Muscle Fatigue during Isometric and Dynamic Efforts in Shoulder Abduction and Torso Extension: Age Effects and Alternative Electromyographic Measures

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Dissertation submitted to the faculty of the
Virginia Polytechnic Institute and State University
in partial fulfillment of the requirement for the degree of

Doctor of Philosophy
In
Industrial and Systems Engineering

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October 28, 2005
Blacksburg, Virginia

Keywords: muscle fatigue, aging, electromyography, fractal

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Abstract

Aging has been associated with numerous changes in the neuromuscular system. Age effects on muscular performance, however, have been addressed only in limited contexts in earlier research. The present work was conducted primarily to investigate age-related effects on muscle capacity (fatigue and endurance) during isometric and dynamic efforts. This work was also motivated by current theories on muscle fatigue as a potential risk factor for musculoskeletal disorders and recent demographic projections indicating a substantial increase of older adults in the working population. Four main experiments were conducted to investigate development of muscle fatigue during isometric and intermittent efforts in shoulder abduction and torso extension at different contraction levels. Two age groups were involved (n=24 in each), representing the beginning and end of working life. Findings from this study demonstrated that the older group exhibited slower progressions of fatigue, though the age effect was more consistent for the shoulder than the torso muscles. This implied a muscle dependency of the influence of age on fatigue. Several interaction effects of age and effort level were also observed, suggesting that both task and individual factors should be considered simultaneously in job design.

The present investigation also sought to develop alternative electromyography (EMG)-based fatigue parameters for low-level isometric and dynamic contractions, two areas in which improvements are needed in the sensitivity and reliability of existing EMG indices. Several alternative EMG indices were introduced, derived from logarithmic transformation of EMG power spectra, fractal analysis, and parameter estimation based on a Poisson distribution. Potential utility of several of these alternative measures was demonstrated for assessment of muscle fatigue.

In the name of Allah, Most Gracious, Most Merciful.

Then which of the favours of your Lord will ye deny?

Koran, Chapter 55, Verse 13

Say: He is Allah, the One and Only.
Allah, the Eternal, Absolute.
He begetteth not, nor is He begotten.
And there is none like unto Him.

Koran, Chapter 112, Verse 1-4

Acknowledgments

First of all, I would like to express my sincere gratitude to my advisor, Dr. Nussbaum. It was a great chance to work with you and to have you as an advisor. I really appreciate your endless support and guidance. My appreciation also goes to all my committee members: Dr. Babski-Reeves, Dr. Lockhart, and Dr. Madigan. All of the feedback you gave made valuable contributions to this document. I have learned a lot from all of you.

I also want to thank Dadi, who served greatly as my mentor and introduced me to Hokies and this academia. Thanks to Miguel and Laura for helping me in polishing this document. To my other friends in the Industrial Ergonomics and Biomechanics Lab, thank you for friendship, help, and support. I would also like to thank all the faculty and staff in the ISE department, and to all my friends in the Virginia Tech HFES. To all my participants, I appreciate your involvement and cooperation in this study.

Finally, I am grateful to my parents, my brothers, my sisters, and my parents in-law for your constant prayers and support. To my wife, Imar, this academic career would not have been possible without your love, patience, support, sacrifice, and understanding. I dedicate this dissertation to you. May Allah reward all beautiful things you have given to me.

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List of Abbreviations and Symbols

α	: Fractal scaling
λ	: Poisson parameter
All	: Concentric + Eccentric
ANCOVA	: Analysis of covariance
ANOVA	: Analysis of variance
Conc	: Concentric EMG
CV	: Coefficient of variation
D	: Fractal dimension
DFA	: Detrended fluctuation analysis
dyn	: Dynamic EMG
Ecc	: Eccentric EMG
EMG	: Electromyography
FFT	: Fast Fourier transfrom
HF	: High frequency component
HFslp	: High frequency slope
ICC	: Intra-class correlation coefficient
LBP	: Low back pain
LF	: Low frequency component
LFBand	: Low frequency band (sum of power)
LFslp	: Low frequency slope
LLEs	: Low level efforts
LMF	: Localized muscle fatigue
MdPF	: Median power frequency
MFCV	: Muscle fiber conduction velocity
MnPF	: Mean power frequency
MVC	: Maximum voluntary contraction
nRMS	: Normalized RMS
O	: Older group
Peak	: Peak frequency
PSD	: Power spectral density
RMS	: Root mean square
RPD	: Rate of perceived discomfort
RS	: Remaining strength
SEM	: Standard error of measurement
st	: Static EMG
WMSDs	: Work-related musculoskeletal disorders
Y	: Younger group

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CHAPTER I
INTRODUCTION

1.1 RATIONALE

Reports have indicated that costs of injuries and work related musculoskeletal disorders (WMSDs) are still high. According to Bureau of Labor Statistics (BLS, 2004), there were over 520 thousands cases of WMSDs with days away from work, accounting for ~34% of total number of injuries and illnesses. Annual costs of WMSDs in 2001 were estimated to be $45.8 billion, or nearly $1 billion per week, an increase of 4% as compared to 1998 (Liberty Mutual Research News, 2003). Of these costs, workers' compensation claims associated with manual material handling accounted for approximately 36% (Dempsey and Hashemi, 1999).

Along with reports of costs, several risk factors associated with WMSDs have also been identified. Overexertion and repetitive motion were considered to contribute roughly 27% and 6% of the total WMSDs cost respectively (Liberty Mutual Research News, 2003). Based on epidemiological studies, physical workload, repetitiveness, and static effort were among the main risk factors mentioned for neck-upper limb disorders (Malchaire et al., 2001) and low back pain (Andersson, 1981; BLS, 2004; Waters, 2004). Since these biomechanical factors are closely related to fatigue, localized muscle fatigue (LMF) has been hypothesized as one possible risk factor (Armstrong et al., 1993), supported by several theories that describe WMSD-precipitating mechanisms that result from LMF (e.g., Forde et al., 2002).

In addition to biomechanical factors, aging has been viewed to impact the prevalence of WMSDs since older people are perceived to be more susceptible to injury. Several studies, particularly in physically demanding occupations, showed a higher rate of complaints by older workers (e.g., de Zwart, 1999). However, the precise role of age as a risk factor of WMSDs is still unclear. Despite this, since age may limit work capabilities and risk of injury could be reduced substantially by designing job demands within worker's capability, quantitative data pertaining to the physical work capacity of the older workers appear necessary. Note that this issue is at current interest due to a projected dramatic increase of older workers (around 46 percent) in the workplace from 2000 to 2010 (Horrigan, 2004).

1.2 RESEARCH ISSUES

An individual's physical work capacity can be determined by aerobic capacity and muscular capacity (de Zwart et al., 1995). More specifically for muscular capacity, muscle strength, fatigue, and recovery have been used as measures of work capacity. Most studies agree that there is a marked decline in muscle strength with advancing age after the fifth decade of life (e.g., Bäckman et al., 1995; Lindle et al., 1997). The trend appears to be similar for both isometric and dynamic strength, but varies depending on the muscle group examined (Bemben et al., 1991).

While a large number of studies have indicated age-related losses in muscular strength, only a few have investigated the effects of age on muscular endurance and fatigue (Allman and Rice, 2002). Two different approaches have been used to measure muscular endurance and fatigue. In early work, a fixed load was typically used, and muscular endurance was found decline with age (Larsson and Karlsson, 1978). Recent studies, however, have used a relative load (% maximum) in their experimental protocols, due to the differences in muscular strength among individuals and as a correction for the decrease in maximum strength with aging. When this method is used, conflicting results have been reported concerning age effects on muscle fatigability during isometric contractions (Larsson and Karlsson, 1978; Bilodeau et al., 2001) or dynamic contractions (Petrella et al., 2005; Lanza et al., 2004). Moreover, the age effect seems to be different across muscle groups (Klein et al., 2001). In short, further studies are needed that quantify age-related differences in fatigability during both isometric and dynamic contractions.

In assessing occurrence and development of muscle fatigue, several methods have been used such as sub-maximal test contraction, electromyography (EMG), mechanomyography, near-infrared spectroscopy, or subjective ratings of discomfort, each with its own limitations (Vøllestad, 1997; De Luca, 1997; Yoshitake et al., 2001). To date, surface EMG has been used most extensively in examining localized muscle fatigue (De Luca, 1984 and Merletti et al., 1992). Though precise relationships between EMG and physiological changes within the muscles are still being debated (Hägg, 1991; Kupa et al., 1995; Vøllestad, 1997), changes in the EMG signals are typically associated with fatigue development (De Luca, 1979; Hagg, 1992; Duchene and Goubel, 1993).

The utility of EMG-based measures, however, seems to be affected by the type of contractions performed. The most common existing EMG-based measures (amplitude and mean/median frequencies) have shown inconsistent changes for prolonged isometric exertions at low-level forces (Krogh-Lund, 1993; Hägg and Ojok, 1997; Oberg, 1994). Conflicting arguments about appropriate parameters for dynamic efforts have also been reported (Komi and Tesch, 1979; Hagberg, 1981; Gamet et al., 1993; Potvin, 1997; Roy et al., 1998; Masuda et al., 2001). Note that low force levels and dynamic efforts are considered here to be more relevant to occupational tasks. Therefore, it is of interest to obtain alternative measures that are reliable and sensitive for low levels of sustained contractions as well as for repetitive-dynamic efforts.

One possible reason for less effective use of existing EMG measures to indicate fatigue is that the measures seem to inadequately characterize EMG signals (Hägg and Ojok, 1997).

Recently, non-linear methods such as fractal analysis have become attractive in biomedical signal processing. Mandelbrot coined fractal from the Latin adjective *fractus* meaning irregular, fragment or fraction (Iannaccone and Khokha, 1995). The essential characteristic of fractals is 'self-similarity', meaning that the properties of the pieces are similar to those of the whole. Many biological structures can be regarded as natural fractals such as the vascular tree, the bronchial tree, and the renal glomerulus. Fractal method has been used to analyze heart rate variability in diagnosing the effects of aging and diseases (Iyengar et al., 1996; Turcott and Teich, 1996). Fractal analysis is also applied to electroencephalographic (EEG) data in characterize pathological condition of the human brain (Accardo et al., 1997). This method may provide similar useful applications for EMG, and may also produce better EMG-based fatigue parameters than the existing ones; further investigation is warranted to determine this.

Applications of EMG have resulted in a number of valuable interventions. EMG has been applied as a parameter for better design of hand tools and posture (Chaffin, 1973) and workstation design (Kofler et al., 2002). EMG has also been used to determine acceptance criteria of muscular loads for job design (Jonsson, 1982; Christensen, 1986) and to estimate time-to-fatigue (Nussbaum, 2001). In the clinical setting, EMG-fatigue analysis has been used in physical rehabilitation (Roy et al., 1995), monitoring muscular strain and fatigue on surgeons (Luttman, 1996), diagnosis of muscle impairment (Gorelick et al, 2003), and prediction of functional and endurance deficits (Maisetti et al, 2002; Van Dieën et al, 1998). More effective interventions can be expected if more reliable and sensitive EMG measures can be obtained.

1.3 RESEARCH OBJECTIVES

It can be concluded from the above discussion that at least two main issues in the area of aging and localized muscle fatigue still remain for investigation. The first relates to the effects of age in muscle endurance and fatigue, and that indicate a need for more quantitative data. This topic is argued to be important, due to the increasing population of older workers in the workplace. The second issue pertains to a need for finding improved EMG-based fatigue measures, in particular for isometric low-effort levels and repetitive dynamic tasks.

The two issues noted above were addressed in the comment work, which had the following main objectives:

a. To investigate the effects of age on muscle endurance and fatigue during isometric and dynamic exercises. In addition to endurance and fatigue, muscle recovery was also particularly addressed.
b. To investigate the adequacy of different processing methods in deriving EMG-based fatigue measures during low-level isometric contractions.
c. To investigate the adequacy of different processing methods in characterizing EMG signals for localized muscle fatigue assessment during intermittent dynamic contractions.

1.4 THESIS ORGANIZATION

This work is reported in five main sections, in addition to this introductory section. The first three parts (Chapter II-V) cover the first objective and describe age-related differences in muscle endurance, fatigue, and recovery from isometric and dynamic efforts. This investigation was focused on shoulder and low back muscles, which are functionally and morphologically different and typically active in industrial work. Chapter II describes the effect of age during isometric efforts, while Chapter III and IV focus on dynamic efforts using the shoulder and low back muscle, respectively. In the last two chapters, several alternative EMG-based fatigue measures for isometric low-effort levels and dynamic efforts are proposed and evaluated in Chapters V and VI, respectively. Summary and suggestions for future work in these areas are provided in Chapter VII.

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CHAPTER II

THE INFLUENCE OF AGE ON ISOMETRIC ENDURANCE AND FATIGUE IS MUSCLE DEPENDENT: A STUDY OF SHOULDER ABDUCTION AND TORSO EXTENSION

Abstract

The purpose of the present study was to examine differences in muscle capacity (endurance, fatigue, and recovery) between older (55-65 years) and younger (18-25 years) individuals. Two groups of 24 participants (gender balanced within each group) performed sustained isometric arm abductions and torso extensions to exhaustion at 30%, 50% and 70% of individual maximal voluntary contraction (MVC). Along with endurance time, manifestations of localized fatigue were determined based on changes in surface electromyographic (EMG) signals obtained from the shoulder (middle deltoid) and the torso (multifidus and longissimus thoracis) muscles. Strength recovery was monitored using post-exertion MVCs over a 15-minute period. Compared to the younger group, older individuals exhibited lower muscular strength, longer endurance time, and slower progressions of local fatigue. Age effects on fatigue were typically moderated by effort level, while effects of gender appeared to be marginal. Comparable (non-linear) relationships between target joint torque and endurance time were observed for shoulder exertions in the two age groups. These relationships differed between age groups, however, for torso extensions. Overall, the effects of age seemed to be more substantial and more consistent for the shoulder muscle than for the torso muscles, indicating possible differences in muscle responses when subjected to sustained exertions and likely related to differences in muscle fiber type composition. In summary, this study suggests that differences in static work capacity do exist between older and younger individuals, but that this effect is influenced by effort level and the muscle tested.

2.1 INTRODUCTION

Aging has well-known associations with sarcopenia (loss of skeletal muscle mass) and decline in muscular strength (Evans, 1995). Though age-related decreases in physical activity may contribute, several underlying changes in the neuromusculoskeletal system have been identified as primary contributors to sarcopenia. These changes include a reduction in the number of active motor units (Campbell et al., 1973; Doherty et al., 1993), a slower firing rate of motor units (Kamen et al., 1995; Connelly et al., 1999; Erim et al., 1999), and a smaller number of muscle fibers (Lexell et al., 1988) and fiber size (Trappe et al., 2001). As a result, age-related declines in muscular strength have generally been reported, but these losses appear to vary among different muscle groups (Bemben et al., 1991; Bäckman et al., 1995).

Along with a loss of muscle fibers, a shift in the relative proportions of muscle fiber types appears to occur with advancing age due to a selective reduction in Type II (fast-twitch) fibers (Lexell et al., 1988). Some studies on the vastus lateralis muscle, for example, have indicated an increase in the proportion of type I (slow-twitch) fibers with aging (e.g., Larsson and Karlsson, 1978) and a substantial reduction of type II/type I area ratio in older groups (Clarkson et al., 1981; Proctor et al., 1995). In the biceps brachii muscle, Klein et al. (2003) similarly found an age-associated decrease in type II/type I area ratio.

While sarcopenia has clear consequences in terms of loss in maximal strength, the shift in fiber type proportion has been hypothesized to play at least a partial role in the higher fatigue resistance that has been reported for older individuals. Examples include longer endurance time with age for sustained elbow flexion at submaximal efforts (Bilodeau et al., 2001a), and for handgrip and knee extensor muscles (Petrofsky and Lind, 1975; Taylor et al., 1995; Smolander et al., 1998), though the age effects found were not always significant. The role of fiber type proportion in causing these age-related differences is supported by evidence of correlations between decreasing Type II fiber area of the vastus lateralis and increasing endurance time for knee extensions (Larsson and Karlsson, 1978), and between relative Type I fiber area of the erector spinae and endurance time for torso extension (Mannion et al., 1998).

In addition to endurance, aging also seems to influence the development of localized muscle fatigue. Based on changes in electromyographic (EMG) measures, older individuals appear to be more resistant to fatigue. For instance, rates of EMG amplitude changes throughout an exercise were reported to be lower in older versus younger subjects (Bilodeau et al., 2001a; Hunter et al., 2004). At the same levels of prolonged submaximal efforts, older individuals have also demonstrated slower rates of decline in EMG spectral measures during exercises involving the elbow flexor (Bilodeau et al., 2001a) and ankle dorsiflexor muscles (Merletti et al., 1992; Yamada et al., 2000).

In contrast to the fairly extensive study of aging and fatigue development, the relationship between aging and recovery from exercise-induced local fatigue has received relatively little attention (Allman and Rice, 2001). Several measures have been used to monitor recovery (e.g., measurements of voluntary strength, EMG changes, twitch response), but existing evidence is often contradictory. For example, older subjects were found to have slower recovery of muscle fiber conduction velocity and EMG median frequency following a sustained maximal force of the abductor digiti minimi (Hara et al., 1998), yet no significant age effects on recovery of voluntary strength and EMG parameters were reported after sustained maximum contraction of the elbow flexors (Bilodeau et al., 2001b). Non-significant age effects were also found, following submaximal intermittent elbow flexions (60% maximum strength), in the amount and rate of force recovery (Allman and Rice, 2001). Further study is needed to address recovery from different submaximal effort levels.

Existing evidence has clearly identified several changes in the neuromuscular system, and demonstrated some consistent age effects on muscular performance and recovery. However, most of this evidence has been obtained using individuals at post-retirement age (i.e. >65 years), and from a limited set of muscles (mostly knee extensor and hand grip muscles; for review see Allman and Rice, 2002). Subject age, gender, and physical activity can potentially moderate the comparative fatigability and recovery that has been observed between younger and older individuals (Garg, 1991; Allman and Rice, 2002). Furthermore, the aging process appears to vary across muscles (Pastoris et al., 2000; Klein et al., 2001), suggesting the potential for differential age effects across different muscle groups. Both issues served as motivation for the current investigation.

The purpose of the present study was to quantify differences in muscular endurance, fatigue, and recovery between older and younger individuals in response to several levels of sustained static efforts. Several reports have demonstrated decreased muscular capacity (i.e. shorter endurance time, more rapid fatigue development) when young and old individuals perform tasks at the same absolute level of effort. As has been done in several more recent studies, efforts used here were normalized to individual capacity in order to isolate endurance/fatigue from confounding effects of age-related strength decrements. It was hypothesized that age-related differences in muscular capacity would be apparent at an earlier point than (post) retirement age, so this study focused on individuals whose age was that typical of that at the end of working life. This pre-retirement age group focus was also based on the projected increase of older worker population in the workplace. More than 11 million people between the age of 55 to 64 years are expected to enter the labor force by 2010, which will result in an increase in older workers of about 46% from 2000 (Horrigan, 2004). Two muscle functions were investigated, specifically shoulder abduction (middle deltoid) and torso extension (lumbar extensors). Occupationally, symptoms and complaints related to the shoulder (Bjelle et al., 1981, van der Windt et al., 2000) and torso (Andersson, 1991; Guo et al., 1999) are common and frequent among industrial workers. These muscles also differ in fiber composition (Manta et al., 1996; Mannion et al., 1997) and can be considered functionally different (task vs. postural muscles, respectively), leading to expected differences in habitual (and occupational) use. Therefore, it was hypothesized that age-related differences in muscular capacity would differ between the two muscle groups.

2.2 METHODS

Fatigue development and recovery were evaluated under different workload conditions using a repeated measures design. To address variations in motor unit control with exertion level (De Luca et al., 1996), three levels of sustained isometric activity were used (30, 50, and 70% of maximum voluntary contraction or MVC), representing relatively low, moderate, and relatively high levels. Two separate experiments were conducted: one addressing shoulder abduction and the other addressing torso extension.

