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ORIGINAL ARTICLE

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Static and dynamic myoelectric measures of shoulder muscle fatigue during intermittent dynamic exertions of low to moderate intensity

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Abstract Despite extensive research on muscular fatigue during prolonged static efforts, there have been relatively few studies of more complex tasks (dynamic and intermittent). A laboratory study of overhead work tasks was conducted to investigate whether electromyographic (EMG) measures can potentially serve as indicators of fatigue, particularly for ergonomic tasks analysis. Sixteen participants performed the tasks until they either developed substantial discomfort or reached a 3-h limit. EMG signals were obtained at intervals throughout the experiment from four shoulder muscles, both statically (during fixed-level test contractions) and dynamically (during task performance). Both EMG root mean square (RMS) amplitude and spectral content (mean and median power frequencies) were examined and compared in terms of their variability and sensitivity. In addition, a new fatigue index was developed to allow for the estimation of substantial fatigue onset. Variability was found to differ significantly between muscles and EMG measures, and was generally lowest for mean power frequencies obtained during static test contractions. Sensitivity was typically greatest for RMS versus spectral measures, and slightly higher for median than mean power frequencies. The results suggest that fatigue during dynamic tasks, while a complex phenomenon, can be monitored and quantified using EMG.

Keywords Electromyography · Shoulder · Localized muscle fatigue · Intermittent dynamic work

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Introduction

With ongoing efforts to reduce the physical loads incurred during occupational tasks, modern work is increasingly characterized by low to moderate efforts of highly stereotyped motions. Epidemiological studies have shown that repetitive tasks are strongly associated with a variety of soft tissue disorders (e.g., National Institute for Occupational Safety and Health 1997; National Research Council 1999), and sufficient evidence has accumulated for some to conclude a causal role of repetition in such disorders. Additional studies have highlighted the continuing prevalence of these disorders, which account for a large fraction of all workplace illnesses (Bureau of Labor Statistics 1998). Considerable efforts have been devoted toward amelioration, and numerous methods have been developed for quantifying exposures and estimating injury risk (Hagberg 1992; Winkel and Westgaard 1992). At present, however, there appears to be no consensus, with respect to work-related musculoskeletal disorders, as to the most accurate, reliable, and valid exposure and risk measures.

These disorders are likely to result from several underlying causal mechanisms, yet the nature of these mechanisms, their possible interactions, and any dose-response relationships are as yet not clear. This has, in part, resulted in the development of a wide range of assessment methodologies and tools. Localized muscle fatigue has received particular attention, and several authors have suggested that it has a contributory role in chronic muscle pain (Herberts and Kadefors 1976; Armstrong et al. 1993), or implicitly assumed this role as motivation for their investigations. Existing evidence supports an important role of local muscle fatigue as at least a surrogate indicator of risk, and as an exposure metric incorporating a range of underlying biomechanical (e.g., moment arms, external loads) and physiological (e.g., energy depletion, metabolite accumulation) influences.

Electromyographic (EMG)-based measures are commonly used in fatigue assessment. Extensive evidence has accumulated to show that the signal power increases and the spectrum density compresses and shifts toward lower frequencies with increasing fatigue. While demonstrated for a range of muscle groups with varying fiber-type compositions, the majority of such evidence is limited to prolonged isometric exertions. In contrast, there have been relatively fewer investigations of more complex tasks, such as those that are intermittent and/or non-isotonic or non-isometric. It is these more complex tasks, however, that are of particular interest in an occupational setting, where prolonged isometric efforts are rare.

Cautions have been raised on the use of EMG, particularly spectral measures, during these more complex exertions. Potential confounding effects related to changing muscle force (Shankar et al. 1989; Gerdle et al. 1991), length (Gerdle et al. 1988b; Rosenburg and Seidel 1989), and velocity (Gerdle et al. 1988c) have been demonstrated. Such results have not always been consistent (e.g., Gerdle et al. 1988a; Öberg et al. 1990; Ament et al. 1996), and may be further influenced by exercise type and intensity. Theoretical considerations related to the nonstationarity of the signal also argue against the validity of EMG spectral transforms obtained under dynamic conditions (Duchêne and Goubel 1993).

EMG-based fatigue measures have produced contradictory results when actually used in intermittent and/or dynamic tasks. Several authors have shown that the same power and spectral changes found in isometric tasks also appear under more complex conditions (Sundelin and Hagberg 1992; Ament et al. 1993; Christensen et al. 1995; Potvin and Bent 1997). These same studies and others (e.g., Nagata et al. 1990; Gamet et al. 1993), however, have shown substantial variability between subjects performing the same tasks, and myoelectric changes in directions opposite to expectations, particularly during low loading conditions. Such discrepancies may have resulted in part from some of the potential confounding factors noted earlier, or from additional confounding effects related to muscle temperature (Duchêne and Goubel 1993), differences in muscle fiber type composition and distribution (Kupa et al. 1995), or changes in underlying motor unit recruitment patterns. It is difficult to measure and control for these confounding influences, and thus difficult to compare directly results across studies. Local muscle fatigue and associated myoelectric changes are apparently not perfectly correlated.

During isometric exercise, clear relationships have been demonstrated between rates of frequency decrease and endurance, and by implication the rates of fatigue development (Hagberg 1981a; Dieën et al. 1993; Mannion and Dolan 1994). Inconsistent results, however, have been reported during dynamic exercise (e.g., Komi and Tesch 1979; Hagberg 1981a; Dieën et al. 1996). Although some have concluded that fatigue (at least as measured by spectral shifts) is not appropriate for vocational risk assessment (Westgaard 1988), EMG-based assessment of fatigue remains attractive, both in experimental and occupational settings, as one of the most applicable and objective approaches available.

