomfort rating, relative to EMG amplitude and spectral indicators. The former measures were highly dependent on differences in distributions of external load and, to a lesser extent, rates of muscle contraction. Other evidence has shown a good agreement between subjective and objective measures of fatigue (e.g., Dedering et al., 1999; Grant et al., 1994; Jørgensen et al., 1988), though a contrasting finding with respect to these two types of fatigue indicator has also been reported (e.g., Dedering et al., 1999). When compared to EMG-derived measures, higher sensitivity for subjective ratings can probably be attributed to the fact that an individual's perception during muscular exertions is based on more complete information facilitated by the peripheral components, the central nervous system, and the central cardiovascular and respiratory functions (Grant et al., 1994). Reports in the literature (e.g., Åhsberg and Gamberale, 1998; Dieën and Heijblom, 1996; Vøllestad, 1997) also seem to suggest the adequacy of muscle strength and subjective ratings, particularly with respect to their sensitivity to differences in task conditions.

This investigation also evaluated CPDE and EVA as data reduction methods, with an expectation that these methods could be utilized to describe external exposures (Hagberg, 1979; Mathiassen and Winkel, 1991) and as manifestations of local fatigue development (Hagberg, 1979; Hagberg and Jonsson, 1975; Nakata et al., 1992). The assumption was that acute responses were (surrogate) indicators of chronic musculoskeletal symptoms and problems (Nussbaum et al., 2001). Adequate exposure metrics, therefore, should be capable of indicating the degree of short and (potentially) long-term effects associated with the exposure. To this end, however, the two methods did not seem to fulfill the above purposes. The potential of using initial CPDEs and EVAs to characterize muscle loading was limited, and using changes in CPDE and EVA-based measures to assess the development of local fatigue is probably of little value, drawbacks that have been noted in earlier studies (Christensen, 1986; Hagberg and Sundelin, 1986; Linderhed, 1993; Mathiassen and Winkel, 1991; Winkel and Bendix, 1984).

It should be noted that, specific to EVA, results of this study indicated a general observation, in which positive rates of change for some EVAs were simultaneously (and somewhat consistently) accompanied by negative values for others. The overall (absolute) magnitude of these changes (SumEVA) was closely associated with differences in external workload, with sensitivity (ω^2) equal to or greater than 0.4. This finding suggests that variability of myoelectrical activities is probably a more important estimator of fatigue, as opposed to changes in the magnitude of EMG amplitudes. Variability of muscular activities may provide important physiological and etiological information (Mathiassen and Winkel, 1991), and has been assumed to lead to reduced risk of musculoskeletal disorders (Möller et al., 2004).

Another issue that is worth considering is the design of EVA matrix, since it could affect study results. The matrix selected here was based on "assigned" loading parameters. Though different classification could have been used (e.g., Mathiassen and Winkel, 1991), there was currently no well-established classification scheme available. A more recent study (Dan et al., 2003) combined EVA cells into "more manageable" clusters, an approach that had the potential to discriminate workers with different muscle loading characteristics. No firm criteria, however, were chosen as a basis for selecting EVA cell boundaries. Another slightly different approach in building an EVA matrix has been based on the associations between EVA-based measures and the presence of musculoskeletal symptoms/problems. A few investigations (e.g., Hägg and Åström, 1997; Jensen et al., 1993; Merletti and Lo Conte, 1997; Veiersted et al., 1993) have addressed this relationship by investigating certain aspects of an EVA matrix, including periods of zero activity (EMG gaps) and very short EMG durations (EMG jerks). These studies typically demonstrated the ability of the measures to discriminate exposure patterns between occupational tasks, or even differentiate groups of healthy workers vs. those with disorders. The investigators further argued that differences in EMG gaps or jerks may provide important etiological information, but agreed that the actual effects (or if the measures can indeed predict future problems) remain unknown. These and findings from the present work point toward the potential value of the EVA, though much future work is still needed.

A critical issue worth discussing pertains to the choice of muscle, particularly since it is not known if activity of the middle deltoid muscle is representative of typical shoulder muscle functions. The shoulder girdle is a complex structure (Kadefors et al., 1976). Movements and control of this shoulder joint are accomplished via the work of several muscles, including the infraspinatus, trapezius, deltoid, and supraspinatus, with the latter two muscles playing an important role during shoulder abductions. It is thus important to provide the arguments for (and against) obtaining surface EMG data from only a single muscle.

With regard to the present work, a major advantage of measuring muscle activities from the middle deltoid is ease of measurement. Recall that participants in this study adopted a supine posture with the intention of eliminating the effect of arm weight. As a result, obtaining EMG from the dorsal muscles would have been impaired by skin contact with the supporting device. Another benefit gained included simple interpretations of EMG data. Difficulty may arise, for example, when two different muscles are investigated, but yielded conflicting results. There is currently no accepted method in deriving conclusions (i.e. relative fatigue rates induced by task parameters) if more than one muscle is involved. It should be noted that several common indicators of fatigue examined in this study (e.g., changes in strength or discomfort rating) were not influenced by the choice of muscle, though these measures were considered secondary.

A likely drawback of investigating only the middle deltoid muscle is the assumption of consistent recruitment (load sharing) between multiple muscles and the associated changes in EMG-based measures. Note, though, that such load sharing is often assumed in biomechanical models. This drawback, however, presented an opportunity if it was found that a single measure from a major agonist gives results comparable to other local fatigue measures (e.g., endurance time or strength changes). Comparable results would justify the use of a single muscle (representing function of the synergistic group), while conflicting results would emphasize a need for EMG recordings obtained from multiple muscles. To a certain extent, this study indicated correspondence between the two data processing methods (particularly the EVA) and the common measures of fatigue. This finding suggests the potential of using the middle deltoid for rough assessments of shoulder abduction efforts, and perhaps the use of a major agonist in general.

5. Conclusions

This investigation aimed to characterize the development of local fatigue during static work with varying external forces. A primary finding of the study was that common indicators (endurance, reductions in muscle strength, and change in perceived discomfort) could be generally employed for fatigue assessment purposes, for efforts that are cyclic and repetitive. In contrast to EMG RMS, spectral measures (MnPF and MdPF) were associated with fatigue development. For the cyclic, repetitive contractions examined, use of CPDE to describe task differences and muscle fatigue was of limited value. The potential of employing EVA for these purposes was also marginal. In short, the use of these two more recent EMG signal processing methods to characterize external exposure is still worth considering, but their use as a basis for indicating fatigue development is questionable. It should be noted, however, that an EVA-derived measure (R_{SumEVA}) did show some relationships with local fatigue development. Further research on this is warranted, particularly in terms of how EVA 'cells' should be grouped.

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