2.2.1 Participants

Two groups of participants (n=24 in each, gender balanced within each group) were recruited from the local community (Table 2.1). Only those with moderate levels of daily physical activities (self reported) and no injuries or disorders (past 12 months) were allowed to participate. Due to the complexity of hardware set-up, participation in shoulder extension exercise was limited to right-handed individuals. All older individuals were screened for contraindications by an occupational physician prior to participation. Informed consent, using procedures approved by the Virginia Tech Institutional Review Board, was obtained prior to the experiment.

Table 2.1 Descriptive data on participants (mean ± S.D).

	Shoulder Abduction		Torso Extension	
	Younger	Older	Younger	Older
n	12M + 12F	12M + 12F	12M + 12F	12M + 12F
Age (yr)	21.7 ± 1.9	61.5 ± 4.3	21.5 ± 1.2	60.8 ± 4.0
Height (cm)	170.8 ± 7.7	167.3 ± 7.8	172.8 ± 9.0	166.9 ± 7.7
Mass (kg)	71.5 ± 13.6	78.5 ± 14.0	71.6 ± 10.9	77.0 ± 14.1

2.2.2 Procedures

Each experimental session consisted of initial warm-up and practice, baseline MVC measurement, endurance testing, and MVC measurements during recovery. The three effort levels used during endurance tests (30, 50, and 70% MVC) were performed on three separate days, with a minimum of two days of rest in between. To minimize potential order effects, the ordering of effort levels was completely counterbalanced for each age and gender groups. Detailed procedures for each experiment are described below.

<u>Experiment 1: Shoulder Abductions</u>

Each participant was comfortably stabilized with shoulder and waist straps in a seated position (Figure 2.1) in a dynamometer (Biodex^{TM} System 3 Pro Medical System, Shirley, New York, USA). This machine has been reported to have acceptable reliability and validity for measuring position and torque (Drouin et al., 2004). The dynamometer's center of rotation was aligned with the center of the shoulder joint, approximated below the acromial process (Nussbaum and Zhang, 2000), while the right elbow was attached to the dynamometer arm using a padded strap. After a brief warm up consisting of six sets of ten repetitions of arm abduction, MVC was determined according to the general methods proposed by Chaffin et al. (1991). Each participant was instructed to maximally abduct their right arm in the frontal plane against the dynamometer padding while the arm was horizontal. The left arm remained resting at the participant's side. Each recorded MVC value was corrected for gravitational effects on the participant's arm and dynamometer attachment. A minimum of three MVCs were performed, with two minutes of rest given between each. Additional MVCs were performed if the torque exerted markedly increased from the previous trial. The largest torque was designated as the participant's MVC.

Figure 2.1 Participant posture for isometric shoulder abduction (right hand at horizontal position with feedback of exerted torque shown on a computer screen).

Following a 10-min rest, each participant performed a static endurance test at one of the three effort levels. All postures were the same as used during the MVC tests. Torque feedback was displayed on a computer screen located directly in front of the participant. After brief practice, each participant was instructed to maintain the exertion level within ±5% of the target torque (shown on the screen) until exhaustion. A series of MVCs was then performed during recovery at 0, 1, 2, 5, 10, and 15 minutes following cessation of the static exertion. Throughout each test (MVC, endurance, and recovery tests), non-threatening verbal encouragement was provided, and efforts were made to ensure that participants were comfortable. All postures and fixture configurations were recorded and kept consistent across experimental days.

Experiment 2: Torso Extensions

Torso extension tasks (MVC, endurance, and recovery) were performed using similar procedures. Each participant was stabilized using a hip fixture designed to maintain an upright posture with hips and knees strapped to padded constraints (Figure 2.2). A force plate (OR6, Advanced Mechanical Technology Inc., Massachusetts, USA) was placed underneath the fixture, and torques exerted (at the L5/S1 level) were estimated from the recorded force and relative distance to the force plate (Granata et al., 1996). Participants' MVC was determined by instructing them to stand upright and to maximally extend the torso in the sagittal plane against fixed padding strapped over the upper torso. All other test procedures were the same as those employed during shoulder abductions.

Figure 2.2 Participant posture for isometric torso extension (the lower body is stabilized using a fixture attached to a force plate).

2.2.3 Surface EMG Signal Acquisition and Processing

Surface EMG signals were obtained from the middle deltoid during shoulder abductions, using a pair of Ag/AgCl electrodes (inter-electrode distance of 2.5 cm) located at the muscle belly (Hermens et al., 2000). During torso extensions, four pairs of the same electrodes were placed bilaterally over the erector spinae muscles at the L1 and L4/L5 levels to target the longissimus thoracis and the multifidus muscles, according to Biedermann et al. (1990), Larivière et al. (2002), and Hermens et al. (1999). For both types of exercise, electrode locations were marked and recorded to ensure consistent placement in subsequent experimental sessions. Prior to data collection, the skin was shaved, gently abraded, and cleaned with rubbing alcohol, with an inter-electrode resistance less than 10kΩ considered acceptable. The clavicle or the C7 vertebral process was used for grounding.

EMG signals were recorded continuously during both experiments using an EMG amplifier system (Measurement System Inc., Ann Arbor, MI, USA). Raw signals were pre-amplified (x100) near the electrode site and hardware filtered at 10-500 Hz. Raw EMG was sampled at 2048 Hz, while EMG root-mean-square (RMS) was sampled at 128 Hz (110-ms time constant). The former were then divided into 2-sec samples, and subsequently divided into three 1-sec overlapping windows (Luttmann et al., 1996). A Hanning window and Fast Fourier transform (FFT) were applied to each sample window. A final power spectral density was calculated as an average value, and used to determine the mean and median frequencies (MnPF and MdPF, respectively) according to Merletti and Lo Conte (1995). EMG RMS was low-pass filtered using software (Butterworth, zero phase-lag, 4^{th} order, 3 Hz cutoff). Each 1-sec sample of EMG RMS taken during endurance tests was averaged and normalized (nRMS) against maximum RMS values obtained during initial MVC.

2.2.4 Analysis

Independent variables included age group, gender, and level of submaximal effort, while dependent measures were endurance time, total work, rates of EMG changes (nRMS, MnPF, and MdPF), rate of MVC decline, and rate of strength recovery. Endurance time was defined as the time during which torque was maintained in the target zone, and total work was computed as the product of torque exerted and endurance time. Linear regressions were applied to determine rates of EMG change (slope) normalized to initial values (intercept), and were used as an indicator of fatigue progression. Rates of MVC decline were the percentage reduction in muscular strength (pre vs. post exertions MVC) relative to individual endurance time. Strength recovery was also represented as a percentage of initial MVC.

A three-way repeated measures analysis of variance (ANOVA) was generally used to examine the presence of main and interaction effects of age, gender, and effort level. Since individual strength (MVC) may be a factor influencing fatigue development, an initial test was conducted to investigate the effect of this potential covariate as suggested by Abebe (2005). If the covariate (MVC) was significant and a common slope parameter existed, a three-factor repeated measures analysis of covariance (ANCOVA) was considered to be more appropriate (Abebe, 2005). Results showed that the prerequisite ANCOVA conditions were not found for any dependent measures during the shoulder experiment, but were observed for all EMG-based fatigue parameters for the torso. Thus, for the latter measures, ANCOVA was employed following Winer et al. (1991).

To analyze strength recovery, post-exertion MVCs were fitted to exponential models, $Y = A * (1 - \exp^{-\lambda t})$, as suggested by Elfving et al. (2002), where λ represents the rate of recovery (min^{-1}). Subsequently, any effects of the independent variables on λ were determined using a three-way repeated measures ANOVA. Where relevant, post-hoc analyses were conducted using Tukey's HSD Test. Repeatability of initial MVC was also of interest (since different MVCs were obtained on different days), and this was assessed using intra-class correlations (ICC 2,1), standard errors of measurement (SEM), and coefficients of variation (CV) using standard procedures (Denegar and Ball, 1993, Elfving et al., 1999). Note that these measures are commonly used to verify the reliability of diverse strength tests (e.g. Essendrop et al., 2001; Symons et al. 2005). Significance for all statistical tests was concluded at $p < 0.05$.

2.3 RESULTS

Results of statistical analyses (Table 2.2) indicated the presence of effort level effects for all measures, and some evidence for age and age x effort level interaction effects for the majority of the measures. More detailed results are described below for each dependent measure.

Table 2.2 Summary of main and two-factor interactive effects for each dependent measure.

		Main effects			Interactions		
Measure	Muscle	A	G	EL	AxG	AxEL	GxEL
Endurance	Shoulder	*		***			
	Torso			***			
EMG RMS	Shoulder			***			
	Torso‡	+ /		***/***		+ / +	
EMG MnPF	Shoulder	***		***	+	*	
	Torso‡	* / **		***/***	+ / *	***/***	
EMG MdPF	Shoulder	***		***		*	
	Torso‡	+ / **		***/***	* / *	***/***	/ +
Rate of MVC Decline	Shoulder	*		***	*		
	Torso			***			
Recovery rate	Shoulder						
	Torso			+			

Note: 1. A = Age, G = Gender, EL = Effort level.
2. * : $p<0.05$, ** : $p<0.01$, *** : $p<0.001$, + : $0.05<p<0.1$ (approaching significance).
3. ‡Values for the longissimus thoracis and multifidus muscles, respectively.

2.3.1 Initial MVC

Age and gender significantly affected isometric strength in both experiments (shoulder and torso). Mean shoulder abduction MVCs were 64.5 and 52.4 Nm, respectively, for younger and older participants (a difference of approximately 19%). An interactive age x gender effect was also significant, with a greater age-related difference among males than females. For the torso, a larger MVC discrepancy between the two age groups (about 27%) was found, with only a marginal age x gender interaction effect (p=0.12). Across the two experiments, initial MVC was found to be highly repeatable (ICC > 0.96) with fairly small variability (SEM = 2.4 Nm for shoulder abductions and 10.5 Nm for torso extensions; CV < 4.9% for both).

2.3.2 Endurance Time

Shorter endurance times were associated with increased effort level (Figure 2.3). The effect of age on endurance time was significant for the shoulder (a difference of 13.2% across effort levels), but not significant for torso extension (p=0.71). Gender was not significant for shoulder endurance (p=0.76), but a trend was observed during torso extensions (p=0.11), with females having 11% longer endurance across effort levels. A marginal age x effort level interaction effect was obtained for the shoulder (p=0.13), showing a greater age-related discrepancy at lower effort levels.

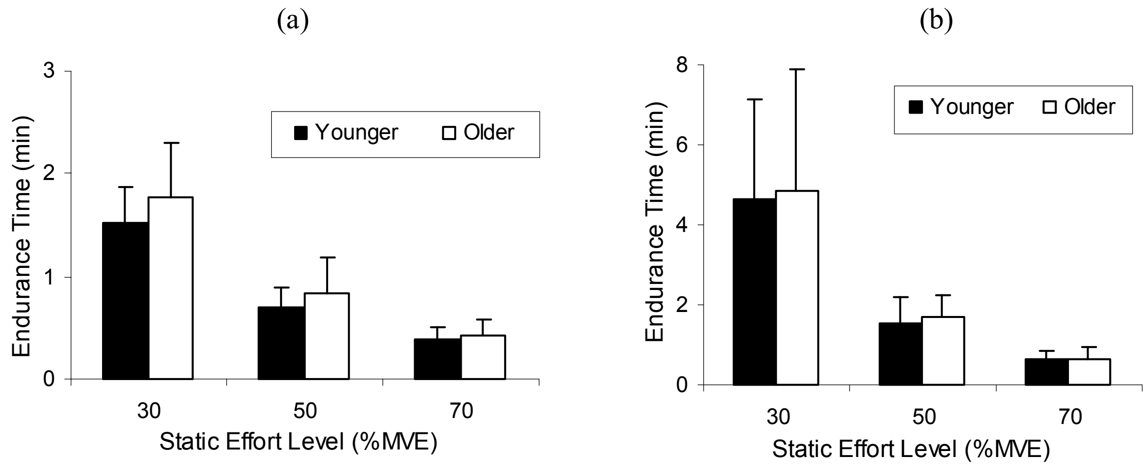

Figure 2.3 Endurance times for (a) isometric shoulder abduction and (b) torso extension.

Endurance time demonstrated an inverse relationship with target torque magnitude, and was well-modeled using a negative exponential function (Figure 2.4). A significant relationship was obtained for both age groups in both experiments (r^2=0.30 - 0.57). The relationships between younger and older individuals, however, differed between the two experiments. No significant difference between both age groups was observed for the shoulder (p=0.64), while a significant difference was found for the torso, with a distinct upward shift in the younger group. Compared to the shoulder, the amount of work generated by the torso was almost 10 times greater. Total work significantly decreased as the effort level increased, and this was found consistently in both experiments. Whereas ~24% less work was generated by older participants in torso extension, no similar age-related effect was observed for the shoulder (p=0.71).

(a) (b)

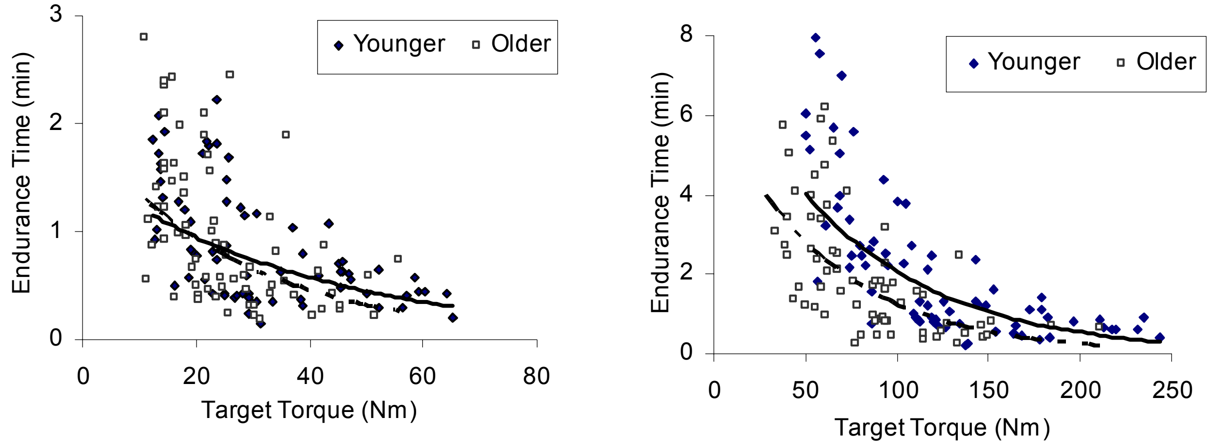

Figure 2.4 Relationship between endurance time and target torque for (a) shoulder abduction and (b) torso extension. Fitted curve for younger group (Y) represented by a *solid line* and for the older group (O) by a *dashed line*. Equations: Shoulder, Y: $y=1.56\ e^{-0.025x}$, $r^2=0.32$; O: $y=1.92\ e^{-0.036x}$, $r^2=0.30$; Torso, Y: $y=7.94\ e^{-0.014x}$, $r^2=0.57$; O: $y=6.29\ e^{-0.016x}$, $r^2=0.42$.

2.3.3 Rate of MVC Decline

At the same submaximal effort level, rates of MVC decline during shoulder abduction were almost twice as high as those for torso extension. Both experiments induced substantial reductions in maximum strength, ranging from about 4% to 40%/min. These declines were significantly greater at higher effort levels, with rates at 70% MVC three to four times greater than those at 30% MVC. Older participants had lower rates of MVC decline (about 30% for the shoulder and 14% for the torso) across effort levels. For the shoulder, while a gender effect was not observed (p=0.36), its interaction with age was significant demonstrated by the smaller differences found between older male and female participants. For the torso, no significant main or interactive effects were found.

2.3.4 EMG-Based Measures

During shoulder abduction, rates of EMG-amplitude increase varied between 4% and 37%/min. The effects of age and gender were not significant, and no two-way interactions were found (p>0.44). EMG spectral measures obtained during shoulder abduction showed consistent decreasing trends (Figure 2.5). Older participants had significantly less rapid spectral changes, with a mean difference of roughly 27%. The interactive effect of age x effort level was significant, indicating greater age-associated differences as effort level increased. Effects of gender were not evident (p=0.74).

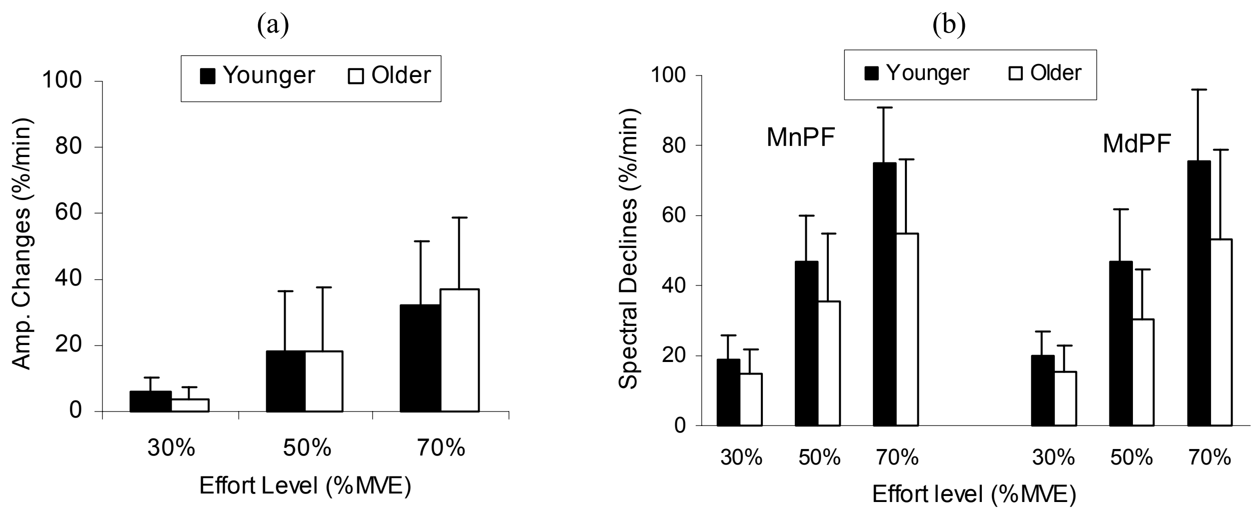

Figure 2.5 EMG (a) amplitude and (b) spectral changes during shoulder abduction.

For torso extension, no significant differences were found in EMG changes between the left and right low-back muscles for either amplitude and spectral measures. Therefore, further analyses were based on mean values of the bilateral data. Although younger individuals generally had more rapid changes in EMG amplitude, age effects were not significant for the multifidus (p=0.8) and only approached significance (p=0.07) for the longissimus. For both muscles, the interactive effect of age x effort level approached significance (p=0.07) with similar tendencies shown by the shoulder. No effects of gender nor interactions with gender were present. With regard to changes in EMG spectral measures, the two torso muscles showed similar patterns (Figure 2.6). Less rapid rates of decline were observed for older individuals, and larger slopes were found as the workloads increased. The interaction between age x effort level was significant, in which greater discrepancies in the slopes between younger and older groups were observed as effort levels increased. While an effect of gender as a main effect was not found (p>0.42), its interactive effect with age was significant. This interaction effect showed significant greater rates of the decline for younger males than for the other three groups.