Occupational applications of experimental fatigue studies have been somewhat limited. From an applied standpoint this is not surprising; extensive research has demonstrated fatigue-related changes, yet little of this work leads to the type of guidance required in ergonomic applications. Endurance data, such as the effort-endurance relationships presented by Rohmert (1960, 1973) are used commonly for this purpose, yet few would argue that failure points should serve as the basis for design (Mathiassen and Winkel 1992). Fatigue is an ongoing and continuous process (De Luca 1984) with no distinct onset time, yet there is a practical need to determine when fatigue (and perhaps injury risk) rises above a baseline level. Several fatigue indices derived from myoelectric measures have been proposed to address this goal (Merletti et al. 1991). Indices with inherent statistical rigor include tests of changes in parameters such as the action potential conduction velocity or mean power frequency versus time (e.g., Lindström et al. 1977). Others have used iterative tests on sequences of such values (e.g., Hagberg 1981b; Hansson et al. 1992; Nieminen et al. 1995; Hanon et al. 1998), changes in values outside of the normal range of variability (Suurküla and Hägg 1987; Öberg et al. 1991; Gamet et al. 1993), or joint analysis of EMG power and spectral changes (Sundelin 1993; Luttmann et al. 1996). Given these diverse approaches, there appears to be no accepted method at present that will satisfy the vocational need for identifying fatigue and risk onset.

The present study involved a laboratory simulation of industrial assembly work, and required specifically intermittent dynamic exertions. The simulation was intended to reflect the large number of jobs that involve short cycle times, low mean loads, and large duty cycles (Mathiassen and Winkel 1991). Several EMG-based measures of fatigue were obtained during both static test contractions and during dynamic task performance, and their relative variability was compared. Interrelationships among EMG-based measures were determined and their sensitivity of these measures was examined, with a goal of evaluating their potential to detect fatigue in different occupational tasks. In addition, a new fatigue index was derived from consideration of intra-subject variability and parameter rates-of-change. This new index was intended to facilitate the development of fatigue-based work design guidelines, and an initial evaluation of its sensitivity and specificity was performed.

Methods

Participants

Participants (eight female, eight male) completed an informed consent procedure approved by the Virginia Tech Institutional Review Board and were compensated for their time. All were righthanded, aged 18–22 years, and reported engaging in moderate levels of physical activity. None reported any history of musculoskeletal injuries in the prior year. An initial training session was held to provide familiarization with the experimental equipment and procedures, including performance of maximum voluntary exertions. Eight subsequent experimental sessions were conducted. A minimum of 48 h of recovery time was provided after the training session and between all experimental sessions to minimize residual fatigue.

Experimental task

A manufacturing task requiring overhead activities was simulated in a laboratory. Participants performed tapping motions between two targets placed above shoulder level, using a small lightweight wand to which a 3-cm-diameter sphere was attached at the end. The targets were arranged parallel to the floor and separated at average shoulder span (50 cm). A fixed rate of 1 tap each 0.75 s was maintained using computer-generated tones. One-minute cycles were performed, with each cycle consisting of a fixed duration of work and rest. Myoelectric recordings were made intermittently during the tapping task as well as during test contractions that occurred in the resting portions of the cycles. Blocks of ten cycles were performed either until participants indicated that they could no longer continue because of substantial discomfort or when 18 blocks were completed (3 h). The time of termination was recorded as the endurance limit (t_{end}).

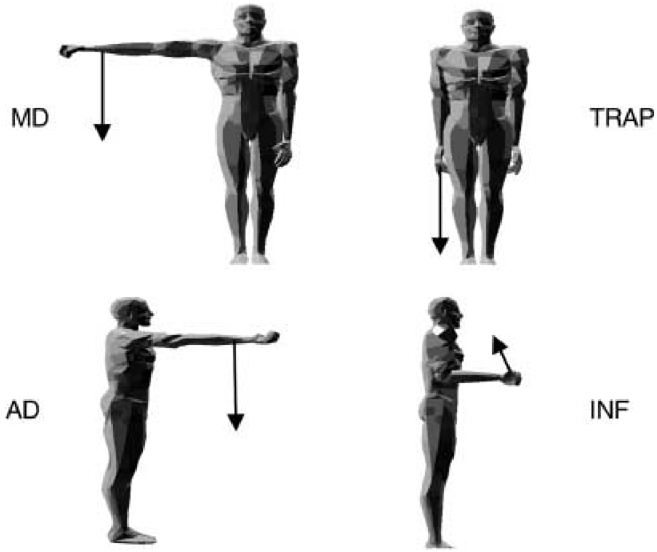

Fig. 1 Postures used during static test contractions (external rotation of the shoulder was used to test the infraspinatus muscle). (*MD* Middle deltoid muscle, *AD* anterior deltoid muscle, *INF* infraspinatus muscle, *TRAP* descending trapezius muscle)

Task variables

In each of eight experimental sessions, three task variables were held at one of two levels. The specific variables were target height, duty cycle, and hand orientation, and were selected to be representative of a range of common overhead industrial tasks. Each participant completed each of the eight factorial combinations, with a double Latin Square design used to minimize order effects (one Square for each gender). Target height was set at either 50% or 75% of an individual's overhead reach, with the proportion determined using the distance from the shoulder center-of-rotation to the maximum overhead grip height. The two duty cycles involved work/rest combinations of 20/40 and 40/20 s. Participants performed the tapping task with their palms either facing the targets (pronated) or facing their body (supinated).