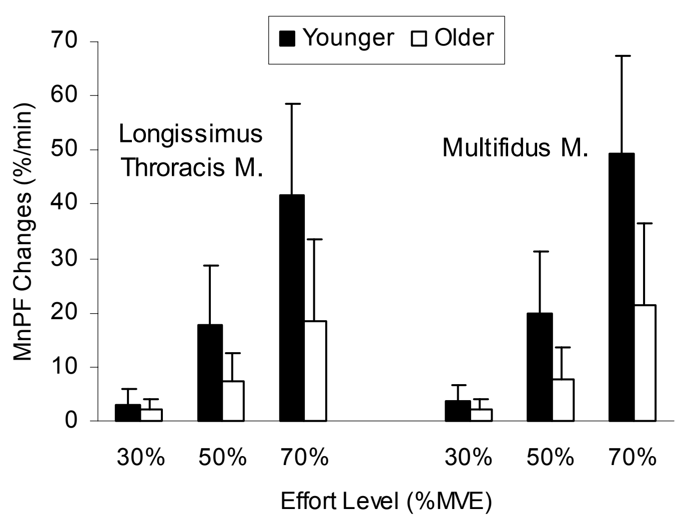

Figure 2.6 Rates of EMG MnPF decrease during torso extension (results similar for MdPF).

2.3.5 Strength Recovery

During the recovery period, a substantial proportion of muscle strength was regained within the first few minutes and leveled off afterwards. This pattern, however, was somewhat inconsistent between exercises (Figure 2.7). The magnitude of strength loss at the cessation of both exercises was significantly lower for the older group, but the difference became non-significant after 10 and 2 minutes for shoulder and torso, respectively. Roughly 75% of trials could be well-fitted using an exponential model. The rest of the data showed linear or non-monotonic trends, and these were almost equally distributed across age and gender groups. Note that only the well-fit trials were analyzed here, with no significant age or gender effects on recovery rate found in either experiment (p>0.15). The effect of effort level approached significance (p=0.07) for the torso, but was only marginal for the shoulder (p=0.12). This effect in both experiments indicated a tendency for larger recovery rates following exercises at lower effort levels.

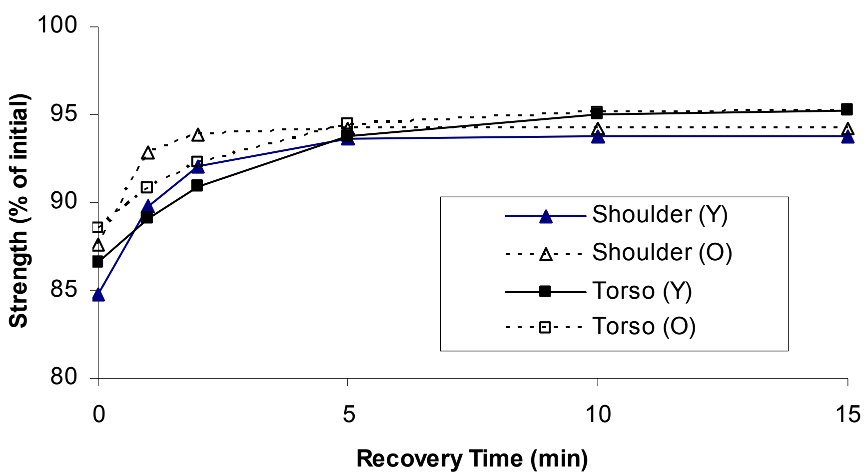

Figure 2.7 Strength recovery following shoulder abduction and torso extension. Equation: Shoulder Y: RS = 93.8 – 8.99 $e^{-0.82 * t}$; O: RS = 94.2 – 6.57 $e^{-1.58*t}$; Torso Y: RS = 95.2 – 8.67 $e^{-0.35*t}$; O: RS = 95.3 – 6.69 $e^{-0.41*t}$, where RS = remaining strength, t=time.

2.4 DISCUSSION

The present study aimed to quantify the effects of age on muscle endurance, fatigue, and recovery from sustained isometric efforts using two different exercises (shoulder abductions and torso extensions). The major findings suggest that older individuals (here 55-65 years) have distinctly increased muscular endurance and less rapid developments of local fatigue when performing efforts at comparable levels of relative effort to younger individuals. Differences due to aging were generally more substantial and more consistent during shoulder abductions, suggesting a muscle dependency of these age effects.

2.4.1 Muscle Strength

Older individuals had about 19% less initial (pre-fatigue) strength in shoulder abduction. This result is comparable to earlier reports using similar exercises and age ranges that showed a difference of 16% (Bäckman et al., 1995) and 23% (Hughes et al., 1999). A slightly greater difference (27%) in torso extension strength was found between the two age groups. This difference, however, is also comparable to the results of Viitasalo et al. (1985), who found that individuals aged 51-55 and 71-75 years had roughly 15% and 35% less torso extension strength, respectively, when contrasted with a younger group (31-35 years).

It is generally accepted that a decline in maximum strength with advancing age occurs after the fifth decade of life (Johnson, 1982; Bemben et al., 1991; Bäckman et al., 1995). This strength decline with aging has been attributed to a reduction in both quality and quantity of excitable muscle mass (Larsson et al., 1979; Vandervoort and McComas, 1986; Roos et al., 1999), primarily due to irreversible fiber damage and a permanent denervation (Brooks and Faulkner, 1994; Lexell, 1995; Payne and Delbono, 2004). The fact that the decline in MVCs in this study differs between muscles is in accordance with earlier observations of variability across muscles (Bemben et al., 1991, Lynch et al., 1999). Muscle dependency in the age effect may be due, in part, to a more substantial reduction in the use of lumbar extensor muscles with aging (or less habitual use) as compared to the shoulder musculature (Garg, 1991; Jørgensen, 1997).

Given that present study focused on endurance and fatigue during participant-specific (relative) effort levels, the results clearly depended on the ability of each participant to exert maximal efforts consistently across experimental sessions. Irrespective of age, this requirement appears to be fulfilled by well-trained and motivated healthy individuals (Vandervoort and McComas, 1986; De Serres and Enoka, 1998; Kent-Braun et al., 2002; Lanza et al., 2004). More specifically, reports by Kent-Braun et al. (2002) and Lanza et al. (2004) also indicated no substantial age-related effects in the ability to maximally activate muscles even at the cessation of a fatiguing task. Further evidence of MVC quality is provided by the reliability of initial MVCs obtained here, as shown by excellent ICC values and relatively small SEMs and CVs observed across days.

2.4.2 Age-related Differences in Muscle Endurance and Fatigue

Findings from this investigation confirmed the presence of age effects in isometric endurance time (Larsson and Karlsson, 1978; Chan et al., 2000, Bilodeau et al., 2001a), though the effect during torso extension was not significant. A similar trend of longer endurance time with non significant age effects has been reported earlier (Petrofsky and Lind, 1975; Taylor et al., 1995; and Smolander et al., 1998). Age-related increases in endurance are generally argued to occur as a results of a shift in the proportion of muscle fiber types (i.e. increased Type I/Type II ratio) that occurs with aging (e.g. Larsson and Karlsson, 1978), and which in turn is a consequence of selective atrophy of fast-twitch fibers (Lexell, 1995). The greater proportion of slow-twitch fibers may result in less lactate accumulation during the type of prolonged efforts studied here (Larsson and Karlsson, 1978). Note that muscle metabolism parameters, such as substrate availability and oxidative capacity, may also affect muscle fatigability (Bemben, 1998).

Aging processes affecting such metabolic activities have been suggested to vary across muscles depending on muscle morphology, utilization of the muscle, and metabolic demands (Pastoris et al., 2000). This may explain the disparity in the age effects on fatigability between shoulder and torso muscles obtained in the present study.

The older group also showed slower declines in EMG spectral measures, and similar findings have been reported in several earlier studies (Merletti et al., 1992; Yamada et al., 2000; Merletti et al., 2002). This effect is likely a consequence of the higher proportion of slow-twitch fibers in older muscles (Merletti et al., 2002; Macaluso and De Vito, 2004). As noted, a muscle with a predominance of slow-twitch fibers has a lower likelihood of lactate accumulation and acidosis (Brody et al., 1991; Kupa et al., 1995; Yamada et al., 2000). Therefore, a lower rate of decline in muscle fiber conduction velocity and EMG spectral measures can be expected (Merletti et al., 2002; Mannion et al., 1998). In the current study, age effects on changes in EMG spectra were found to be consistent across effort levels for both muscle groups tested. Similar consistency across different workloads was reported by Merletti et al. (2002) for sustained isometric elbow flexions at 20%, 40%, and 60% MVC. In addition, Bazzucchi et al. (2005) found that an age effect was present only at a high effort level (80% MVC) during prolonged elbow flexion and knee extension. Note that an interactive age x effort level was also found in the present study, with a more considerable difference observed at higher effort level. As speculated by Bazzucchi et al. (2005), this phenomenon could be explained by a relatively greater recruitment of fast-twitch muscle fibers among younger individuals at higher effort level.

In contrast to spectral measures, no age effects were found on changes in EMG amplitude (RMS) for either muscle group. Despite some controversy regarding the use of EMG RMS as a fatigue measure, including questions about reliability, this finding may be partly explained by age-related variations in motor unit behaviors in generating different levels of effort (Gerdle et al., 1997). At low-moderate effort levels (30 and 50% MVC) in both exercises, younger individuals exhibited greater progressive increases in EMG RMS. This result confirmed a similar finding by Bilodeau et al. (2001a), suggesting that the older individuals require lower muscle activation levels (smaller number of recruited motor unit and/or smaller increase in firing rate) in maintaining a target torque (Macaluso et al., 2002). A lower level of activation is also likely given that the older participants, as a whole, performed at lower levels of absolute (vs. relative) torque.

A delayed development of fatigue among older individuals was also supported by the lower rates of MVC decline, with this measure often considered as the "gold standard" for fatigue assessment (Vøllestad, 1997). Rates of MVC decline were lower for the older group in both experiments, though the effect was only substantial for the shoulder. In addition to endurance time and fatigue, muscle dependency was also observed in the age effect on total work. Results indicated a significant age effect for the torso, but a non-significant effect for the shoulder. It seems that for the shoulder, the lower muscular strength in older individuals can be compensated by a longer endurance time, resulting in comparable work to younger. However, a similar phenomenon was not observed for the torso, implying that designing occupational tasks for older individuals, where work-capacity is of interest, is muscle dependent.

In some cases, the effects of age on fatigue were moderated by gender (age x gender interaction). This interactive effect was found to be substantial on rates of MVC decline during shoulder abduction and on rates of EMG spectral changes during torso extension. Both indicated a greater effect of gender observed in younger individuals. The existence of an age x gender interaction effect on muscle fatigability has previously been documented by Hunter et al. (2004)

and Ditor and Hicks (2000), though the effect was found for different measures. Hunter et al. (2004), for example, reported similar interactive effects on endurance time and changes in RMS EMG during intermittent elbow flexion at 50% MVC, whereas Ditor and Hicks (2000) found the effect on rates of strength decline during intermittent contraction of the adductor pollicis muscle. Note that these two studies may not be comparable to the present study due to the differences in muscle groups and exercise type (sustained vs. intermittent). This interactive effect could be due to a mixed influence of muscle histology, sex hormones, and muscle capillarization, as speculated by Ditor and Hicks (2000). However, it is not clear why this effect was inconsistent across muscles and across dependent measures.

2.4.3 Muscle Endurance at Relative vs. Absolute loads

It should be noted that effort levels examined here were relative (i.e. normalized against an individual's MVC) and, hence, it is not known whether the observed age effects on endurance might also be present if absolute effort levels had been used. As indicated by Larsson and Karlsson (1978), an age-related decline in muscular endurance may be expected at comparable absolute loads due to age-associated reduced muscle strength resulting in an increased relative load in older individuals. Interestingly, however, a recent study using strength matched individuals to examine age-associated differences during sustained elbow flexion showed otherwise (Bazzucchi et al., 2005). Although only a small number of subjects were recruited (six per age group), older individuals were found to have more resistance to fatigue based on endurance time and EMG measures. In contrast, the present study involved a relatively large number of participants with a goal of minimizing several participant-related confounding factors and increasing external validity.

Indeed, the results here also allowed for examining the relationships between endurance time and absolute torque, which were found to differ between muscles (cf. Figure 2.4). During shoulder abductions, the results suggest similar abilities between younger and older individuals in performing the task, even at an absolute load. However, young and old results were not comparable during torso extensions, with lower endurance times expected for older individuals at a similar load magnitude. This muscle dependency may stem from differences between the shoulder and torso muscles in terms of muscle fiber type composition and atrophy rate. As previously mentioned, regular physical activity may also have a role, in which reductions in muscle activity among older individuals are more substantial for the torso than the shoulder.

2.4.4 Age Effects on Strength Recovery

For both experiments, muscle strength was substantially regained within the first few minutes. Several studies have found similar exponential patterns of strength recovery following a sustained effort (e.g. Clarke and Stull, 1969; Elfving et al., 2002), and this pattern has also been found for EMG-based parameters and subjective ratings of perceived fatigue (Baker et al., 1993; Elfving et al., 2002). The present results indicated that age effects on the rate of strength recovery from sustained isometric exercise may be marginal. In addition, both age groups also demonstrated similar incomplete strength recovery (~95%) at 15-minutes of recovery. Non-significant age effects on recovery rates have previously been reported after fatiguing elbow flexions at intermittent (Allman and Rice, 2001) and at maximum efforts (Bilodeau et al., 2001b). It seems that recovery from metabolic inhibition, which is mainly responsible for fatigue during a short sustained contraction, may not be affected by age due to the normalized loads used (Allman and Rice, 2001). This implies that different age effects may be expected from a longer-duration exercise (due to lower effort level or dissimilar exercise type), since such efforts may result in substantial peripheral failures at the level of excitation-contraction coupling (Baker et al., 1993).

2.4.5 Study Limitations

A few limitations in the present study warrant discussion. Pure isometric efforts in a controlled posture as studied here may have little relevance in terms of the demands of daily living or occupational tasks. Instead, muscle activities are more typically performed dynamically with different postures adopted and different muscles may be activated with varying motor unit recruitment strategies. The use of isometric efforts, however, allowed for control of numerous sources of variability associated with dynamic efforts. Similar results, though, are expected if the exercise is performed at different postures since the findings are assumned to be associated with differences in muscle morphology and histology. Further work will be conducted to address age effects using repetitive isokinetic efforts of the same muscles.

Only two tasks were examined, involving a limited number of muscles. These tasks were selected, as noted above, given differences in fiber type composition and function of the muscles involved. Results obtained may thus represent extreme cases of a muscle dependency in the influence of age on muscular performance. During shoulder abductions, only the middle deltoid was monitored via EMG. Other synergistic muscles (e.g., supraspinatus) may also involve during such exercises providing additional information about fatigue development, but requiring more complicated signal acquisition procedures. While EMG is only one measure used in this investigation, sole focus on the deltoid muscle was based on its assumed dominant contribution to torque generation during these tasks.

2.5 CONCLUSIONS

Age is a critical factor affecting an individual's physical capacity. Compared to younger participants, the older individuals exhibited decreased muscular strength, longer endurance times, and slower progressions of local fatigue, but showed with little difference in rates of recovery. With respect to fatigue, the age effect was shown to be influenced by effort level, with more pronounced effects observed as the workload increased. Age effects also appear to be muscle dependent, with more substantial differences found at the middle deltoid when compared to lumbar extensor muscles. Effects of gender were found to be less substantial than age, but an interactive effect of gender with age on fatigue was found in some cases. Further work is warranted to determine if similar effects are present among other muscles, among different age groups and during more complex efforts.

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CHAPTER III
AGE AFFECTS FATIGUE AND RECOVERY FROM INTERMITTENT ISOKINETIC SHOULDER ABDUCTION

Abstract

The present study was conducted to examine age-related differences in muscle fatigue and recovery from repetitive shoulder abductions. Two groups of 24 participants (younger: 18-25 years and older: 55-65 years) with gender balanced within each group were recruited and performed the exercise until exhaustion, at peak moments of 30% and 40% of individual maximum strength (MVC) with cycle durations of 10 and 20 secs. Development of muscle fatigue was determined based on changes in electromyographic signals (recorded from the middle deltoid muscle) and changes in ratings of perceived discomfort (RPD). After completion of the exercise, strength recovery was monitored using a series of post-MVC tests over 15-minute period. Based on endurance time, changes in EMG measures, and rates of RPD, this study indicated the existence of an age-related fatigue resistance. In agreement with this, examination of relationship between absolute effort level and rate of MVC decline indicated a downward shift for the older group. The age effect was typically influenced by effort level, whereas the effect on strength recovery was found to be non significant. In addition to above results, this study demonstrated the possibility of using standard EMG analyses for dynamic contractions.

3.1 INTRODUCTION

3.1.1 Localized Muscle Fatigue and Musculoskeletal Disorders

Work-related musculoskeletal disorders (WMSDs) appear to result from complex interactions among numerous causal mechanisms (Mathiasen and Winkel, 1992; Kumar, 2001), yet localized muscle fatigue (LMF) has been hypothesized as one of the possible risk factors (Edward, 1988; Armstrong et al., 1993). Previous research has indicated that workers performing fatiguing tasks have a variety of adverse outcomes, such as chronic rotator cuff tendonitis (Herberts et al., 1984; Bjelle et al., 1981), trapezius myalgia (Veiersted et al., 1993), and cervicobrachical disorder (Maeda, 1977). Other studies have also shown associations between WMSDs and several occupational exposures which are closely related to muscle fatigue such as muscle overuse, highly repetitive motions, sustained static muscle loading, and non-neutral body postures (Edwards, 1988; Malchaire, 2001; Forde et al., 2002).

Several theories have been proposed that describe WMSD-precipitating mechanisms as a result of local fatigue (e.g. Sejersted and Vøllestad, 1993; Forde et al., 2002). For example, fatigue caused by prolonged static postures may lead to muscular imbalances between overused and underused muscles. The overused muscles cause a reduction in peripheral circulation and may result in myalgia (Forde et al., 2002). High levels of repetition, which can also be fatiguing, may increase the pressure around peripheral nerves and initiate chronic nerve compression (Forde et al., 2002). Ischemia-reperfusion injury (Appell et al., 1999) and overload in "Cinderella" motor units (Sjøgaard and Jensen, 1997), both of which can occur along with or subsequent to fatigue, have also been mentioned as contributing to WMSDs. Thus, LMF may serve as a useful surrogate measure of injury risk, as it is responsive to exposures to these primary risk factors and there is growing evidence for a causal role of fatigue in WMSDs.

3.1.2 Aging and Localized Muscular Fatigue

In addition to LMF, age has been viewed as a potential risk factor for WMSDs (de Zwart et al., 1995). Rates of musculoskeletal complaints increase with age, in particular for physically demanding occupations (de Zwart et al., 1997). The specific role of age as a risk factor is still unclear, as there is insufficient evidence on the mechanisms responsible for the associations that have been identified (Garg, 1991). Moreover, aging occurs along with other potentially confounding variables and older workers may have an advantage due to longer workplace attachment (Pransky et al., 2005). Riihimäki et al. (1994), for example, found that previous history of back pain appears to be a better predictor of back problem than age, yet the prevalence of historical back pain increases consistently with age. Nonetheless, it is safe to conclude that age is an important factor to consider in the context of occupational injury and illness.

Since LMF and aging have likely roles in increasing WMSD risk, studies addressing both factors have grown in number over recent decades (Allman and Rice, 2002). This interest is also motivated by a dramatic projected change in population demographics. It has been estimated that older people (> 55) will account for 25% of the total population of the United States in 2010 (Horrigan, 2004). In parallel, the number of older workers between 55 and 64 years is expected to increase 46% between 2000 and 2010 (Horrigan, 2004). Compared to younger individuals, older workers are likely to have diminished muscular capacity, and these changes in capacity need to be quantified to ensure that task demands are still within older workers' capacity. To characterize physical capacity, the majority of aging studies have focused on muscle strength and muscle fatigability (Larsson and Karlsson, 1978, Johnson, 1982; Bemben et al., 1991; Frontera et al., 1991; de Zwart et al., 1995), with primary emphasis on isometric contractions (Allman and Rice, 2002). Recently, this emphasis has expanded to include dynamic contractions, as these are relevant to actual work tasks, and are also the focus of the present study.