Myoelectric measures

EMGs were obtained from four shoulder muscles: the middle deltoid (MD), descending trapezius (TRAP), anterior deltoid (AD), and infraspinatus (INF). These muscles were selected because of their accessibility to surface measurement, prior evidence of recruitment and fatigability during overhead work (Sigholm et al. 1984; Habes et al. 1985; Christensen 1986; Gerdle et al. 1988b), and the ability to functionally isolate them during the test contractions described below.

Pairs of disposable surface electrodes were adhered to the skin over the four muscles as follows: MD – midway between the acromion and the deltoid insertion; TRAP – 2 cm lateral to the midpoint of a line connecting the C7 spinous process and the acromion; AD – midway between the lateral third of the clavicle and the deltoid insertion; INF – 3.5 cm medial to the border of scapula and 3 cm below the spine of scapula. These locations were adjusted as necessary to place the electrodes over the belly of the respective muscles and parallel to the principle fiber directions. The inter-electrode resistance was maintained at less than 10 kΩ, and the reproducibility of locations between sessions was ensured both by marking the skin with a waterproof pen and by recording distances from prominent landmarks.

Maximum voluntary exertions (MVEs) were conducted prior to the tapping task in each experimental session. All MVEs were performed in consistent postures (see Fig. 1), with postural variations minimized using a special fixture built for this purpose. Participants either pulled on a handle (TRAP, INF) or against a strap placed on the lower forearm (MD, AD). All MVEs were conducted over $\cong$5 s using a ramp-up, hold, and ramp-down effort. Forces were measured using load cells, sampled at 100 Hz, and low-pass filtered at 10 Hz. Peak force values were recorded and used to estimate peak joint torques for the respective muscles using a three-dimensional kinematic model scaled to each participant's anthropometry.

Two separate sets of EMG data were obtained during the experimental sessions. Dynamic EMG data were recorded while participants performed the tapping tasks, for 10 s at the end of the working portion of every fifth cycle (every 5 min). Both raw and root mean square (RMS) EMG data were recorded. Static EMG data (raw and RMS) were obtained during fixed-level test contractions conducted during the resting portions of the tapping cycles. One muscle was tested during a given rest period in a consistent order, and no test contractions were performed in the remaining fifth cycle. Thus, static exertions were performed for each muscle every 5 min. The contraction level was set at 30% of maximum, using weight-filled buckets that generated external shoulder torques of 30% of a participant's maximum for each muscle. A level of 30% was selected based on pilot work showing that this was roughly the highest levels of muscle activity, and which also maximized the tradeoffs between lower EMG magnitude and signal-to-noise ratios at lower levels and inducement of fatigue at higher levels. Fixed levels were used to avoid the possible confounding effects of contraction level on EMG amplitude and spectral content, and to allow for inter-individual comparisons. The test contractions also resulted in muscle activity that was comparable to what was observed during the experimental tasks, thus helping to ensure that the same motor units were used in both (Søgaard 1995). Fixed postures were used to avoid the potential confounding effects of changing muscle lengths and shoulder loads. Test contractions were performed in postures designed to functionally isolate the muscles (Fig. 1), although no fixture was used. The weighted bucket was held for 3 s, during which a 2-s EMG recording was made, and the entire test contraction procedure was completed in approximately 5 s.

Raw myoelectric signals were preamplified ($\times$100) near the electrode sites, then hardware amplified and band-pass filtered (30–

500 Hz). Both raw and RMS signals were A/D converted and sampled, with the latter obtained using a 110-ms time constant. RMS data were sampled at 100 Hz, and subsequently low-pass filtered (second order Butterworth; 10 Hz cutoff). Raw EMG data were obtained during the static test contractions at 1024 Hz. Hardware limitations at the time of the study, however, limited the sample rates of raw dynamic EMGs to 512 Hz.

Data reduction

RMS EMG data were averaged over the respective sampling windows (2 s during static test contractions and 10 s during dynamic test contractions), and normalized to a percentage of the maximum RMS values measured during initial MVE trials. The 2-s samples of raw EMG obtained during static test contractions were processed (Hamming window and fast Fourier transform) and the power spectrum used to determine median and mean power frequencies (MdPF and MnPF, respectively). The 10-s samples of raw dynamic EMG were first broken into ten 1-s samples, then similarly processed to obtain the MdPF and MnPF.

All EMG data were analyzed in terms of changes in time with respect to initial values. The use of changes, rather than magnitudes, was necessary to facilitate comparisons across conditions and participants. Linear models were fit to percentage changes versus time, since the data did not exhibit any consistent non-linearities. Student's *t*-tests were used to determine if the linear trends were significant.

As there is no accepted standard for deciding when, on the basis of EMG, fatigue has occurred or become significant, a new operational measure was employed. The first step was to determine variability (σ) in a time-series of EMG data, from a single participant, condition, and muscle. Variability was quantified as the residual error $\sqrt{\mathrm{MSE}}$ from linear regression and expressed as a proportion of an initial value, with the latter estimated as the intercept (b_0) of the regression line (Eq 1). Significant differences ($P<0.1$) in σ were found between participants and muscles, but not between experimental conditions. In all subsequent analyses and to facilitate comparisons between participants, values of σ specific to each participant were used, after averaging across conditions for each muscle.

$$\sigma = \frac{\sqrt{\mathrm{MSE}}}{b_0} \tag{1}$$

An operationalized measure of time-to-fatigue (TTF) was subsequently defined as the time at which the best-fit line (EMG-based measure versus time) had changed by more than half the variability ($\sigma/2$). TTFs were calculated from a simple function of the variability and the regression line slope (b_1) normalized to initial values (Fig. 2; Eq 2). These TTF values were assumed to indicate when the EMG-based manifestation of fatigue had become substantial. If TTF was predicted to be greater than 3 h (the maximum experimental duration), it was set instead to 3 h. TTF values were computed separately for the 4 muscles and the 6 types of EMG data (RMS, MdPF, and MnPF during static and dynamic tests), thus yielding 24 estimates of fatigue onset for each subject and condition.