3.1.3 Age-Related Differences in Dynamic Work Capacity

Patterns of deterioration in dynamic muscular strength with aging have been reported to be similar to that for isometric strength (Johnson, 1982, Larsson et al., 1979; Lindle et al., 1997; Jubrias et al., 1997). Both isometric and dynamic muscular strength generally exhibit a considerable decline after the fifth decade (Bemben et al., 1991; Frontera et al., 1991; Hurley, 1995). For isometric strength, this decline is mainly attributed to a reduction in both quality and quantity of excitable muscle mass (Larsson et al. 1979; Vandervoort and McComas 1986; Roos et al. 1999). For dynamic strength, in addition to above factors, the decline may also result from age-related impairments in explosive contractile velocity (Larsson et al., 1979; Dean et al., 2004). The strength loss with aging seems to occur for both concentric and eccentric contractions (Lindle et al., 1997; Christou and Carlton, 2002).

While age-related declines in dynamic strength seem to be well-accepted, mixed evidence has been found on the effects of age on muscle fatigability during dynamic contractions. Older adults have been found to be less capable of maintaining maximum concentric velocity during repetitive knee extensions (Petrella et al., 2005). In contrast, no age-associated differences in fatigue rate (i.e. relative reduction of muscle force) have been reported during repeated maximum isokinetic contractions in knee extension (Larsson and Karlsson, 1978; Lindström et al., 1997). For a similar exercise, Johnson (1982) found no significant age effect on muscle endurance which was defined as total time to maintain at least 50% peak torque. Interestingly, a recent report showed that older men fatigued less than younger men, based on a reduction in relative peak power produced after performing 90 cycles of maximum dynamic contraction of the ankle dorsiflexors (Lanza et al., 2004). As a whole, existing evidence indicates that the influence of age on fatigue during dynamic tasks is still not fully understood.

3.1.4 Aging and Recovery from Fatigue

In addition to fatigue development, several investigators have examined age-related differences in the ensuing recovery process. After an isometric exercise, age effects on recovery have been reported to be significant (with a slower recovery in an older group; Hara et al., 1998) and non-significant (Bilodeau et al., 2001; Allman and Rice, 2001). The latter is in agreement with our results obtained in the previous chapter (Chapter 2). Compared to recovery from static exercise, less information is available pertaining to age effects on muscular recovery following dynamic exercise. It has been suggested that the recovery rate following dynamic exercises tends to be slower in comparison to isometric tasks (Clarke, 1962), due to a relatively longer exercise duration performed during dynamic efforts, resulting in involvement of non-metabolic factors at the level of the cell membrane (Baker et al., 1993). This implies that age effects on recovery may differ following isometric and dynamic exercises. The work of Lanza et al. (2004) is one of limited studies examining age-associated differences in both fatigue and recovery processes from a dynamic exercise. After the same number of repetitions of maximum dynamic ankle dorsiflexion, younger and older individuals showed similar patterns of strength recovery. Note that in their study, repetitive maximum exertions were employed to induce fatigue. Thus, it is also of interest to determine whether there are age effects on recovery following fatiguing dynamic exercises performed at submaximal efforts, given that both daily and occupational tasks are characterized by submaximal effort.

3.1.5 Fatigue Assessment Methods

The presence of fatigue localized to a muscle group has been observed using a number of measures, including declines in maximum strength, changes in electromyography (EMG) signals, and increases in subjective rating of discomfort. A reduction in maximum strength is considered as the "gold standard" indicator of fatigue (Vøllestad, 1997). However, this approach requires the use of periodic strength tests, which disturb and/or interrupt the task being investigated (De Luca, 1997). Moreover, this method does not allow for real-time observations.

An alternative method, assessing LMF using surface EMG, is often preferable for ergonomic and clinical studies, and it is assumed that processed EMG signals can reflect underlying muscle states. For many forms of exercise, LMF is associated with increases in the EMG signal amplitude and/or shifts of the power spectra towards lower frequency (Kadefors, 1978; Komi and Tesch, 1979; Petrofsky and Lind, 1980). An increase in the EMG amplitude likely reflects increased motor unit recruitment and/or firing rate (De Luca, 1979; Suzuki, et al., 2002). In turn, spectral shifts result, at least in part, from a decline in the muscle fiber conduction velocity (MFCV) due to a decrease in intracellular pH and accumulation of extracellular potassium (Lindstrom et al., 1977; Hägg, 1992; Lowery et al., 2000). In addition to changes in MFCV, an increased synchronization of motor unit action potentials and alterations in the motor unit action potential shape can contribute to the spectral shift (Yoshitake et al., 2001; De Luca, 1979; Merletti et al., 1990; Brody et al., 1991). These trends of increasing EMG amplitude and shifting the power spectra towards lower frequency are well-established for static contractions.

In contrast, there is a lack of consensus on appropriate methods for deriving EMG-based fatigue measures during dynamic contractions. Stationarity, a requirement for standard spectral analysis (i.e. fast-Fourier transform, FFT), has been questioned for dynamic EMG due to modulations of muscle force, length, and velocity during movement (Shankar et al., 1989; Duchene and Goubel, 1993; Farina et al., 2001). As a result, several new methods of processing dynamic EMG have been proposed such as the use of time-frequency analysis or wavelets (Roy et al., 1998; Knaflitz and Bonato, 1999; Karlsson et al., 2001). Despite this, several studies have found that standard EMG analysis methods are still appropriate and reliable for dynamic EMG (Christensen et al., 1995; Potvin and Bent, 1997). Some studies, however, have found relatively high variability in comparison to static EMG (during so-called test contractions) assessed from the same exercise (e.g., Nussbaum, 2001). Given these existing discrepancies, further study is warranted to determine the utility of standard analysis methods for dynamic EMG.

Along with objective methods, subjective ratings of pain/discomfort have often been used to measure of fatigue. Among the different subjective rating scales available, Borg's CR-10 (Borg, 1990) is one of the most commonly used. The scale ranges from 0-10, where 0 corresponds "no discomfort at all" and 10 corresponds to "extremely strong (almost maximal)" discomfort. Note that while Borg's scale was developed (and defined) to assess perceived exertion, it is typically used (unmodified) to assess perceived discomfort in fatigue research. Dedering et al. (1999) found that ratings using Borg's scale were closely related to endurance time and to the slope of EMG spectral measures, and the scale also appears to have relatively high repeatability (Elfving, 1999). Nonetheless, it is of interest to determine the utility of subjective measures in quantifying fatigue, particularly during complex (dynamic) efforts.

3.1.6 Purpose of the Study

The present study determined age-related differences in fatigue and recovery following dynamic (i.e. intermittent-isokinetic) efforts. One main limitation of previous LMF and aging research is a lack of external validity, particularly in terms of the accordance between experimental protocols and actual work situations (Garg, 1991). The majority of previous studies have focused on isometric efforts (for review see Allman and Rice, 2002), which are of relatively low relevance to either daily or occupational tasks. It has also been suggested that muscle activation patterns differ between isometric and dynamic actions (Linnamo et al., 2003), and that fatigue effects may be task-dependent (Hunter et al., 2002; Christou and Carlton, 2002). In addition, physiological factors (e.g. lactate accumulation) and neural mechanisms (e.g. central activation) may play a greater role in fatigue development during dynamic situations in comparison to isometric conditions (Lanza et al., 2004). Thus, results obtained from investigations on isometric conditions may not be applicable to dynamic tasks, and further investigation of age effects on fatigability during dynamic exercise is needed.

In the current investigation, dynamic arm abduction exercises were performed via submaximal efforts, given that these effort levels conform to the majority of work tasks, in contrast to the majority of previous studies that focused on repeated maximum contractions (e.g. Larsson et al., 1979; Johnson, 1982; Lindström et al., 1997; Lanza et al., 2004). This suggests that past research on maximal contractions may be of limited applicability in the submaximal exertion realm. Since aging seems to result in changes in neuromuscular and physiological systems, with responses to an exercise depending on type and effort level (Hunter et al., 2002; Crenshaw et al., 1997), it was hypothesized that age effects on fatigue and recovery during submaximal dynamic efforts might not mirror those resulted in previous research on isometric and repeated maximum efforts. It was also of interest to compare the applicability of different fatigue indices (subjective and objective) during dynamic efforts. Furthermore, it was expected that EMG might not be applicable for fatigue assessment during such complex efforts.

3.2 EXPERIMENTAL METHODS

Fatigue development was assessed during isokinetic exercises comprised of intermittent cyclic arm abduction. All combinations of two effort levels (both low-moderate) and two cycle durations (short and long) were used, resulting in four different workload conditions. Participants performed exercises at peak moments of 30% and 40% of individual maximum strength, with 5 and 10 consecutive cycles of arm abduction followed by an equal rest period (work-rest ratio of 1:1). It was assumed that effects of effort level and cycle duration could specifically be concluded from these selected conditions.

3.2.1 Participants

A total of 24 younger and 24 older participants were recruited from the local community (Table 1), with gender balanced within each group. Target age ranges within each group (18-25 and 55-65) were selected to represent individuals at the beginning and end of working life. Participants were screened for shoulder injuries or disorders sustained during the previous 12 months and were also examined by an occupational physician. Due to the complexity of hardware set-up, participation was limited to right-handed individuals. Only those with non-extreme (moderate) levels of self-reported daily physical activity were included. All participants completed an informed consent procedure approved by the Virginia Tech Institutional Review Board.

Table 3.1 Descriptive data on study participants (mean ± SD).

	Younger	Older
Age (yr)	21.5 ± 1.89	61.2 ± 4.43
Stature (cm)	170.2 ± 8.15	166.8 ± 8.73
Mass (kg)	71.7 ± 13.53	75.9 ± 15.37

3.2.2 Procedures

In addition to an initial practice session, participants completed four experimental sessions. The four workload conditions (30%/10s, 30%/20s, 40%/10s, 40%/20s, representing combinations of effort level and cycle duration) were performed on different days. Each experimental session consisted of pre-fatigue MVCs (maximum voluntary contractions), an endurance test, and post-fatigue MVCs (recovery period). To minimize potential order effects, a

Latin-square design was used to counterbalance treatment presentations. A minimum of two days of rest was given between sessions, with additional rest was given if residual discomfort was reported. All postures and fixture configurations were recorded and maintained across days.

Pre-fatigue MVCs

Each participant sat on a dynamometer (Biodex^{TM} System 3 Pro Medical System, Shirley, New York, USA) with securing straps at the shoulder and waist. The dynamometer's center of rotation was aligned with the participant's shoulder joint center (Figure 3.1) which was estimated below the acromial process (Nussbaum and Zhang, 2000). The dynamometer was activated in isokinetic mode. After a brief warm up, each participant was instructed to generate a maximal effort (as fast and hard as possible) in the scapular plane (abduction) against the dynamometer attachment, with a set speed of 90º/sec. High reproducibility has been reported for this type of strength test (Holmbäck et al., 1999; Dvir and Keating, 2001) and for the particular dynamometer (Drouin et al, 2004). Recorded torques were corrected for gravitational effects on arm and attachment mass. Three trials were conducted with two minutes of rest given between trials. If the torque exerted markedly increased (more than 10%) from the previous trials, an additional trial was carried out. The largest torque was determined as the participant's MVC.

Figure 3.1 Equipment and fixtures for isokinetic shoulder abduction (instructions shown on a computer screen).

Endurance Test

Following a 10-min rest, each participant performed intermittent-isokinetic efforts either to their limit of endurance or up to one hour, whichever occurred first. The test required repetitive cycles of shoulder abduction-adduction (from arm at side to arm horizontal, range of motion = 90°). A small pad, that contacted the dorsum of the hand when the arm was horizontal, was used to control the abduction motion limit. Repeated sets of efforts were performed, consisting of two cycle durations: either 5 motions with 10 seconds rest between each or 10 with 20 seconds rest. Both combinations resulted in the same amount of work (torque-time product). During the last cycle, a two-second sub-maximal static test contraction was maintained in the middle of the RoM (Figure 3.2). Consistent instructions (up, down, hold, and rest) were provided on a computer screen that facilitated control of the movement velocity and rest period. Metal weights were attached above the arm padding so that peak moment was equal to 30 or 40% of peak MVC torques. The peak moment was determined with the arm and attachment in a horizontal position. During the exercises, the angular position of the dynamometer attachment was recorded at 2048 Hz

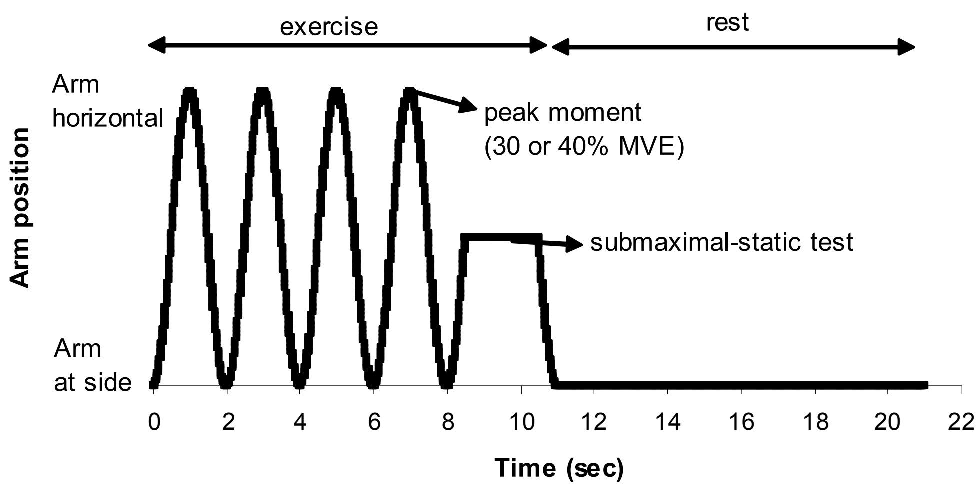

Figure 3.2 Arm positions with 5 cycle durations followed by 10 sec rest.

Post-fatigue MVC Test

A series of recovery MVCs was performed 0, 1, 2, 5, 10, and 15 minutes after exercise termination. All procedures and postures were the same as those used during pre-fatigue MVC. Note that throughout all of the tests, efforts were made to ensure that participants were comfortable, and non-threatening verbal encouragement was given continuously.

3.2.3 Surface EMG Signals

Signal Acquisition

As recommended by Hermens et al. (2000), a pair of electrodes (Ag/AgCl) were placed over the contracted belly of the middle deltoid muscle, with a 2.5 cm inter-electrode distance. Prior to electrode placement, the skin was shaved, lightly abraded, and cleaned with rubbing alcohol; inter-electrode resistance was maintained at less than 10kΩ. The ground electrode was located on the clavicle. Electrode locations were recorded relative to anatomical landmarks to ensure reproducibility between sessions.

EMG signals were recorded continuously throughout the tests. Raw signals were preamplified (x100) near the electrode sites, then hardware (Measurement Systems Inc., Ann Arbor, MI, USA) amplified and band pass filtered between 10-500 Hz. Raw signals were sampled at 2048 Hz. Root-mean-square (RMS) data were obtained using a 110-ms time constant, sampled at 128 Hz, and subsequently low-pass filtered using a software-based Butterworth filter (zero phase-lag, 4^{th} order, 3 Hz cut off).

Data Reduction and EMG-Based Measures

EMG signals recorded were grouped as static and dynamic. The static EMG were obtained from the sub-maximal static test contractions performed at the end of cycles. One second of EMG data were extracted from the middle of the two second samples. Amplitudes were averaged then normalized against maximum RMS values obtained during MVCs ($nRMS_{st}$). Raw EMG were divided into three 0.5-sec overlapping windows (Luttmann et al. 1996); Hanning window and FFTs were applied to each of these sample windows, and then median and mean power frequencies were determined ($MdPF_{st}$ and $MnPF_{st}$, correspondingly) as the average across the windows.

Dynamic EMG were obtained from the exertion cycle immediately prior to the sub-maximal static contractions. Within this cycle, a window was identified (using positional data) near the maximum joint movement angle, specifically where the shoulder joint angle was between 60º and 90º (horizontal). Data segments within these ranges (30º of movement) were considered at least weakly stationary (Potvin and Bent, 1997). The length of the data windows was at least 500 ms. RMS EMG within these windows were averaged and expressed as normalized values against maximum RMS values from MVC trials ($nRMS_{dyn}$). Raw EMG were processed as above, with the addition of zero padding, to determine median ($MdPF_{dyn}$) and mean frequencies ($MnPF_{dyn}$).

3.2.4 Subjective Measures

Ratings of perceived discomfort were obtained throughout the endurance test, using Borg's (1990) CR-10 scale. The scale is continuous, ranging from 0-10 in which 0 means "no discomfort at all" and 10 corresponds to "extremely strong (almost maximal)" discomfort. The scale was visible to participants, and ratings were collected every four minutes.

3.2.5 Analysis

Independent variables included age group, gender, effort level, and cycle duration. The dependent measures were rates of MVC decline (post- vs. pre-exercise), rates of change of surface EMG parameters, rates of subjective rating change, and strength recovery during post-fatigue MVCs. Rates of MVC decline were determined as reductions in muscular strength as a percentage of pre-fatigue MVC, normalized against individual endurance time. EMG parameters and subjective ratings of discomfort were analyzed in terms of changes in time throughout the exercise with respect to initial values. Based on inspection of the data, linear regression models were used for data fitting dependent (fatigue) measures obtained as slope/intercept. Strength recovery (post-fatigue MVC at $t = 0$ until $t = 15$ min) was expressed as normalized values against pre-fatigue MVC.

An initial test was conducted to investigate the effect of MVC as a potential covariate according to Abebe (2005) as described in the previous chapter (Chapter 2). Results suggested that MVC was a significant covariate for endurance time and rates of MVC decline. For these two dependent measures, a four-factor (age, gender, effort level, cycle duration) repeated measures Analysis of Covariance (ANCOVA) was used following Winer et al. (1991). For the remaining dependent measures, a four-factor repeated measures analysis of variance (ANOVA) was employed. For analyzing post-fatigue MVCs (recovery), data were fitted to exponential models $Y = A * (1 - \exp^{-\lambda t})$ following Elfving et al., (2002). Afterward, any effects of the independent variables on λ (the rate of recovery in min^{-1}) were determined using similar four-way repeated measures ANOVA.

For EMG, two additional factors: Data type (static or dynamic) and Parameter (RMS, MnPF, or MdPF) were included in the ANOVA model to identify differences between static vs. dynamic EMG in terms of rates of EMG changes and within trial variability. Variability of EMG was expressed as a percentage, calculated as the square root of mean square of residual error (MSE) from linear regression divided by intercept of the regression line. Where relevant, post-hoc comparisons were done using Tukey's Honestly Significant Difference Test (Tukey HSD). Correlation analysis was also conducted between pairs of dependent measures. Reliability of MVCs was analyzed using intra-class correlations (ICC), standard errors of measurement (SEM), and coefficients of variation (CV), following Shrout and Fleiss (1979), Denegar and Ball (1993), and Elfving et al. (1999). Significance for all statistical tests was concluded at $p<0.05$.

3.3 RESULTS

3.3.1 Baseline MVC and Endurance Time

Initial MVC was significantly different between age groups and gender. Mean MVCs for the younger and older groups were 56.5 and 46.82 Nm, respectively, a difference of roughly 17%. Males had 44% higher MVCs than females across age groups. A trend indicating an age x gender interaction was observed (p=0.11), with a greater age-related difference in males (18%) vs. females (11%). Excellent repeatability was obtained for initial MVCs across days (ICC = 0.99 for both age groups). Both age groups showed relatively small variability, with SEM = 1.94 and 1.74 Nm and CV = 3.56 and 3.87 %, respectively, for the younger and older groups.

Older participants tended to have longer endurance times (Figure 3.3), but the age effect only approached significance (p=0.08). Roughly 63 and 69% of participants of the younger and older groups, respectively, were able to last the full hour for exercises at 30% effort level. Endurance time was significantly affected by effort level and cycle duration, with lower effort levels and shorter cycle durations typically resulting in longer endurance times. The interaction between age and effort level was significant, with differences in endurance times between the age groups occurring only at the 40% effort level.

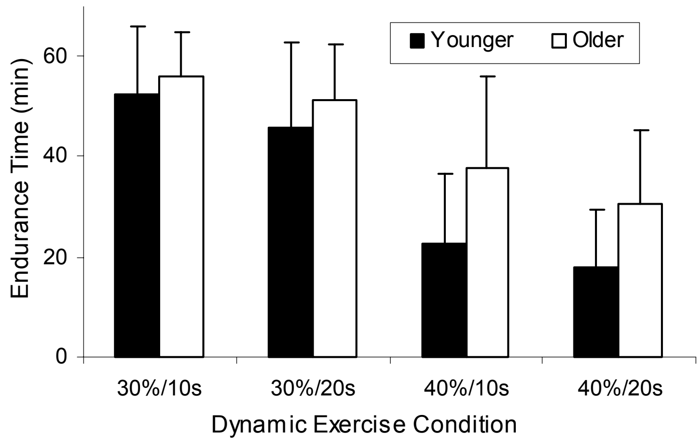

Figure 3.3 Endurance times of younger and older groups as a function of exercise condition (% effort level and cycle duration).

3.3.2 Rate of MVC Decline

Rates of MVC decline were significantly affected by age, effort level, and cycle duration. Greater rates were associated with the younger group, higher effort level, and longer cycle duration (Figure 3.4a). A significant age x effort level interaction was found, showing a greater difference in the rates between the two age groups at the higher effort level (40%). Gender was not significant as a main effect, but there was a gender x effort level interaction. Specifically, gender-related differences in the rates at the 30% level were almost double that at the 40% effort level. Exponential relationships ($r^2 = 0.23$ and 0.27) were found between rates of MVC decline and absolute effort levels (in Nm) for each age group, with results for 20 sec cycle durations shown in Figure 3.4b (though results were similar for 10 sec durations). Both curves had similar shapes (p=0.22), but significantly differed in intercept indicating a downward shift for the older group.

(a) (b)

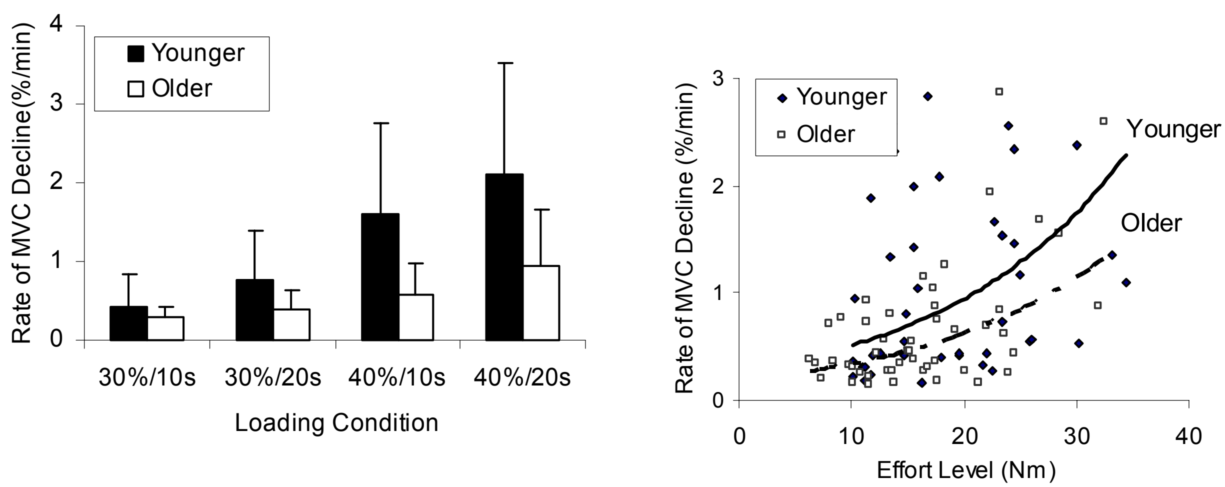

Figure 3.4 (a) Rates of MVC decline, and (b) Relationships between absolute effort level and fatigue rate at 20 sec cycle duration (Y: $y=0.27\ e^{-0.06x}$, $r^2=0.23$; O: $y=0.19\ e^{-0.06x}$, $r^2=0.27$).

3.3.3 EMG-Based Measures

The proportions of significantly decreasing slope were comparable among spectral measures, at 82%, 72%, 86%, and 77%, respectively for $MnPF_{st}$, $MdPF_{st}$, $MnPF_{dyn}$, and $MdPF_{dyn}$. $nRMS_{st}$ showed a substantially higher percentage of increasing slopes than $nRMS_{dyn}$ (75% vs. 54%). Rates of change (normalized slopes) for each EMG parameter are presented in Table 3.2, calculated from the two data types (static and dynamic EMG). The slopes of static and dynamic EMG tended to be different, though non-significant (p=0.13). Variability was significantly lower for EMG spectral measures than EMG RMS (Figure 3.5). An interaction between EMG parameter and data type was observed, showing a substantially greater difference in variability between the two data types for EMG RMS than EMG spectral measures. Variability was higher for static vs. dynamic EMG for RMS. An opposing result was obtained for MdPF and MnPF, but variability between $MnPF_{st}$ and $MnPF_{dyn}$ was not significantly different. Variability was lower for the older group, but an interaction effect between age x EMG parameter was also present. This interaction was evident as a significant difference in variability between the two age groups for nRMS (14.7 vs. 12% for younger and older groups, respectively), but not for MnPF or MdPF.

Table 3.2 Amplitude and spectral rates of change (%/min) for parameters derived from static and dynamic EMG.

		Static EMG			Dynamic EMG		
	Exercise Condition	$nRMS_{st}$	$MnPF_{st}$	$MdPF_{st}$	$nRMS_{dyn}$	$MnPF_{dyn}$	$MdPF_{dyn}$
Younger	30%/10s	0.12	0.21	0.19	0.09	0.22	0.23
	30%/20s	0.27	0.26	0.21	0.29	0.35	0.37
	40%/10s	0.62	0.70	0.63	0.52	0.78	0.76
	40%/20s	1.00	1.05	0.99	0.89	1.26	1.30
Older	30%/10s	0.09	0.15	0.11	0.10	0.14	0.12
	30%/20s	0.12	0.15	0.12	0.10	0.17	0.15
	40%/10s	0.37	0.23	0.21	0.20	0.25	0.25
	40%/20s	0.46	0.37	0.34	0.35	0.47	0.45

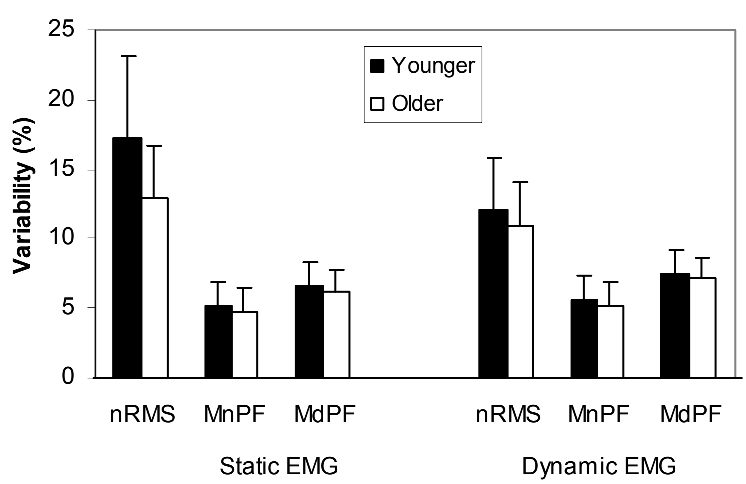

Figure 3.5 Variability of EMG-based measures (percentage of initial values).

Changes in EMG amplitude (RMS) were significantly affected by age and effort level, but not by gender (p=0.2). The rates of change in EMG amplitude for younger individuals were nearly double that for the older group. An interactive effect of age x effort level was also found, with a more pronounced difference in the rates of RMS change between the two age groups observed at 40% MVC. These rates were also significantly affected by cycle duration and its interaction with effort level. Longer cycle durations resulted in greater increases in EMG RMS, though the difference was only significant at 40% MVC.

Age, gender, and effort level had significant effects on slopes of MnPF and MdPF, with greater spectral declines found in the younger group, male participants, and higher effort level. Two-way interactions among these three factors were also present, indicating a greater age influence for males, and greater effects of age and gender at the 40% effort level. Similar to results obtained for EMG RMS, the effect of cycle duration and its interaction with effort level for spectral measures were also significant.

3.3.4 Ratings of Perceived Discomfort (RPD)

Rates of RPD increase ranged from 0.1 to nearly 0.7/min, and were significantly affected by age, effort level and cycle duration (Figure 3.6). An interactive effect of age x effort level was present, with a larger influence of increasing effort level among the younger vs. older groups. The longer cycle duration produced a significantly greater increase in RPD. The effect of gender was not significant (p=0.67), and no additional interactive effects were found.

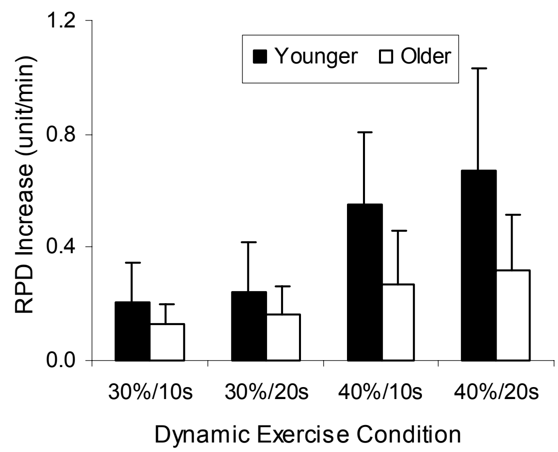

Figure 3.6 Changes in ratings of discomfort.

3.3.5 Strength Recovery (Post-fatigue MVC)

Both age groups regained a substantial proportion of muscle strength early in the rest period (Figure 3.7). The proportion of strength loss at the cessation of exercise was evidently lower for the older group, but the difference became non-significant after 10 minutes. Nearly 76% of trials demonstrated exponential changes over time. The remaining data, roughly balanced across age and gender groups, showed linear or non-monotonic patterns. Rates of recovery, computed based on exponentially well-fitted trials, were not different across age, gender, effort level, and cycle duration (p>0.19). No interactive effects were found.

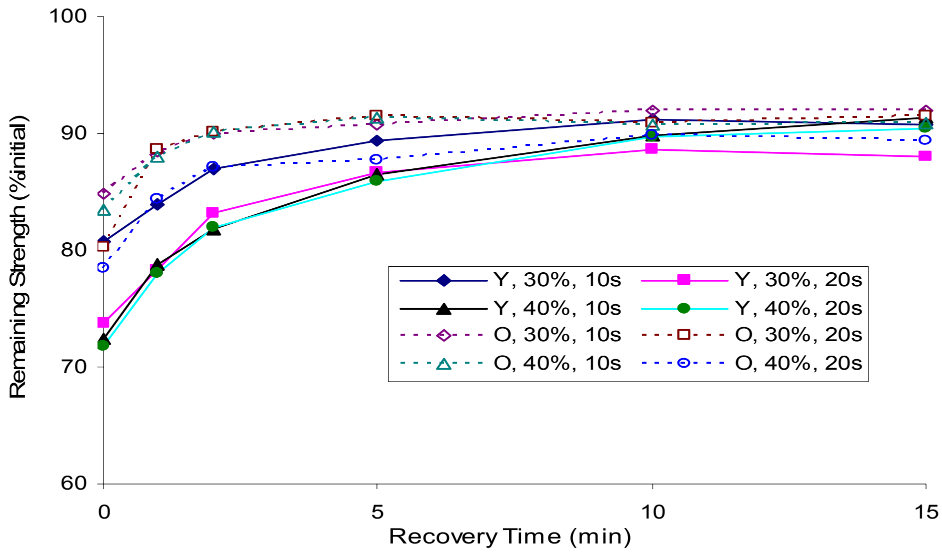

Figure 3.7 Patterns of post-fatigue MVC during the recovery period.

3.3.6 Correlations among Measures

Fairly high correlations were obtained among endurance time, RPD slope, and rate of MVC decline when correlations were calculated independently for the two age groups (Table 3.3). MVC had relatively weaker relationships with other measures, and more specifically for the older group, most relationships were found to be non-significant. Endurance time had moderate correlations with slopes of EMG parameters (r = 0.3-0.6), with relatively better correlations found for the younger group. Similar results were also observed in the relationships between slopes of EMG parameters and RPD slope, as well as MVC decline. The degree of correspondence between changes in parameters derived from EMG static and EMG dynamic to other dependent variables showed relatively similar results. Among EMG measures, higher correlation values were observed for the older group.

Table 3.3 Correlations (*r*) among dependent measures.

	MVC	$nRMS_{st}$	$MnPF_{st}$	$MdPF_{st}$	$nRMS_{dyn}$	$MnPF_{dyn}$	$MdPF_{dyn}$	RPD slope	MVC Decline
Endurance	ns/-0.29	-0.56/-0.43	-0.5/-0.3	-0.51/-0.29	-0.58/-0.37	-0.59/-0.38	-0.56/-0.39	-0.86/-0.83	-0.74/-0.71
MVC		0.25/ns	0.32/ns	0.35/ns	ns/ns	0.31/ns	0.37/ns	ns/0.28	0.31/0.3
$nRMS_{st}$			0.61/0.77	0.62/0.76	0.56/0.87	0.71/0.8	0.68/0.8	0.62/0.66	0.68/0.62
$MnPF_{st}$				0.96/0.96	ns/0.86	0.88/0.95	0.9/0.93	0.61/0.42	0.72/0.47
$MdPF_{st}$					0.2/0.85	0.87/0.91	0.9/0.9	0.61/0.42	0.69/0.49
$nRMS_{dyn}$						0.48/0.88	0.4/0.88	0.58/0.51	0.51/0.54
$MnPF_{dyn}$							0.95/0.97	0.7/0.49	0.78/0.55
$MdPF_{dyn}$								0.61/0.52	0.77/0.57
RPD									0.83/0.82

Notes: *ns* = non-significant (p>0.05); values in each cell represent younger / older groups,

3.4 DISCUSSION

The present study was conducted with a main goal of comparing the development of local muscle fatigue from intermittent-isokinetic exercises, and the subsequent recovery process, between younger and older individuals. There were two main motivations for this work. First, while a number of studies have characterized muscle fatigue during purely isometric exertions, few have assessed fatigue associated with dynamic muscular contractions. Second, there has been increasing attention paid to age-related differences in muscular capacity due in large part to anticipate dramatic changes in future workforce composition. The exertion type selected, arm abduction, is considered one important factor contributing to work-related musculoskeletal disorders of the upper limb (Aptel et al., 2002; Fagarasanu and Kumar, 2003). More generally, symptoms and complaints in the shoulder region are common among industrial workers (e.g. Bjelle et al., 1981; Herberts et al., 1984; van der Windt et al., 2000; Bjorksten et al., 2001). The current study was also designed to address limitations of previous aging studies, as reviewed by Garg (1991), including subject biases and task-related factors. Here, a relatively large number of participants were involved with ages intended to represent extremes in the workforce. In addition, the selected exercise conditions (combination of effort level and cycle duration) were intended to represent a range of occupational demands.

3.4.1 Isokinetic MVC

In the present study, voluntary exertion capacity was used as a baseline measure. Cautions have been raised on the reliability of dynamic MVC tests performed with the elderly, and extensive discussion on factors affecting reliability has been reported elsewhere (Holmbäck et al. 1999; Dvir and Keating, 2001; Symons et al., 2005), including aspects of test protocols, measurement parameters, individual factors, and number of tests, in addition to the accuracy of dynamometer. All of these issues, indeed, have been considered in our experimental protocols, and the older participants recruited here were physically active with moderate levels of daily exercise. It is worth noting that there is some evidence of no age-associated differences in the maximum ability to dynamically activate muscles before and after a fatiguing exercise (e.g. Lanza et al., 2004). Moreover, compared to one report on ankle dorsiflexor strength of young adults (Holmbäck et al., 1999), the reproducibility obtained from this work was slightly better in terms of ICC, SEM and CV.

In this study, a ~17% difference in muscular strength was found between the two age groups, comparable to that obtained for the isometric MVC tests (19%) discussed in Chapter 2. This finding also agrees with previous studies that have identified age-associated peak torque declines in isokinetic strength at slow movement speeds that are similar to reductions for isometric conditions (Larsson et al., 1979; Johnson, 1982). At faster speeds, the aging-related loss in isokinetic muscle strength is typically more evident than for isometric muscle strength (Hurley, 1995; Thelen et al., 1996; Jubrias et al., 1997) due to age-related impairments in explosive contractile velocity (Petrella et al., 2005). In addition to muscle atrophy, other factors have been postulated as reasons for age-related losses in dynamic strength. These factors include impaired motor neuron function, and reduced intramuscular blood flow and protein metabolism with aging (Larsson et al., 1979; Roos et al., 1999; Payne and Delbono, 2004).

3.4.2 Age and Gender Effects on Fatigue

The main experimental finding was that the older participants were more resistant to fatigue at normalized levels of dynamic exercise. Significant differences were found between age groups for all dependent measures including rate of MVC decline, ratings of perceived discomfort, and changes in EMG measures. Similar observations were obtained by Lanza et al. (2004) for both isometric and dynamic exercises of ankle dorsiflexor muscles. After 90 trials of maximum dynamic contraction, participants in the younger group exhibited larger reductions in peak power than their older counterparts. Comparable trends of fatigue resistance in older individuals have also been reported by Larsson et al. (1979), although the age effect was non-significant. Further investigation may be needed to confirm this result by using strength-matched individuals, as done by Bazzucchi et al. (2005) though with a limited number of participants.

The evidence of smaller proportions of Type II (fast-twitch) muscle fibers in older individuals (e.g., Lexell et al., 1988) has been proposed as a causal factor for age-related differences in endurance and fatigability during isometric efforts. Taking together results from this investigation and the previous study (Chapter 2), it seems that aging may, to some extent, have similar impacts on fatigue resistance during isometric and intermittent dynamic efforts at low-moderate effort levels. This is contradictory to an expectation that age-related fatigue resistance observed in isometric efforts may be diminished on dynamic exercises due to task complexity in terms of neural and physiological demands. Further, this implies that age-related changes in physiological systems may not limit endurance and fatigability for older individuals during intermittent-dynamic exercise at low level efforts. As reported by Aminoff et al. (1997), similar physiological responses in heart rate, blood pressure, and blood lactate occur during dynamic exercise (arm-cranking and leg-cycling) at light workloads.

It is noteworthy that the results of the present study also allowed for examination rates of MVC decline at different absolute effort levels. As previously mentioned, a reduction in MVC has been considered as a 'gold-standard' indicator of localized fatigue (Vøllestad, 1997), and from practical point of view, absolute efforts may be more meaningful than relative efforts. The present work demonstrated that the relationship between absolute workload and fatigue rate (measured via MVC decline) could be modelled using an exponential curve (cf. Fig 3.4b) for both age groups. This exponential fit was chosen based on previous studies (e.g. Hunter et al., 2004b). Interestingly, a downward shift curve was obtained for the older goup suggesting that rates of fatigue may be lower for this group even at absolute effort levels. This, indeed, supports the existence of age-related fatigue resistant in performing exercise, particularly for dynamic shoulder abduction. Further investigation appears necessary to verify this finding, for example, by using matched-strength individuals taken from both age groups.