$$\mathrm{TTF} = \frac{\left(\frac{\sigma}{2}\right)}{\left(\frac{b_1}{b_0}\right)} \tag{2}$$

Statistical analyses

Preliminary analyses of variance (ANOVA) were used to determine if any order effects were present due to serial presentation of the conditions. Subsequent ANOVAs were used to investigate differences in data variability (σ) and time-dependent changes (normalized slopes) between the various EMG-based measures. Several of the TTF measures were found to be non-normally distributed, probably due to the ceiling limit of 3 h. The TTF measures were thus evaluated using non-parametric tests to examine the effects of the three task variables and their two-way interactions (ANOVA by ranks). Effects in all statistical tests were considered significant when $P<0.1$. This P value was chosen in order to increase the likelihood of identifying interesting effects in spite of slightly higher probability of type 1 error compared with $P<0.05$. The results are presented as the mean (SD) unless indicated otherwise.

Results

Example data

An example of RMS and power frequency results obtained from the MD muscle during static test contractions performed every 5 min is illustrated in Fig. 2. In this trial, the expected increases in RMS and decreases in MdPF and MnPF values were observed, although these changes were not seen in all trials, as described below. A regression line fit to the normalized RMS time series yielded an intercept of 21.9% and a slope of 0.027%·min^{-1}. For this participant's MD muscle, the average variability (σ, Eq. 1) was 8.5% of initial values, which equates to 1.86% in the present sample, and a TTF (Eq. 2) of 34.3 min was obtained. An example of MnPF and MdPF results obtained from EMG data collected during task performance (dynamic) from the same participant and condition is illustrated in Fig. 3.

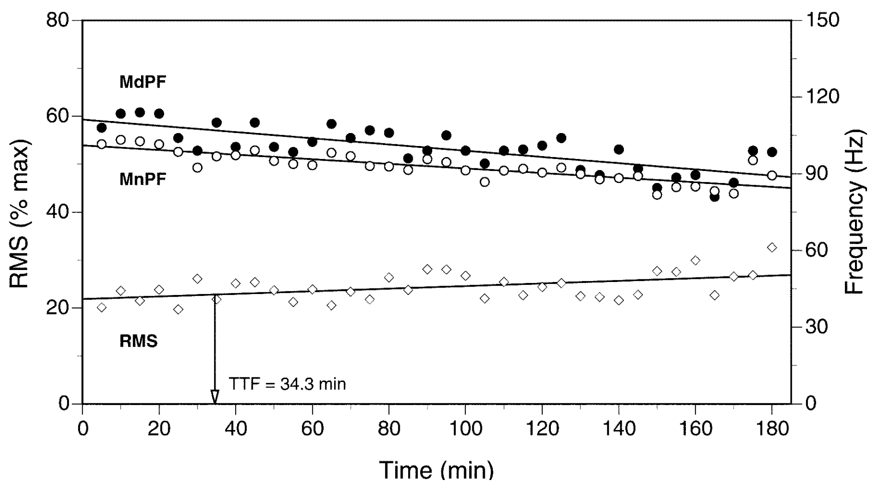

Fig. 2 Sample root mean square (*RMS*, *open diamonds*), median power frequency (*MdPF*, *filled circles*), and mean power frequency (*MnPF*, *open circles*) results obtained from the MD muscle during static test contractions (50% reach, hand supinated, 40/20 s work/rest duty cycle). Regression lines fit to all three electromyography (EMG)-based measures exhibited non-zero slopes ($P<0.01$). Procedures used to obtain time-to-fatigue (*TTF*) measures are illustrated for the RMS data

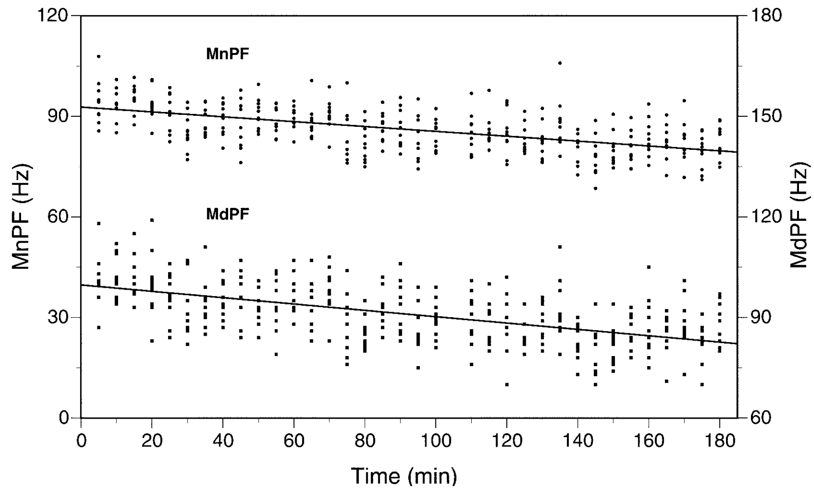

Fig. 3 Sample (dynamic) results obtained from the MD muscle during task performance (same subject and experimental condition as in Fig. 2). Regression lines fit to both the MdPF (right-hand ordinate) and MnPF (left-hand ordinate) measures exhibited non-zero slopes ($P < 0.001$)

Again, the expected decreases in power frequency values were observed (although not consistently in all trials).