Age effects detected in this study were found to be more important than gender effects. The effects of gender were limited to changes in EMG spectral measures (MnPF and MdPF) and accompanied by age x gender and gender x effort level interactions. Somewhat similarly, Hunter et al. (2004a) found a higher fatigability in males based on changes in EMG amplitude during intermittent elbow flexions. Note that the gender effect in our study was present for EMG spectral measures instead, not for EMG amplitude. It can be speculated that this result suggests more lactate accumulation (resulting a larger decline in fiber conduction velocity) and higher motor unit synchronization in males that could be related to gender-related muscle histology and morphology (Hicks et al., 2000). It is worth noting that, in general, gender effects on muscle fatigability are still controversial, though investigators have used strength-matched individuals in their study (Fulco et al., 1999 vs. Hatzikoutoulas et al., 2004). Interactive age x gender effects on fatigue have also been reported by Ditor and Hicks (2000) and Hunter et al. (2004b), though with varying results, for sustained isometric efforts that may not applicable for this investigation (Clark et al., 2003). Typically, the interaction was indicated by greater age-related differences for males, in agreement with our results here, that could be explained by interactive influences of changes in muscle proportion and sex hormone with aging (Hunter et al., 2004b). However, further study is warranted to verify this finding and examine further the effect of gender on muscle fatigue during dynamic exercise.

3.4.3 Effects of Effort Level and Cycle Duration

As expected, greater workload contributed to faster development of fatigue as shown by a significant effect of effort level on all dependent measures. Interestingly, the majority of results also indicated the presence of an age x effort level interaction with more prominent age-related differences in fatigue resistance at the higher effort level (40% MVC). Similar findings were obtained by Bazzucchi et al. (2005) in a study of isometric elbow flexion, in which older individuals had longer endurance times than younger individuals at 50% and 80% MVC, but not at 30% MVC. As postulated by Bazzucchi et al. (2005), the finding of more pronounced age effects at higher workloads could be explained by the order of motor unit recruitment. At a low level of effort, slow-twitch motor units are recruited in comparable fashion between younger and older groups. At higher effort levels, more recruitment of fast twitch motor units is expected for younger individuals due higher percentage of fast-twitch fibers (Lexell et al., 1988).

Intermittent tasks in the present study were conducted at short and long cycle durations. Consistent results were obtained among the main measures indicating an important effect of cycle duration, where shorter durations resulted in a slower development of fatigue. Metabolically, this result could be explained by improved restoration of blood flow in the shorter duration cycles. On the other hand, a higher metabolic demand and thereby circulatory impairment can be expected from longer contraction durations (Petrofsky et al., 2000). The lack of an interaction effect between age and cycle duration also support the intuitive conclusion that shorter work duration can benefit all age groups.

3.4.4 Fatigue Assessment using Dynamic EMG

Results of the present study indicated that changes in dynamic EMG can be used to monitor development of localized muscle fatigue. A similar conclusion has been reported by others (e.g. Hagberg, 1981; Christensen et al., 1995; Potvin and Bent, 1997; and Nussbaum, 2001), though with differing experiment protocols. Christensen et al. (1995), for example, examined slow dynamic contractions of the brachial biceps muscle at low force levels. Nussbaum (2001), similarly, compared variability and sensitivity of EMG measures obtained during fixed-level test contractions and dynamic overhead work. The results obtained in this study suggest two interpretations. First, it may be the case that static and dynamic tasks resulted in differences in EMG-based fatigue measures within a data subset. However, these differences become negligible if data are analyzed in terms of changes over time. The second follows

Nussbaum (2001), who hypothesized that motor units recruited and measured during dynamic tasks might mirror those used during static tasks. It should be noted, however, that this conclusion may be limited to contractions at a low to moderate level of effort, and when task and 'test contraction' effort levels are comparable (as here).

It has been argued that standard spectral analysis requires EMG signals need to be at least weakly stationary (Shankar et al., 1989). This 'stationarity' condition can be satisfied for most constant-force isometric contractions for epochs of 0.5-2.0 sec (Roy et al., 1998), conditions which should have been fulfilled by the data acquired from the static test contractions (static EMG). For dynamic EMG, the possibility of non-stationary data was anticipated by the use of consistent data subsets (only when the shoulder joint angle was between 60º and 90º), following earlier methods (Shankar et al, 1989; Potvin and Bent, 1997). In this study, eccentric and concentric phases were assumed to equally contribute to the power spectrum of dynamic EMG due to the nature of isokinetic movement. Further analysis will be conducted (Chapter 6) to investigate possible differences in the recruitment pattern between these two phases.

Spectral indicators (MnPF and MdPF) derived from static and dynamic EMG were found to be comparable in terms of the proportions of significantly decreasing slopes. A similar degree of variability was also observed between $MnPF_{dyn}$ and $MnPF_{st}$. EMG amplitude was found to be associated with higher variability and lower sensitivity to fatigue than the spectral measures. These results are in agreement with Potvin and Bent (1997) and Crenshaw et al. (2000), who both concluded that MnPF is a more effective variable for quantifying local fatigue than RMS for repetitive isokinetic contractions. Nussbaum (2001) reported that variability was lowest for MnPF, also consistent with the results of this study, though he found that sensitivity was slightly better for RMS than the spectral indicators. The discrepancy likely stems from the relative sensitivity of each measure to physiological changes within muscle. These changes include modifications in firing rate, motor unit synchronization, and metabolite accumulation (Lindstrom et al., 1977; De Luca, 1979; Hägg, 1992; Suzuki et al., 2002), all of which can be influenced by experimental protocols.

The fairly high correlations (0.42-0.78) observed between changes in dynamic EMG measures and ratings of perceived discomfort (as well as rates of MVC decline) support the utility of dynamic EMG in detecting local fatigue. Correlations for dynamic EMG were also comparable to or slightly better than those for static EMG. Among changes in EMG measures, results of the present study showed relatively higher correlation values for the older participant group. For example, correlation values between $nRMS_{dyn}$ and $MnPF_{dyn}$ were 0.48 and 0.88 for the younger and older groups, respectively. It was not clear why this discrepancy was observed. One possible reason might be related to age-related differences in muscle fiber morphology (e.g. threshold and motor unit firing rate) as previously discussed.

3.4.5 Age Effects on Recovery

The exponential trend of increasing strength recovery observed in this investigation is consistent with earlier evidence (e.g. Clarke, 1962, Elfving et al., 2002, Chapter 2). After a resting period of 15 minutes, about 90% of the initial strength was regained. Similar incomplete recovery following dynamic exercise was reported by Kroon and Naeije (1988). It has been postulated that long duration exercises, such as intermittent or repetitive eccentric-concentric contractions, may induce changes in the histochemistry of muscle, resulting in failure at the level of excitation-contraction coupling (Kroon and Naeije, 1988; Baker et al., 1993). Consequently, a long recovery process is required.

In this study, a larger strength loss was observed in the younger group, particularly at the end of the exercise period. Though younger participants seemed to have steeper recovery curves, the age effect was found non-significant. A similar nonsubstantial age effect was reported by Lanza et al. (2004) following repetitive dynamic MVCs. The explanation for this result is not clear. It is worth noting that fatigue and recovery seems to be independent, though the latter is a reverse process of the former. Results of this study suggest that age-related effects on both processes are different. Furthermore, this may simply be due to comparable tasks between both age groups as discussed by Allman and Rice (2001), who also found no age effect on recovery following intermittent elbow flexion. However, further study is suggested to examine age effects in the recovery process.

3.4.6 Design Implications

Given increasing numbers and proportion of older individuals in the workplace, knowledge of age-related differences in muscular performance during dynamic tasks has clear utility for ergonomic design and interventions. Results of the present study indicate an age-related increase in resistance to fatigue during dynamic tasks at normalized effort levels. This effect appears to be more prominent at higher levels of muscular exertion (i.e., age x effort level interaction). It is worth noting that older workers generally have less muscular strength than their younger counterparts, implying that assigning them less physically demanding tasks would also be a simple solution to address discrepancies in absolute effort level. Doing so may also have the added benefit of reducing fatigue development in older workers, and possibly related WMSD risk. As a whole, the results highlight the importance of considering individual factors (i.e., aging) and effort level when designing and assigning workers to occupational tasks.

Results of this study clearly demonstrate the benefit of shorter cycle duration, which might be an expected finding (although there is conflicting evidence in the literature). This result stresses the importance of designing appropriate rest breaks into the work schedule. Note that more investigation should be completed prior to reaching general conclusions, since the guideline of preferable shorter work cycles with more frequent rest breaks may only be applicable for tasks that are similar to those investigated in this study. Here, the tasks were limited to intermittent shoulder efforts with work durations lasting between 10 and 20 s (work-rest ratio 1:1), and many occupational tasks involve either shorter or longer cycles.

Findings from this investigation support the use of EMG recordings to monitor fatigue development during dynamic tasks, though specifically for intermittent shoulder efforts at low to moderate levels of contraction with speeds approximately 90°/sec. It has been proposed that static test contractions should be additionally incorporated to tasks being studied in order to obtain static EMG and allow for standard EMG processing methods. This approach has obvious disadvantages such as interruption of the task being studied and the possibility of different motor units being detected. Instead, the current results support previous reports indicating that EMG can be used directly for fatigue assessment in dynamic tasks, albeit with a slight increase in variability.

3.4.7 Study Limitations

Age effects appear to be muscle and task dependent, as previously discussed (Chapter 2). Implementation of the above guidelines may be limited to work conditions similar to those examined in this study, which simulated a dynamic task involving simple intermittent shoulder abduction-adduction performed at a constant speed. Speed of movement has been noted to affect muscle performance for older individuals (e.g. Larsson et al., 1979; Christensen et al., 1995). In this investigation, the speed was chosen based on workplace observations and to represent typical task demands. Actual work settings may be more complex, and may require movements about other axes such as flexion-extension or shoulder rotation. Despite these limitations, this study provides additional knowledge of age effects on dynamic task performance and its interaction with effort level and cycle duration.

It was assumed that the middle deltoid muscle is representative of the muscles active during the abduction-adduction movements, although other muscles such as trapezius or supraspinatus may also be involved. It is expected that similar muscle activations to those observed for the deltoid muscle would be obtained for these agonist muscles, since an assumption of consistent load sharing is made. Note that changes in arm posture may also alter muscle involvement, but this problem was minimized by asking the participants to follow consistent instructions.

Another limitation is related to the stationarity assumption employed for dynamic EMG signals and the possibility of geometrical artifacts resulting from skin movement during task performance. This limitation, however, has been anticipated with consistent data subsets following methods used in previous studies. Thus, traditional EMG methods still appear valid for assessment of fatigue during intermittent-isokinetic work. Moreover, the results obtained here showed that both data of static and dynamic EMG provide similar information when processed in terms of changes over time.

3.4.8 Conclusions

Accumulating evidence has shown the existence of age effects on muscle performance during isometric contractions. The present study was aimed to investigate similar issues in a more realistic work setting simulation (i.e. dynamic exertions). Results as a whole are similar those previously obtained for isometric contractions: older individuals are more resistant to fatigue at normalized effort levels. Age also interacted with effort level, but not with cycle duration. Gender effects appear to be less important than age, and only significant for changes in EMG measures. Age effect on recovery was found non-significant, though younger participants seemed to have steeper recovery curves. Aside from above aging issues and more specifically pertaining to the fatigue assessment method, support is provided for the use of standard EMG analyses for intermittent-dynamic exercise.

3.5 REFERENCES

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CHAPTER IV
AGING AND LOCALIZED MUSCLE FATIGUE DURING REPETITIVE DYNAMIC TORSO EFFORTS

Abstract

This study was carried out to investigate age-related differences in muscle fatigue during intermittent isokinetic efforts of torso extension. Two groups of 24 participants (older: 55-65 years, younger: 18-25 years) with gender balanced within each group performed repetitive torso extensions until exhaustion, at effort levels of 30% and 40% of individual maximum strength (MVC) with cycle durations of 15 and 30 secs. Electromyographic signals were obtained bilaterally from the longissimus thoracis and multifidus muscles, and were used to evaluate fatigue progression during the exercise. In addition, fatigue development was determined based on changes in ratings of perceived discomfort and declines in MVC. The results indicated that effects of age and gender on fatigue were non-significant, but interactive effects of age and gender with effort level were observed. This study further suggests the applicability of fatigue assessment method based on MVC reduction, but that assessment based on changes in EMG measures warrants further investigation.

4.1 INTRODUCTION

Back pain represents one of the largest musculoskeletal problems in the United States (Melhorn, 2003). Dempsey and Hashemi (1999) reported that 29.5% of total claims related to manual material handling are associated with the lower back. Baldwin (2004), in her review, noted that back problems accounted for 15-24% of claims and 23-31% of costs of US workers' compensation. According to Liberty Mutual Research News (2003), annual costs of WMSDs in 2001 were estimated to be $45.8 billion, or nearly $1 billion per week. Strains appear to be the most frequently reported causes for lower back injury (Dempsey and Hashemi, 1999). These strains can result from one or more interactions among several work-related factors, such as heavy physical work, static work load, prolonged sitting, dynamic workload, heavy manual handling, frequent lifting, trunk rotating, pushing/pulling, and vibration (Riihimäki et al. 1989).

Since the majority of the work factors contributing to low back pain (LBP) are closely related to fatigue, localized muscle fatigue has been considered to have a direct role or can be used as a surrogate in development of back pain (Leinonen, 2004). The mechanism of fatigue resulting in LBP is still inconclusive (van Dieën et al., 2003), though several theories have been proposed. In a fatigued state, there may be an increase in trunk muscle co-contraction of both agonist and antagonist muscles (Potvin and O'Brien, 1998). Though required to maintain spinal stability, an increased co-contraction may increase muscle strain and thereby risk of injury (Shier et al., 2003). A higher risk of LBP can also be expected if excessive fatigue is followed by insufficient recovery time. Kumar (2001) argued that prolonged and repeated loading, which can be fatiguing, may cumulatively reduce stress-bearing capacity, given that visco-elasticily of biological tissues. In his previous study, he found that workers with LBP had substantially higher cumulative exposure to spinal compressive load than a no-pain group (Kumar, 1990).

The incidence and prevalence of LBP may increase with age, due to degenerative changes of vertebral tissue and decline in work capacity with aging (de Zwart et al., 1995). However, conflicting results based on cross-sectional studies have been reported, showing a significant (Andersson, 1981; de Zwart et al., 1997) or non-significant (Bigos et al., 1991; Riihimäki et al., 1994) role of age as a risk factor for LBP. More evidence thus is needed, considering that aging is becoming an important issue in the workplace due to a dramatic increase in the proportion of older worker (Horrigan, 2004).

Age-related differences in fatigability during repetitive torso extensions have received little attention. Results of earlier studies suggest that aging may result in slower progression of fatigue in performing isometric efforts (e.g. Bilodeau et al., 2001). Similar results were obtained by Lanze et al. (2004) who investigated repetitive dynamic contractions of the ankle dorsiflexors at maximum effort. However, observations on different muscles (such as knee extensors) demonstrated no significant age effect on dynamic performance (Larsson and Karlsson, 1978; Lindström et al., 1997; Johnson, 1982). Further study is needed to obtain a better understanding of age effects on fatigability of torso muscles.

Surface electromyography (EMG) has been used widely as an assessment tool for monitoring localized muscle fatigue. Changes in EMG signals have beeen associated with metabolic process within the muscles (Roy et al., 1995). In real-world applications, such as for identification of physical impairments or rehabilitation programs of low back muscles, EMG-based assessment methods seem to be better than evaluations based on endurance time due to less dependency on motivation and fear of injury (van Dieën et al., 1993; Jorgensen, 1997; Leinonen et al., 2000). It has been shown, as another advantage of EMG, that changes in EMG parameters during isometric torso extension have been reported to be a good predictor of endurance time (van Dieën et al., 1993). However, such an application has never been reported for dynamic tasks.

During more complex exercises (e.g. dynamic contraction), questions have been raised concerning validity in using EMG for fatigue assessment (Roy et al., 1998). The lumbar spine is a complex muscular structure consisting of multiple layers of muscles with different orientations (Jorgensen, 1997; Clark et al., 2003). As a consequence, load sharing among muscles and 'cross talk' may occur affecting EMG signals obtained (Clark et al., 2002; Larivière et al., 2003; Stokes et al., 2003). Moreover, dynamic movement may result in non-stationarity of EMG data due to modulations on muscle force, length, and velocity that thereby violate the assumption for using standard spectral analysis (Shankar et al., 1989; Duchene and Goubel, 1993). Note also that accumulating results obtained in isometric situations may not be applicable for dynamic conditions. Shier et al. (2003) reported that there is low correlation between results obtained from static and dynamic trunk exercises, suggesting a need for further study in the area of dynamic effort which is more relevant to actual work settings.

The main purpose of this study was to examine age-related differences in the development of local muscle fatigue during dynamic torso efforts. The effort was determined intermittently at submaximal (low-moderate) levels, representative of the majority of work tasks. As described in the previous chapter (Chapter 3), age effects on muscle fatigue during dynamic efforts have received little attention with only few muscles investigated (e.g., Lindstrom et al., 1997 and Petrella et al., 2005 for knee extension; Lanza et al., 2004 for ankle dorsiflexion), and to our knowledge, no study has examined the torso. Though a similar issue has been addressed for the shoulder abduction in Chapter 3, it was hypothesized that results would differ due to muscle differences in occupational use and fiber composition (Manta et al. 1996; Mannion et al. 1997). This hypothesis was also partly inspired by our results in Chapter 2 which suggested a muscle dependency for age effect on fatigability during isometric efforts.

In this investigation, several fatigue assessment methods were employed and compared. EMG, as an objective indicator of muscle fatigue, was collected in companion with other methods, including maximum voluntary contractions (prior and after exercise) and subjective ratings of discomfort. Thus, this study also addressed the efficacy of those assessment methods, specifically in terms of sensitivity and variability. We expected that changes in traditional EMG measures, such as amplitude and spectral measures might be less consistent if applied for dynamic EMG data obtained from torso exercise due to the complexity of the back musculature.

4.2 EXPERIMENTAL METHODS

Fatigue development was assessed using comparable experimental protocols to those described in the Chapter 3. Four workload conditions, which were combinations of two effort levels and two cycle durations, were used. Levels of exercise were determined at 30% and 40% of individual's maximum strength with 5 and 10 consecutive cycles of torso extension followed by an equal period of rest (work-rest ratio of 1:1).

4.2.1 Participants

Two groups of participants (24 younger and 24 older with gender balanced) recruited from the local community (Table 1) completed all experimental sessions. Participants were selected whose ages were representative of the beginning and end of working life. Participants were screened using interview for any past (12 months) torso injury and disorders, and were also examined by an occupational physician. Participants had an exercise level of 3.4 ± 0.8 and 3.5 ± 1.1, respectively for younger and older groups, based on a self-reported assessment using a 5-point scale, where 1 correspond to "never" and 5 means "everyday'. Prior to the experiment, informed consent was obtained, using procedure approved by the Virginia Tech Institutional Review Board.

Table 4.1 Descriptive data on study participants (mean ± S.D).

	Younger	Older
Age (yr)	21.5 ± 1.2	60.8 ± 4.0
Stature (cm)	172.8 ± 9.0	166.9 ± 7.7
Mass (kg)	71.6 ± 10.9	77.0 ± 14.1

4.2.2 Procedures

In addition to a practice session, each participant completed four experimental sessions (30%/15s, 30%/30s, 40%/15s, 40%/30s representing combinations of effort level and cycle duration) on different days, with a minimum of two days of rest between. Each experimental session consisted of initial warm-up and practice, pre-fatigue MVCs (maximum voluntary contractions) test, endurance test, and post-fatigue MVC test. A Latin-square was used to counterbalance treatment presentation, to minimize potential order effects.

Pre-fatigue MVCs

Each participant stood upright and stabilized using a hip fixture (Figure 4.1), with a force plate (OR6, Advanced Mechanical Technology Inc., Massachusetts, USA) placed underneath the fixture. Padding was strapped over the upper back and connected to the lever arm of a dynamometer (Biodex™ System 3 Pro). The rotation center of the dynamometer was aligned with the participant's L5/S1 level, which was estimated based on bony landmarks. The torques exerted (at the L5/S1 level) were estimated from the recorded force and relative distance to the force plate (Granata et al. 1996). During the MVC test, the dynamometer functioned in isokinetic mode with a speed of 45º /sec. Each participant was instructed to maximally extend the torso (as fast and hard as possible) in the sagittal plane against the dynamometer attachment, with an initial posture of upper body flexed 60º and ending posture of completely upright. Three trials were performed, with two minutes of rest given between trials. If the torque exerted markedly increased (more than 10%) from the previous trials, an additional trial was carried out. The largest torque was recorded as the participant's pre-fatigue MVC.

Figure 4.1 Equipment and fixtures for isokinetic torso extension (instructions shown on a computer screen).

Endurance Test

After a 10-min rest, each participant performed intermittent-isokinetic efforts to their limit of endurance or up to one hour, whichever came first. The test required repetitive cycles of torso flexion-extension (the upper body moving between standing up-right and flexed 60º). A pad was placed to contact the chest when the upper body flexed 60º, to ensure that the range of motion (RoM) was covered consistently. Each participant performed repeated set of efforts consisting of either 5 cycles of motion with 15 seconds rest between or 10 with 30 seconds rest. During the last cycle, a two-second sub-maximal static contraction was performed at the end of the RoM (Figure 4.2). Metal weights were attached to the lever arm of the dynamometer such that external moment at L5/S1 level when the upper body flexed 30º (at the middle of the RoM), was equal to either 30 or 40% of MVC. During the exercises, the angular position of the dynamometer attachment was sampled at 2048 Hz. Instructions (up, down, hold, and rest) were provided on a computer screen for motion guidance and to facilitate consistent pacing.

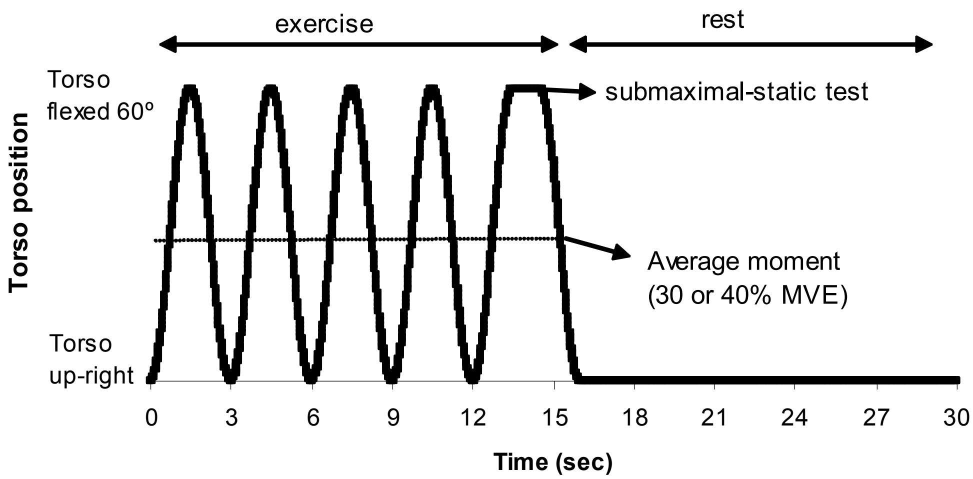

Figure 4.2 Torso positions with 5 cycle duration of flexion-extension followed by 15 sec rest.

<u>Post-fatigue MVC</u>

Immediately after cessation of the exercise, the metal weight was taken off from the dynamometer attachment and a MVC test was performed using similar procedures and posture used during the pre-fatigue MVCs. Note that during all of the tests, non-threatening verbal encouragement was given continuously and efforts were made to ensure that participants were comfortable.

4.2.3 Surface EMG

Four pairs of surface electrodes (Ag/AgCl) were placed bilaterally over the erector spinae muscles at the L1 and L4/L5 levels to target the longissimus thoracis and the multifidus muscles, according to Biedermann et al. (1990), Larivière et al. (2002), and Hermens et al. (1999), with a

2.5 cm inter-electrode distance. These medial muscles were selected based on the work of van Dieën et al. (1998) and Biedermann et al. (1990), showing that these medial muscles were more active during extension effort as compared to the lateral muscles, such as the iliocostalis. The ground electrode was placed on either the clavicle or the C7 vertebral process. To obtain good electrode-skin contact, the skin was shaved, gently abraded and then cleaned with rubbing alcohol. Inter-electrode resistance was kept below 20kΩ.

EMG signals were recorded continuously throughout the tests. EMG signals were preamplified (x100) near the electrode sites, then hardware amplified and band-pass filtered at 10-500 (Measurement System Inc., Ann Arbor, MI, USA). Raw data were sampled at 2048 Hz. Root-mean-square (RMS) data were obtained using a 110-ms time constant, sampled at 128 Hz, and then low pass filtered using software (Butterworth, zero phase-lag, 4^{th} order, 3 Hz cut-off).

EMG signals recorded were categorized into static EMG and dynamic EMG. One second sample of static EMG was extracted from the two-second samples of static test contractions based on position data of the dynamometer attachment. For each sample, RMS data were normalized against maximum RMS values obtained during MVCs ($nRMS_{st}$). Raw EMG were divided into three 0.5-sec overlapping windows (Luttman et al., 1996). Hanning window and FFTs were applied for each sample window, and then median and mean power frequencies ($MdPF_{st}$ and $MnPF_{st}$, correspondingly) were determined from the average across the windows.

Dynamic EMG were obtained from the contraction cycle immediately prior to the sub-maximal static contractions. Within this cycle, a window was determined (based on positional data) near the end of the range of motion, where the torso flexed between 35º and 60º. According to Potvin and Bent (1997), this data segment can be considered at least weakly stationary. For each window, normalized RMS ($nRMS_{dyn}$), the median ($MdPF_{dyn}$), and mean frequencies ($MnPF_{dyn}$) were determined from using similar processing methods for static RMS EMG and raw EMG, with the addition of zero padding for the latter.

4.2.4 Subjective Measures

In addition to EMG signals, ratings of perceived discomfort were also collected throughout the endurance test, using Borg's (1990) CR-10 scale. The scale is continuous (0-10), in which 0 means "no discomfort at all' and 10 means "extremity strong" discomfort. The scale was visible to participants, and the ratings were collected every four minutes.

4.2.5 Analysis

Independent variables included age group, gender, effort level, and cycle duration. The primary dependent measures were endurance time, rates of MVC decline (post- vs. pre-exercise), changes in EMG-based fatigue parameters, and changes in perceived discomfort rating. Rates of MVC decline were determined as reductions in muscular strength as a percentage of pre-fatigue MVC, normalized against individual endurance time. EMG parameters and ratings of discomfort were analyzed in terms of time-dependent changes throughout the exercise with respect to initial values.

An initial test was conducted to investigate the effect of MVC as a potential covariate according to Abebe (2005), similar to that described in Chapter 2 and 3. Results suggested that MVC was not a significant covariate for any dependent measure. Thus, a four-factor (age, gender, effort level, cycle duration) repeated measures analysis of variance (ANOVA) was employed to determine main and interactive effects on each independent measure.

For EMG, two additional factors, Muscle (longissimus thoracis or multifidus) and Data type (static or dynamic), were included in the ANOVA model. Paired t-tests were used to examine difference between data collected from the bilateral muscles. Where relevant, post-hoc comparisons were done using Tukey's Honestly Significant Difference Test (Tukey HSD). Reliability of pre-fatigue MVCs was analyzed using intra-class correlations (ICC), standard errors of measurement (SEM), and coefficients of variation (CV), following Shrout and Fleiss (1979), Denegar and Ball (1993), and Elfving et al. (1999). Significance for all statistical tests was concluded at $p<0.05$.

4.3 RESULTS

4.3.1 Baseline MVC

Age and gender significantly affected initial (pre-fatigue) MVCs. Mean magnitudes of MVCs were 285.1 and 220.2 Nm, respectively, for the younger and older groups (a difference 22.8%). Males had 38% higher MVCs than females across age groups. An interactive age x gender effect was also significant, with males having a greater age-related difference (24.9% for males vs. 19.2% for females). Excellent repeatability was found for initial MVCs across experimental sessions (ICC =0.99 and 0.98, respectively for the younger and older groups). Both age groups showed relatively small variability, with SEMs (8.42 and 7.03 Nm) and CVs (3.04 and 3.30%), respectively, for the younger and older groups.

4.3.2 Endurance Time

Nearly 77 and 67% of trials at the 30% effort level, for the younger and older groups respectively, were performed for the full one-hour exercise period. These proportions were roughly equal between genders. Overall, endurance times were comparable between the two age groups (Figure 4.3). The effects of age and gender on endurance time were not significant (p>0.13), nor was the interaction (p=0.5). Endurance time was significantly affected by effort level, but not influenced by cycle duration (p=0.85). An interactive effect of gender x effort level was found, with a greater gender-related difference observed at 40% effort level. At this effort level, females had longer endurance time (~15%) than males. No additional interactive effects were found (p>0.33).

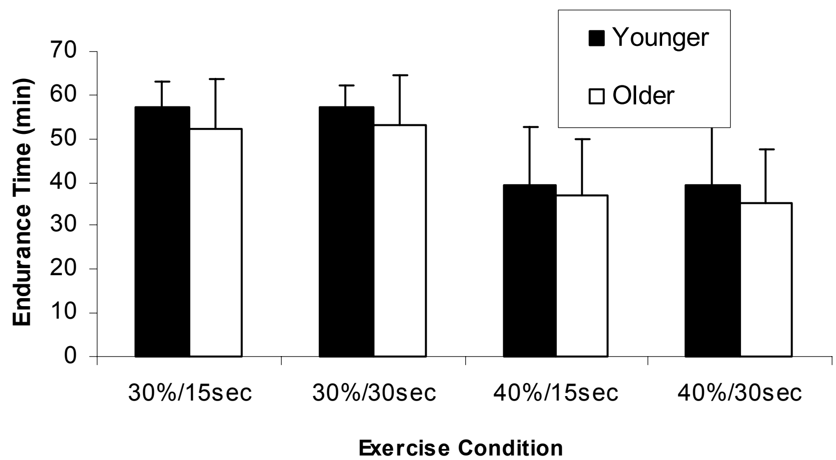

Figure 4.3 Endurance times of younger and older groups at four exercise conditions.

4.3.3 MVC Decline

At the end of exercise period, mean reductions of MVCs were 11.3 and 12.7%, respectively for the younger and older groups (non-significant difference; p=0.23). The reductions were also not significantly different between genders (p=0.20). Effects of age and gender were also not significant (p>0.16) for rates of MVC decline (Figure 4.4). Rates of MVC decline were significantly greater at higher effort level. While main effect of cycle duration was not significant (p=0.52), its interactive effect with effort level was, demonstrated by a smaller effect of effort level for the shorter cycle duration. No other interaction effects were significant (p=0.33).

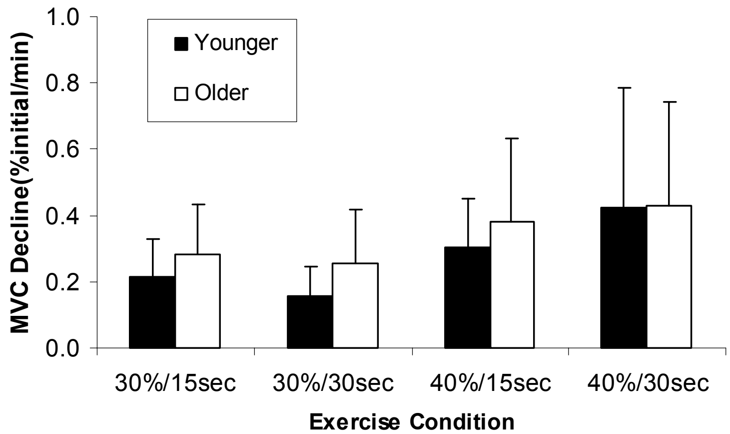

Figure 4.4 Rates of MVC decline of younger and older groups at four exercise conditions.

4.3.4 EMG-Based Measures

Table 4.2 shows time-dependent changes in the EMG measures when linear fits were applied. Since the 'expected' increasing trends on RMS and decreasing trends on MnPF/MdPF were not consistently obtained (less than 50% of trials), rates of EMG changes were determined based on means of three cycles obtained at the beginning and end of each exercise. Paired t-tests showed non-significant differences in rates of either amplitude or spectral measures changes between the left and right muscles for both muscles. Further analyses, therefore, were conducted based on mean values of the bilateral data. For each EMG measure, rates of change were not significantly different between the two data types (static and dynamic EMG) as shown in Table 4.3.

Table 4.2 Proportion of linear changes (-/0/+).

		Longissimus M.		Multifidus M.	
		Left	Right	Left	Right
Static EMG	$nRMS_{st}$	25/36/39	23/31/46	28/29/43	23/33/44
	$MnPF_{st}$	29/47/24	34/45/21	32/47/21	31/51/18
	$MdPF_{st}$	20/59/20	28/53/19	29/55/16	25/59/16
Dynamic EMG	$nRMS_{dyn}$	19/36/45	18/35/47	26/34/40	20/32/47
	$MnPF_{dyn}$	23/53/24	25/57/18	33/46/20	31/51/19
	$MdPF_{dyn}$	20/64/17	21/60/19	22/60/18	18/61/21

Note: -/0/+ represents significant negative, no trend, significant positive trends respectively.

Table 4.3 EMG amplitude and spectral changes (rates of change, in %/min).

		Exercise Condition	Longissimus M. nRMS	Longissimus M. MnPF	Longissimus M. MdPF	Multifidus M. nRMS	Multifidus M. MnPF	Multifidus M. MdPF
Static EMG	Younger	30%/15s	0.20	0.09	0.11	0.22	0.12	0.16
		30%/30s	0.23	0.07	0.10	0.25	0.10	0.11
		40%/15s	0.38	0.15	0.20	0.37	0.16	0.23
		40%/30s	0.34	0.16	0.23	0.46	0.18	0.26
	Older	30%/15s	0.19	0.12	0.16	0.18	0.10	0.14
		30%/30s	0.23	0.11	0.17	0.16	0.09	0.14
		40%/15s	0.28	0.20	0.23	0.29	0.14	0.21
		40%/30s	0.31	0.14	0.21	0.26	0.12	0.20
Dynamic EMG	Younger	30%/15s	0.17	0.11	0.12	0.17	0.11	0.12
		30%/30s	0.15	0.08	0.12	0.19	0.11	0.16
		40%/15s	0.46	0.32	0.34	0.49	0.32	0.35
		40%/30s	0.33	0.13	0.16	0.24	0.15	0.22
	Older	30%/15s	0.13	0.10	0.11	0.16	0.11	0.17
		30%/30s	0.15	0.09	0.14	0.21	0.13	0.17
		40%/15s	0.23	0.16	0.21	0.20	0.14	0.23
		40%/30s	0.25	0.23	0.30	0.29	0.15	0.24

Rates of RMS change were influenced by age, with significant effect for the multifidus and a marginal effect (p=0.1) for the logissimus muscle. For both muscles, the rates were significantly greater at 40% vs. 30% effort levels. An interaction effect between age x effort level was also found for both muscles, with a greater age-related differences at the higher workload. Gender was only significant for the multifidus muscle. Though not significant, there was an indication of interactive effects of gender x effort level for the multifidus (p=0.06) and longissimus muscles (*p*=0.12). This effect showed that the rates between the two effort levels were significantly different only among males. An effect of cycle duration was not found (*p*>0.82), and no additional interactive effects were significant.

While the age effect on rates of MnPF change was not significant for the logissimus muscle (*p*=0.78), a marginal effect (*p*=0.1) was observed for multifidus muscle. The latter demonstrated a roughly 22% difference between age groups, with older individuals associated with less rapid changes. Age effects significantly interacted with cycle duration for the longissimus muscle and with effort level for multifidus. These effects, respectively, indicated a greater difference for the younger group as the cycle duration changed and a greater age-related discrepancy at 40% workload. The effect of cycle duration was only significant for longissimus muscle. For both muscles, effects of effort level and its interaction with gender were observed. The explanation for the interactive effect was similar to that of RMS. Effects of gender were not evident for either muscles (*p*>0.26). Similar results were obtained for effects of dependent measures on rates of MdPF change.

4.3.5 Ratings of Perceived Discomfort

Rates of RPD increase varied from 0.1 to nearly 0.5/min (Figure 4.5). These rates were significantly greater at the higher effort level (0.1/min at 30% MVC vs. 0.21 at 40% MVC). While effects of age and gender were not significant (*p*>0.44), their interactive effects with effort level were observed, with a significant gender x effort level interaction and a trend of age x effort level interaction (*p*=0.07). Both interaction effects were indicated by greater age- and gender-related differences at the higher effort level. No additional main nor interactive effects were present (*p*>0.13).

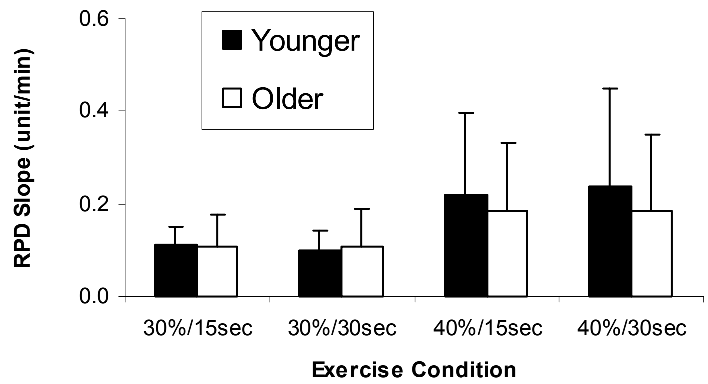

Figure 4.5 Rates of RPD increase of younger and older groups at four exercise conditions.

4.4 DISCUSSION

The present work addressed age-related differences in fatigue development during dynamic torso extension efforts, as a continuation of the previous study which focused on shoulder abduction efforts. This study was argued to be important due to (1) a high prevalence of low back pain in the workplace that could be partly linked to a result of fatiguing, repetitive back contractions, and (2) an increase in proportion of aging worker during the next decade, which suggests the need for more quantitative data on older individuals' capacity.

A question may be raised concerning the validity of initial MVCs, especially of older individuals, due to factors such as fear of injury. This would clearly affect the validity of the overall study due to the focus on participant-specific (relative) effort levels. Note that no previous study, to our knowledge, has addressed age-related differences in isokinetic torso strength. Despite this, the following evidence supports our argument that our older participants have demonstrated close to their "true MVCs". First, the magnitude of the strength loss (23%) was comparable to that of static exercises (~27% in Chapter 2, or within the range of 15-35% according to Viitasalo et al., 1985). In comparison to other muscles, our results in Chapter 3 also suggest a comparable age-related decline in shoulder strength between isometric and dynamic strength if the latter is performed at a slow speed. This is in agreement with Larsson et al. (1979) and Johnson (1982) who also found comparable age-related strength declines between both contraction types for the quadriceps muscles. Second, the initial MVCs had good repeatability and small variability across experimental days. Indeed, care had also been taken in selecting older participants who were healthy and highly motivated.

4.4.1 Muscle and Task Dependency of Age Effect

Overall, results of this study support the hypothesis that the age effect on fatigability during dynamic efforts is muscle dependent. While results of Chapter 3 suggest that aging may result in a higher fatigue resistance in performing dynamic shoulder abductions, results of the present study did not support a similar conclusion, since no difference was observed between the younger and older groups in terms of endurance time, rates of MVC decline, rates of perceived discomfort changes, and the majority of changes in EMG-based fatigue measures. The discrepancy between age effects obtained in this study and the previous study appears to be mainly due to differences in muscle fiber composition (see below for further discussion).