Order effects and muscle activation

Order as a main effect was not significant for t_{end} ($P = 0.9$). Several muscles had significant order effects for TTF based on static EMG measures. In contrast, none of the TTF measures based on dynamic EMG showed significant order effects. No obvious trends with respect to order could be found across the different sets of TTF results, except that they were similar between values based on MnPF and MdPF.

Slightly higher levels of activity were seen in the AD and MD muscles versus the TRAP or INF muscles across all subjects and conditions. Mean levels of normalized RMS activity (%max) were 19.6 (9.4) and 19.1 (7.9) for the AD and MD, respectively compared to 16.9 (7.5) for the INF and 16.0 (6.9) for the TRAP. All pair-wise differences were significant ($P < 0.01$), except for AD versus MD and TRAP versus INF. Muscle activity was also dependent upon the task conditions. Duty cycle had significant effects ($P < 0.02$) on all muscles except the TRAP ($P = 0.2$), although this effect was relatively small, with activity levels 2–4% higher in the 40/20 condition. At the lower and higher tapping heights, average MD activity was 22% versus 16% respectively ($P < 0.0001$). No other significant task effects were associated with differences in mean activity greater than 2%.

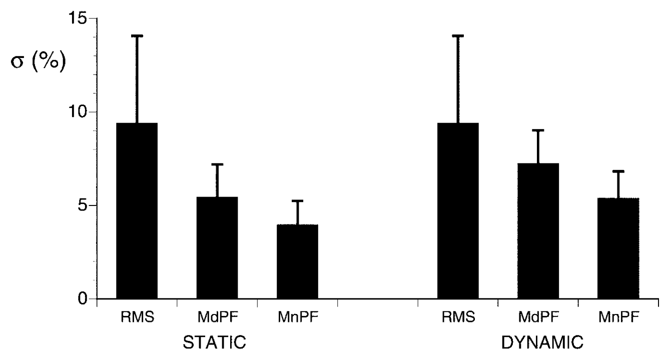

Fig. 4 Variability (σ: percentage of initial values; Eq. 1) of EMG-based measures obtained during static and dynamic tests. Error bars indicate one standard deviation

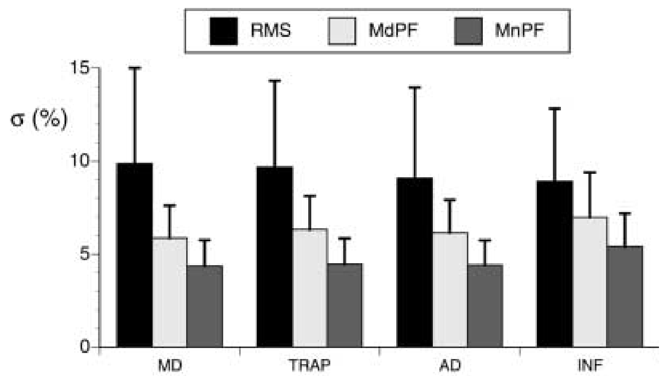

Fig. 5 Variability (percentage of initial values; Eq. 1) of EMG-based measures obtained for each of the four muscles tested (RMS *black bars*, MdPF *light gray bars*, MnPF *dark gray bars*). Error bars indicate one standard deviation

Variability

As a percentage of initial values (σ, Eq 1), the variability of the several EMG-based measures differed significantly between the type of test (static or dynamic; $P < 0.001$), type of EMG data (RMS, MdPF, MnPF; $P < 0.0001$), and muscle ($P = 0.004$). Variability was lower during static versus dynamic tests, with means of 6.3 (3.8)% and 7.3 (3.4)%, respectively. Variability was lowest for MnPF [4.7 (1.5)%], slightly higher for MdPF [6.3 (2.0)%], and highest for RMS measures [9.4 (4.7)%]. All pair-wise differences between the three types of EMG data were significant ($P < 0.05$). Despite the significant main effect of muscle, there were no significant pair-wise differences between muscles, and mean

Table 1 Regression slopes. Rates of change of normalized electromyogram (EMG) parameters (expressed as a % change/h) for each muscle, type of test (*Static* and *Dynamic*), and type of EMG measure. Mean values are taken across participants and experimental conditions. Distributions are also given to represent the percentages of slopes that were significantly positive (+), significantly negative (–), or nonsignificant (0). (*RMS* Root mean square, *MdPF* median power frequency, *MnPF* mean power frequency, *MD* middle deltoid muscle, *TRAP* descending trapezius muscle, *AD* anterior deltoid muscle, *INF* infraspinatus muscle)

		Static				Dynamic			
		MD	TRAP	AD	INF	MD	TRAP	AD	INF
RMS	Mean	6.01	9.22	8.23	7.47	10.99	12.89	6.55	10.08
	–/0/+	14/46/40	8/46/46	14/47/39	12/43/45	9/28/63	5/38/57	16/35/49	10/31/59
MdPF	Mean	0.37	–0.91	–0.09	–1.76	–1.00	–1.29	–1.96	–1.37
	–/0/+	34/49/17	41/48/11	33/51/16	34/50/16	50/25/25	61/23/16	58/31/11	49/31/20
MnPF	Mean	0.18	–0.80	–0.40	–1.35	–0.86	–1.51	–1.75	–1.03
	–/0/+	34/47/19	43/44/13	29/55/16	38/46/16	53/20/27	63/22/15	60/25/15	48/32/20

values ranged from 6.7 to 7.1%. A significant interaction between type of test and type of EMG was found ($P < 0.0001$), and exhibited as a slight difference in the relative effect of EMG type between static and dynamic tests (Fig. 4). The interaction between type of EMG and muscle was also significant ($P < 0.0001$), and was similarly a result of differences in the relative effect of EMG type between muscles (Fig. 5).

Time-dependent changes

Rates of change (normalized slopes) are compiled across participants and conditions in Table 1. The values tended to be substantially higher (as a percentage) for RMS versus power frequency measures. The percentage of the RMS measures that exhibited significant positive linear changes was slightly higher than power frequency changes that were negative (i.e., the "expected" directions of change). Rates of change were significantly ($P < 0.0001$) different between the three types of EMG data, being largest in magnitude for RMS measures with a mean of 8.73 (18.5)%$\cdot$h^{-1}. Normalized slopes were comparable for MdPF and MnPF measures, at –1.00 (6.1)%$\cdot$h^{-1} and –0.94 (4.9)%$\cdot$h^{-1}, respectively. Although the main effect of type of EMG was not significant

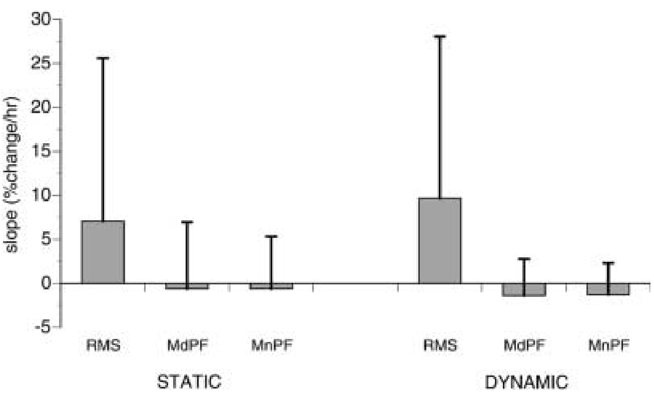

Fig. 6 Rates of change (percentage initial values) of EMG-based measures obtained during static and dynamic tests. Error bars indicate one standard deviation

($P = 0.39$), there was a significant ($P = 0.001$) interaction between type of test and type of EMG data (Fig. 6). There were no significant main or interactive effects of the different muscles ($P > 0.29$). The proportion of significant "expected" slopes was largest for RMS data, at approximately 57% and 43% for dynamic and static measures, respectively. Both spectral measures had comparable proportions of significant negative changes, at roughly 55% and 36% for dynamic and static measures, respectively.

Time-to-fatigue

Effects of the task variables on TTF values (Eq. 2) showed some level of consistency between the static and dynamic measures and three types of EMG measures (Table 2). With a single exception, duty cycle had a significant effect on TTF across all muscles, both types of tests, and all three EMG measures. Factor effect on the two power-frequency-based measures tended to be more consistent than either was with RMS-based TTF. No substantial differences between muscles were apparent in terms of the number of factor effects observed. TTF obtained from RMS measures resulted in more significant factor effects during static test contractions, in contrast to power-frequency-based TTF where more factor effects were found during dynamic tests.

Correlation matrices were created to examine the relationships between t_{end} and the several TTF measures. When comparing subject-specific values (16 subjects $\times$ 8 conditions = 128 data points), simple correlations (R^2) between t_{end} and TTF were generally low and in the order of 0.2. Results were similar for static (0.20) and dynamic (0.21) measures, but were smaller for RMS (0.17) than either MdPF (0.23) or MnPF (0.21) measures. Higher correlations were found for the MD (0.20), AD (0.23) and INF (0.22) than for the TRAP (0.17). Correlations were substantially higher (Table 3), however, when comparisons were made on subject-averaged data (eight conditions). These correlations were also slightly higher for dynamic and power-frequency-based measures, and for the INF versus the remaining muscles. Standard errors obtained from re-

Table 2 Factor effects on time-to-fatigue (*TTF*) estimates obtained from each muscle, type of test, and type of EMG measure. Factor effects represent the task variables as follows: A = arm height; H = hand orientation; D = duty cycle

Type of test	Factor	RMS				MdPF				MnPF			
		MD	TRAP	AD	INF	MD	TRAP	AD	INF	MD	TRAP	AD	INF
Static	A	**											
	H	*			***	**				**			
	D	***	***	*	***	***	***	***	***	***	***	***	***
	A×H												
	A×D		*							**			
	H×D		**	***					*		*	*	
Dynamic	A	***				***		***					
	H			*				*				*	*
	D		***	***	***	***	***	***	***	***	***	***	***
	A×H							**			**	*	
	A×D												
	H×D							**				***	

*Significant at $P < 0.1$; **significant at $P < 0.05$; ***significant at $P < 0.01$

Table 3 Correlations among TTF measures. Adjusted coefficients of determination (R^2) and standard errors (*SE*; minutes) resulting from comparison of endurance times (t_{end}) and TTF measures obtained for each muscle, type of test, and type of EMG measure. Values were obtained from linear regression of mean t_{end} and TTF values across participants

Variable		Static				Dynamic			
		MD	TRAP	AD	INF	MD	TRAP	AD	INF
RMS	R^2	0.35	0.49	0.14	0.64	0.00	0.79	0.27	0.64
	SE	29.1	25.9	33.5	21.6	36.1	16.7	30.9	21.6
MdPF	R^2	0.93	0.58	0.95	0.95	0.90	0.86	0.77	0.91
	SE	9.9	23.3	7.8	8.4	11.3	13.3	17.4	10.4
MnPF	R^2	0.79	0.52	0.94	0.92	0.93	0.88	0.72	0.81
	SE	16.5	24.9	8.7	9.9	9.5	12.8	19.1	15.6

gression of t_{end} and the TTF measures (Table 3) paralleled those obtained for R^2.

Discussion

The present work was part of a larger effort to derive design guidelines for a set of occupational tasks requiring prolonged intermittent efforts of the shoulder musculature. If it is assumed that fatigue is a valid (or surrogate) measure of injury risk, design based on endurance was considered disadvantageous. Rather, acknowledging that fatigue is a continuous and ongoing process, there was a need to evaluate measures to assess fatigue and to identify a point in this process at which risk could be considered to have risen above a baseline level. To support these goals, several EMG-based measures of fatigue were compared in parallel during differing task conditions. In addition, a new fatigue index was developed in order to provide the desired guidelines.

Several limitations were inherent in the described study. EMG measures obtained under non-isometric and non-isotonic conditions are potentially confounded by variability in muscle force, length, and velocity. There are additional, although incompletely understood, changes in motor unit recruitment in response to fatiguing contractions that are variable both within and between individuals. Many of these limitations related to muscle kinetics and kinematics, however, were likely to have been minimized by the cyclic nature of the task investigated. Specifically, the cyclic effects of these factors can be expected to parallel the stereotyped patterns of muscle activity used in the task. By sampling roughly six or seven such cycles, data derived from the dynamic EMG should be relatively independent of these confounds as a result of averaging, and more reflective of any underlying fatigue (Roy et al. 1998).

Static test contractions were performed at a level of 30% of maximum, which exceeded the typical levels observed during the tasks. Different motor units may therefore have been recruited during the test contractions. Given that motor unit recruitment order is typically stereotyped (Henneman et al. 1965), it is expected that the same motor units used (and fatiguing) during the task, and measured by the dynamic EMG, were also measured during the static EMG. Any additional motor units recruited during the higher-level test contractions would have been similar across samples in an experimental trial and across conditions. The expected effect would have been a "dilution" of fatigue-related changes, which may have contributed both to the relatively lower variability (σ) and lower sensitivity (slopes) of the static EMG-based measures.

Signal stationarity is generally argued to be a prerequisite for valid spectral analysis (Duchêne and Goubel 1993). Since the static test contractions were of fixed levels and short duration, EMG stationarity was ensured. In contrast, the myoelectric signals during dynamic tests were likely to have been non-stationary. The use of consistent measurement procedures, and examination of within-subject changes, minimized confounding effects of non-stationarity. Several other observations argue that non-stationarity was not problematic. Variability was only moderately higher in dynamic versus static EMG spectral measures, although the two sets of measures were obtained under very different conditions. In addition, the rates of change of dynamic measures were generally larger than those from static test contractions (Fig. 6), implying a higher sensitivity of the dynamic measures. Finally, the number of factor effects on static and dynamic TTF (Table 2), and the correspondence between both sets of TTF and endurance times (Table 3) were indistinguishable.

Some degree of recovery may have occurred before the static EMG measures were obtained. To facilitate rapid measurement, the four muscles were tested in a consistent sequence, with only one muscle tested at a time. Recovery of mechanical measures such as maximum force or twitch tension is relatively slow, with reports ranging from 15 min (Moussavi et al. 1992) to more than 24 h (Kroon and Naeije 1988). Myoelectric measures, in contrast, recover more quickly, in the order of 2–10 min (Mills 1982; Ament et al. 1993; Buttelli et al. 1999). Since the static test contractions were performed within ≅5 s following the task, only negligible levels of recovery are likely to have occurred. The use of consistent test procedures suggests further that recovery levels were consistent both within and between subjects. Since within-trial changes and rates of change were the focus of our analysis, recovery is not considered to have been a significant confounding factor.

Linear changes versus time were assumed for the EMG-based measures, although there is mixed evidence in the literature for such a relationship. During prolonged isometric exertions, amplitude and spectral myoelectric changes, as well as the action potential conduction velocity, are typically nonlinear and often best characterized as exponential (Lindström et al. 1977). Some investigators have identified two fatigue "phases" during such efforts, the first characterized by a rapid change and the second by a plateau (Tesch et al. 1983; Gerdle and Fugl-Meyer 1992). Less consistent results are reported during intermittent and dynamic efforts, although linear changes tend to be observed at low exertion levels (Krivickas et al. 1996; Masuda et al. 1999). In the present work, nonlinearity was not comprehensively tested, yet no evidence for this was found in empirical tests of a subset of data or from inspection of all the data. Linear fits to the data (c.f., Figs. 2, 3) greatly simplified the analysis and appeared to provide a good characterization of the time-dependent changes. It is possible, though, that the EMG measures would have reached a plateau if the experiments were continued to exhaustion, rather than terminating at 3 h or at moderate–high levels of perceived discomfort.

Subjects performed each task condition only once, and test-retest or day-to-day reliability was not addressed. Several reports have indicated a fair level of both types of reliability for time and frequency domain measures during isometric exertions for several muscle groups over a wide range of levels (Yang and Winter 1983; Roy et al. 1989; Dieën and Heijblom 1996; Dedering et al. 2000). The repeatability of phasic muscle activity (Kadaba et al. 1985) has also been demonstrated, although under more familiar motor demands (gait). Measures obtained during intermittent and/or dynamic exertions also appear repeatable, but to a lesser extent than under continuous isometric conditions (Linssen et al. 1993; Kankaanpää et al. 1997). Discrete values such as the initial MdPF are typically more reliable than time-dependent changes such as slopes under both static and dynamic conditions (Krivickas et al. 1996; Oliver et al. 1996; Peach et al. 1998).

Reliability was addressed in this study within subjects and within trials, by quantifying variability (expressed relative to initial values) about a best-fit linear temporal relationship. Consistent with existing reports, static test contractions yielded more reliable (less variable) myoelectric measures than dynamic conditions. Greater control over exertions in the former, slightly larger exertion magnitudes, and the possible differences in motor unit recruitment noted above, may explain the relative difference in reliability between the two types of measures. Reliability was also substantially better for spectral (MnPF and MdPF) than for RMS measures. Changes in muscle coordination between samples (perhaps an adaptation to fatigue), and the relatively lower sensitivity of spectral measures to changes in contraction levels probably account for these differences. The slightly lower variability in MnPF versus MdPF observed in the present study is consistent with the reported higher stability of the MnPF (Hary et al. 1982), and quite close in terms of relative differences to one earlier experiment (Dieën and Oude Vrielink 1996). Despite fundamental differences between isometric exertions, in which these earlier results were obtained, and the present intermittent dynamic tasks, the MnPF would appear to be the most reliable spectral indicator, although perhaps less sensitive.

The number of significant regression slopes (Table 1) can be considered an estimate of sensitivity, since each was obtained in parallel during each trial. Static and dynamic RMS measures exhibited both the largest percentage changes and a slightly larger proportion of changes (40–60%) that were significant in the expected direction. Rates of change of both spectral measures were an order of magnitude smaller, yet were still significant in 35–60% of trials. Substantially larger proportions of significant slopes were found with dynamic versus static measures, and both amplitudes and spectra appear to be equally sensitive indicators during the studied tasks. The amplitude measures, however, may have had some level of increased activity, reflecting additional recruitment as a non-fatigued synergist. As an example, Dieën et al. (1993) found that during isometric trunk extensions the latissimus dorsi muscle had both the least spectral change and the lowest initial amplitude, suggesting that the observed temporal increases in myoelectric amplitude of this muscle were compensatory changes to fatigue in other trunk extensors. This type of changing load sharing between multiple synergists may have occurred in the present study, but such patterns of amplitude and spectral changes were not observed with any consistency.

Results obtained for MnPF and MdPF sensitivity were similar, although MdPF slopes were roughly 5% larger in magnitude overall. This higher sensitivity, despite the lower reliability compared to MnPF, supports earlier suggestions (Dieën and Oude Vrielink 1996; Merletti and Roy 1996) that the MdPF is a slightly more responsive indicator of fatigue during an intermittent dynamic task. It is notable that the proportions of significant negative spectral slopes overall, and the proportion of tasks that were terminated before 3 h due to subjective discomfort, were nearly identical (45%). Although this is a high population-level correspondence, a significant slope and an early termination were not always observed in the same trials.

Significance of the different factor effects on the derived TTF measures (Table 2) can be viewed as an estimate of specificity. This interpretation assumes that a measure that can discriminate between different task levels is not only sensitive enough to detect fatigue-related changes, but also specific enough to yield distributions of TTF between conditions that are sufficiently distinct. The influence of duty cycle was large enough that all TTF measures (with one exception) could detect it. No clear pattern emerges from inspection of the numbers of other factor effects, which are nearly identical between EMG type and between the static and dynamic measures. Whether all measures have equivalent specificity cannot be concluded with certainty, given that earlier analysis showed that duty cycle also had the only consistent effects on endurance and the onset time for subjective discomfort (Nussbaum et al. 2001). The relatively high correspondence between subject-averaged TTF suggests that each of the four muscles exhibited comparable responses during the experimental tasks and that each of the different EMG measures used to derive TTF were undergoing parallel changes as a result of fatigue. The relatively low correspondence between subject-specific TTF, as opposed to subject-averaged values (Table 3), emphasizes the substantial variability between individuals performing very similar tasks.

This new index of fatigue (TTF) was created in order to facilitate the development of design guidelines. Numerous studies have presented indices of fatigue that quantify the level of fatigue, typically as percentage changes in some parameter (e.g., conduction velocity, EMG amplitude, MnPF). While providing a numerical measure to follow fatigue progression, and thereby allow for comparison between tasks, these approaches do not yield directly an indication of fatigue onset and thus give no indications of how long a given task should be performed. Other studies have determined when a measure, assumed be reflective of underlying fatigue-related intramuscular processes, had changed to an extent outside of normal variability. Examples include *t*-tests on the slopes of lines fit iteratively to increasing sequences of data (Lindström et al. 1977), or tests based on a priori specifications of baseline variability (Hagberg 1981b; Gamet et al. 1993). The present approach uses a combination of both of these latter methods, in which baseline variability is first determined for each subject and then logic similar to a one-sided *t*-test is used to obtain the time at which baseline variability is exceeded.

Benefits of the TTF approach include explicit inclusion of individual differences and provision of distinct onset times that can be compiled as design guidelines. Consistent variability (σ) was found here in all EMG-based measures within-subjects and across eight conditions, each performed at least 2 days apart. This consistency, in turn, formed a basis for the TTF method. A recent report (Pincevero et al. 2000), however, suggests that the variability in surface EMG measures is itself variable and relatively unreliable, although the focus was on maximal efforts. It is thus possible that the task dependency of TTFs found here may also reflect some level of task dependency of the parameter variabilities, although this level is likely to be small based on the present results. Ergonomic application of the TTF metric to differentiate between tasks is therefore supported, albeit requiring prior determination of intra-subject σ magnitudes.

Fatigue is clearly a multifaceted process. Myoelectric measures, while used extensively in isometric conditions, require ongoing study to determine how they can best be processed and employed under more complex (yet more realistic) exertions. The present results support the use of standard EMG analyses during intermittent dynamic contractions, and the use of a TTF metric for ergonomic task evaluation. Since metabolic processes appear to differ between sustained and intermittent exercise (Christmass et al. 1999), and since dynamic contractions may lead to a dissociation between some myoelectric phenomena (Masuda et al. 1999), further investigation is still warranted to explain and account for these findings.

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