Torso muscles are characterized by a predominance of type I fibers. The longissimus muscle, for example, has on average 71% type I fibers (Jørgensen, 1997). In contrast, the shoulder muscle (i.e. deltoid) is comprised of a higher proportion of type II fibers with around 31.4% to 47.4% (Manta et al., 1996). Though there is a lack of data available, it can be speculated that the magnitude of type II fiber reduction, associated with aging, may be less for the torso. Since this association shift in fiber type proportion has been hypothesized to play at least a partial role in higher fatigue resistance, a less substantial age effect on fatigability of the torso muscles can be expected.

In addition to a difference in muscle fiber composition, another explanation for the discrepancy between results from the torso and shoulder is a more substantial reduction in the use of lumbar extensor muscles (or less habitual use) with aging (Garg, 1991). A reduction in use possibly leads to muscle atrophy and disease and will further affect muscle performance. This phenomenon may be more prominent for the aging torso muscle, although its role in the present study may be marginal since the older individuals recruited were active and physically healthy.

The results of this study, taking together with those from Chapter 2, also suggest a task dependency of the aging effect on muscle fatigability. Recall that the study conducted in Chapter 2 focused on isometric torso extension, and that study also involved many of the same participants that participated in the present study. While age effects were generally reported for the isometric torso exertions, a similar result was not obtained here for the dynamic torso. This discrepancy presumably stems from differences in metabolic and neurological responses between both exercise types. Note that dynamic tasks seem to be more demanding and more complicated, and aging may result in declines in metabolic capacity and in the ability to deal with a complex task (Shephard, 1999; Garg, 1991). It is worth adding that a similar task dependency was not observed in isometric-dynamic shoulder efforts (Chapter 3) or the ankle dorsiflexors (Lanza et al., 2004). This discrepancy between studies may be explained by different muscle roles and muscle sizes. Torso fatigue can be expected to have a greater impact on whole body metabolite demands than fatigue of these the two smaller muscle groups.

4.4.2 Effects of Individual and Task Factors

While a main effect of age or gender was not observed, interactive effects of both variables with effort level were found for the majority of fatigue measures. The results indicated a greater age- or gender- related difference as the effort level increased from 30% to 40% MVC. This finding is in agreement with the previous study on dynamic shoulder abduction (Chapter 3). According to Bazzucchi et al. (2005), this interactive effect can be explained by the order of motor unit recruitment. At 30% MVC, motor units recruited are dominantly slow-twitch, which seems to be comparable between age groups. At the higher workload, increased recruitment of fast twitch motor units can be expected for younger individuals due a higher percentage of these fibers (Lexell et al., 1988).

The same postulate can also be used to explain the effect of the gender by effort level interaction obtained in this study. Note that previous studies have shown a higher proportion of type I area in the females vs. males (e.g., Mannion et al. 1997). Muscles with a higher proportion of type I fibers have been considered to be more fatigue resistant since this fiber type is associated with a better ability to metabolize lactate, a larger content of myoglobin and mitochondria, and lower potassium loss (Larsson and Karlsson 1978; Jørgensen, 1997). Similar to above explanation, increased recruitment of fast twitch motor units can be expected for males than females at higher effort levels.

Greater effort level consistently contributed to faster fatigue development as shown by a significant effect of effort level on all dependent measures. This further suggests that a change of 10% in workload will substantially affect muscle performance. In this study, a reduction of 30-40% in endurance time was observed. Pertaining to the effect of cycle duration, results of this study demonstrated that both selected cycle durations were non-significantly different in affecting muscle fatigue, which conflicts with the results of dynamic shoulder abductions (Chapter 3). Though the cycle numbers were comparable between the two studies (5 and 10 cycles per set of effort), their speeds of movement differed from 3 sec per cycle for torso to 2 sec per cycle for shoulder. Since both muscle efforts had the same 1:1 work-rest ratio, a longer rest period between each set of efforts was associated with the torso (15 and 30 sec), in comparison to the shoulder (10 and 20 sec). Both factors (slower speed and longer rest period) might have led to similar fatigue accumulation and recovery rates, resulting in non-significant differences between the two cycle durations during torso exercises.

4.4.3 Longissimus vs. Multifidus Muscle

In the present study, two torso extensor muscles were monitored: longissimus and multifidus. Overall, results concerning effects of individual and task factors on fatigue based on EMG changes were comparable between these two muscles. However, muscle fatigue seemed to be typically faster for multifidus. This result can be related to the characteristics of both muscles. Histochemical studies have shown that these muscles differ slightly in fiber composition. Though both have similar relative type IIa fiber content (approximately 20%), multifidus has a higher proportion of type IIb fibers (24% vs. 11%; Jørgensen, 1997) which are less resistant to fatigue. It should be note, though, that there is also evidence of a comparable muscle fiber composition between these two muscles (Thorstensson and Carlson, 1987). Interindividual variability may account for this disagreement. Functionally, both muscles are activated during sagittal flexion-extension movements. In spite of this, the multifidus, which is located at the lower sites, may be more active during the movements. As a consequence, this muscle may fatigue faster. Note that this needs further justification, due to the complexity of torso muscles and limitations of EMG assessment methods employed as discussed below.

Previous studies have suggested that load sharing may occur among back muscles during an exercise due to an integrated system of the lumbar skeletal-muscles consisting of multiple layers of muscle. Leinonen et al. (2000) and Clark et al. (2002) found that hip extensor muscles were activated in addition to lumbar paraspinal muscles at the end of dynamic torso flexions. A similar phenomenon might have occurred in the present study. Moreover, activation of antagonist muscles may also increase during the exercise that differs across individuals or even between age groups. Macaluso and colleagues (2002) reported a marked increase in coactivation of antagonist muscle for older subjects. Further study is indeed necessary to investigate the pattern of this load sharing among synergistic and antagonist muscles and their concomitant age effects.

4.4.4 Non-linear Changes in EMG Measures

In contrast to our previous study on the shoulder muscle, time-dependent changes in EMG-based fatigue measures in the present work did not follow a linear trend. A number of factors may be responsible for this finding, assuming that a linear change is a valid sign for fatigue. The first factor is related to the multifaceted low back muscles. Even though multifidus can be expected to fatigue faster, this muscle is deeply located (non-superficial muscle). The signals recorded may be influenced by cross-talk (Stokes et al., 2003). Note that this effect may be marginal during isometric contractions since good repeatability has been reported for this situation (Larivière et al., 2002). Our previous study in Chapter 2 also found that changes in the EMG measures during isometric torso extension can be fitted using linear regression. During dynamic situations, the reliability may be reduced due to the additional possibility of geometric artifacts resulting from skin movement during task performance (Rainoldi et al., 2000). Nevertheless, this factor may be minor since non-linear changes were also observed for static EMG. Moreover, this issue was accommodated in data processing by selecting a short window of dynamic EMG.

Another possible explanation for non-linear changes in EMG measures pertains to load sharing. As previously mentioned, studies have shown that dynamic torso exercise may trigger activation of other muscles including hip muscles (Kankaanpää et al., 1998; Leinonen et al., 2000, and Clark et al. 2002). In another report, Clark et al. (2003) demonstrated a derecruitment of the lumbar muscle (after increase up to 55% of maximum) and a concomitant increase in hip extensor muscle activity during isotonic trunk extensions. The last explanation relates to the applicability of the methods employed to process dynamic EMG. The standard EMG processing methods using FFT may not be applicable for such data as suggested by Roy et al. (1998) and Knaflitz and Bonato (1999). Further studies will be conducted to examine alternative methods with a goal of obtaining more sensitive fatigue measures (Chapter 6).

4.4.5 Limitations

A few important limitations of the present study should be noted. It was assumed that multifidus and longissimus thoracis are representative of the muscles active during the torso flexion-extension movements. However, as previously discussed, other low back muscles (including hip extensor) may also be involved. Load sharing among these muscles may confound consistency of EMG changes. Nonetheless, EMG was only a portion of the measures used in this study and its results were comparable with other dependent measures such as endurance time, rates of MVC decline, and rates of RPD change. Note that changes in body posture may also alter muscle involvement, but this problem was minimized by asking the participants to follow consistent instructions.

4.4.6 Conclusions

Age effects on muscle fatigability during dynamic torso efforts have received little attention in previous studies. In this study, intermittent-tasks at low-moderate levels were conducted. The results indicated that effects of personal factors such as age and gender were non-significant, but interactive effects of age and gender with effort level were observed. This study also supports a muscle and task dependency of age effects on muscle fatigue. Pertaining to the fatigue assessment methods, this study revealed that assessment based on MVC reduction appeared to be acceptable, but assessment based on changes in EMG measures warrants further investigation.

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CHAPTER V
EVALUATION OF EMG-BASED FATIGUE MEASURES DURING LOW LEVEL ISOMETRIC EFFORTS

Abstract

Previous studies have indicated that commonly used electromyographic (EMG) parameters appear to be inconsistent in monitoring fatigue during low level isometric efforts. The purpose of the present study was to evaluate alternative EMG-based fatigue measures intended for these efforts. These alternative measures were derived from the frequency band method, a Poisson-plot method, logarithmic-power frequency, and fractal analysis. Data from prior experiments (shoulder abduction and torso extension, Chapter 2) were further analyzed. An additional experiment (10 participants) was also conducted that investigated fatigue during prolonged shoulder abductions at 15% and 30% of maximum strength (MVC). Tests of repeatability were conducted at 15% MVC. Both the alternative and traditional fatigue measures were analyzed for 'utility', in terms of sensitivity, variability, repeatability, and predictive ability. A main finding of this study is that parameters derived from fractal analysis demonstrated high utility. This suggests the potential use of such EMG-based fatigue measures in assessing fatigue during low level isometric efforts.

5.1 INTRODUCTION

5.1.1 Localized Muscle Fatigue at Low Level Efforts

Localized muscle fatigue may develop even during low level efforts (LLEs). For example, participants in Garg et al. (2002) reported a very high subjective rating of localized pain in the shoulder girdle after 15 min of holding a weight at 5% maximum voluntary contraction (MVC). Similarly, Sjøgaard et al. (1986) observed the occurrence of muscle fatigue (measured via EMG recording, strength reduction, and rating of perceived exertion) during isometric knee extension at 5% MVC, though the muscle was found to have adequate blood flow supply. Since localized muscle fatigue has been associated with a risk of work-related musculoskeletal disorders (WMSDs), it has been suggested that LLEs may also lead to these disorders (Winkel and Westgaard, 1992; Armstrong et al., 1993; Mathiassen and Winkel, 1996). As evidence, Ranney et al. (1995) found that 54% of female workers performing highly repetitive jobs at LLEs in an industrial setting had signs of WMSDs in the upper limb. Another report indicated an increasing case of upper extremity disorders associated with computer uses from 1986 to 1993, considering that these tasks are typically characterized by LLEs (Fogleman and Brogmus, 1995).

Investigators have proposed "Cinderella hypothesis" which describes a possible fatigue-related mechanism leading to WMSDs during sustained isometric contractions at LLEs. The hypothesis argues that prolonged static loading may overload low-threshold motor units (Type I) which are initially recruited and remain continuously active for a long period of time (Hägg and Ojok, 1997; Forsman et al., 2002). Recent investigations support this hypothesis. For example, motor units of shoulder abductor muscles were found constantly active with a low degree of motor unit rotation during 30-minutes exertion at 11-12 % MVC (Jensen et al., 2000). Similar findings were reported in studies investigating static contraction of the extensor digitorum communis muscle for 25-min (Forsman et al., 2002) and the trapezius muscle for 60-minutes (Thorn et al., 2002). However, this hypothesis has been criticized for disregarding the "muscular wisdom" of the body (Forde et al., 2002). This criticism argues that the body should have protective mechanisms against overload in muscle fibers, such as motor unit rotation (Westgaard and De Luca, 1999; Garland and Gossen, 2002).

Along with controversy pertaining to the validity of the "Cinderella hypothesis", non-conclusive information is available concerning the physiological processes underlying fatigue during prolonged isometric at LLEs. In prolonged static contractions at moderate-high effort levels, an increase in intramuscular pressure (IMP) is assumed to be an important cause of fatigue by restricting blood flow and thereby impairing oxygen availability to the muscle (Yoshitake et al., 2001). Several studies suggested that similar mechanisms likely occur during tasks at LLEs. Murphy et al. (2001), for instance, showed a marked reduction in tissue oxygenation (TO_2) within the extensor carpi radialis muscle during brief isometric contraction at LLEs. Similarly, Crenshaw et al. (1997) demonstrated an increase in IMP during isometric knee extensions to fatigue at LLEs. In contrast, other investigations have indicated that other mechanisms not related to IMP and TO_2 may have a major role in causing muscle fatigue at LLEs (Sjøgaard et al., 1986; Blangsted et al., 2005). These authors found that IMP and TO_2 remained constant at such effort levels, but changes in EMG signal indicated the occurrence of muscle fatigue. In short, the physiological mechanisms underlying fatigue during LLEs is still not clear.

From a neuromuscular perspective, fatigue at LLEs seems to be more affected by a failure in peripheral processes than by central failure system. Central fatigue is characterized by a reduction in voluntary muscle activation (Bigland-Ritchie et al., 1986; Gandevia et al., 1995) due to a lack of adequate central nervous system (CNS) drive to the motoneuron pool, while peripheral fatigue is associated with a failure in the contractile ability within a muscle regardless of the adequacy of excitatory impulses. The peripheral process involves activation of the surface membrane, propagation of action potential along the t-tubular system, release of calcium (Ca^{2+}), and activation of the contractile elements (Fitts, 1996; Jones, 1996). Peripheral failure in the latter process, at the level of excitation-contraction (E-C) coupling, has been hypothesized as a main cause of fatigue during isometric contractions at LLEs (Bigland-Ritchie et al., 1986; Sjogaard et al., 1986; Krogh-Lund, 1993). This phenomenon can be associated with a reduction in Ca^{2+} release and an increase in intracellular Ca^{2+}, partly caused by low pH and metabolic end products (Green, 1998). Evidence for failure in E-C coupling has been reported in some studies using low frequency muscle stimulation (Jones, 1996; Green, 1998).

Muscle strategies while maintaining a defined target force appear to be dependent upon contraction level. Commonly observed strategies include recruiting new motor units, increasing firing rate (Suzuki et al., 2002), and synchronization (tendency of grouping) of discharge (Krogh-Lund and Jorgensen, 1993). While generating a contraction at LLEs, additional strategies may be involved such as decreasing firing rate and de-recruitment of motor units, as observed by Kamo (2002) during 5-min contraction of knee extensor muscles at 5% MVC. Following this de-recruitment, larger motor units will be recruited (Hägg and Ojok, 1997; Jensen et al. 2000). As a result, motor units seem to demonstrate fairly complex behaviors when the muscle is exerted at LLEs.

5.1.2 EMG-Based Fatigue Measures for Low-Level Effort

The occurrence and development of muscle fatigue during isometric contractions at moderate-high effort levels can typically be detected from changes in surface electromyography (EMG) signals. An increase in the EMG amplitude (root-mean-square, RMS) has commonly been interpreted to an indication of increased motor unit recruitment and/or firing rate (Suzuki et al., 2002). However, mechanisms underlying changes in EMG RMS potentially differ between low and high effort levels (Gerdle et al., 1997). During contractions at LLEs, in addition to changes in motor unit recruitment and firing rate, EMG amplitude may also be influenced by varying trends in firing rate (Kamo, 2002) and additional recruitment of larger motor units (Hagg and Ojok, 1997). Moreover, EMG RMS measured during LLEs may be distorted due to a relatively low EMG signal-to-noise ratio (Baratta et al., 1998). As a consequence, the reliability of EMG RMS as a fatigue indicator may decrease.

Common EMG spectral parameters, such as the median (MdPF) (Stulen and De Luca, 1981) and mean (MnPF) power frequencies (Schweitzer et al., 1979), have yielded inconsistent results in monitoring fatigue during isometric contractions at LLEs. While the majority of previous studies have reported decreasing trends of EMG MdPF or MnPF during contraction (Jorgensen et al., 1988; Bystrom and Kilbom, 1990; Hagg, 1992; Hansson et al., 1992; Jensen et al., 2000; Madeleine et al., 2002), several investigations have documented contradictory findings. For example, increasing trends of MnPF were found in some shoulder muscles while maintaining an elevated arm position (Hagberg, 1981). Non-conclusive trends of MnPF responses were also observed for the upper trapezius muscle during isotonic and isoelectric endurance tests at 10-15%MVC (Hagg and Ojok, 1997). Moreover, no marked changes in MnPF have been reported during isometric contraction of the trapezius muscle at 15-20% MVC (Oberg et al., 1994) and 5% MVC (Thorn et al., 2002). The latter authors suggested that measures such as MnPF may not reflect Type-I fibers of the trapezius muscle, which are mainly activated during LLEs. Therefore, further study is needed to find EMG measures sensitive to fatigue during contractions at LLEs.

Several other EMG-based fatigue indices have been proposed as complementary to the existing parameters (RMS, MnPF and MdPF). These indices include quartile or decile frequencies (Linssen et al., 1993), mode frequency (frequency at the highest spectrum peak; Hagg, 1992), and half-width (spectral width at half maximum amplitude; Nargol et al., 1999). These indices represent single spectral values (Merletti and Lo Conte, 1997). Other proposed indices considering more than a single spectral value are obtained from changes in certain frequency ranges such as low frequency band (Dolan et al., 1995; Maïsetti et al., 2002), mid-frequency region (Lowery, 2000), and ratio of EMG power between high and low frequency bands (Moxham et al., 1982; Allison and Fujiwara, 2002). However, the reliability and sensitivity of these indices for fatigue assessment at LLEs have not been confirmed. Further, several of these indices have produced inconsistent results when applied in different studies (Kumar et al., 2001), suggesting that further research is necessary for obtaining improved EMG-based fatigue indices during contractions at LLEs.

5.1.3 Alternative EMG Processing Methods

Recently, non-linear concepts have been employed in deriving new EMG-based fatigue parameters, for example by using logarithmic transformation. Transformation of the EMG power spectral density (PSD) onto logarithmic axes seems to yield a relatively smooth shape (Figure 5.1). Yassierli and Nussbaum (2003) developed alternative methods on this basis, logarithmic power-frequency, and generated novel fatigue parameters by detecting changes in both lower and higher frequency components (Figure 5.2). However, they noted that the derived parameters produced a relatively higher variability in terms of time-dependent changes than MdPF or MnPF. A follow up investigation using this method is needed with a goal of determining more reliable EMG-based fatigue indices, more specifically for LLEs.

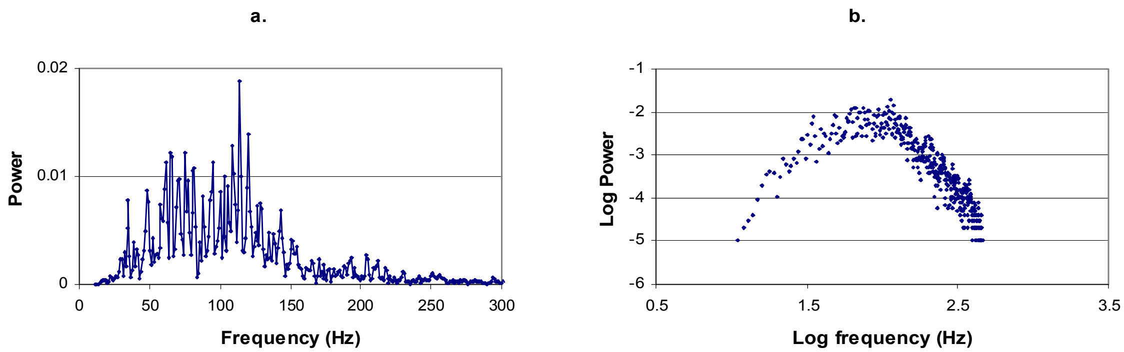

Figure 5.1 Sample of EMG power spectrum distribution a) 'normal' PSD, and b) PSD after transformed to logarithmic power frequency.

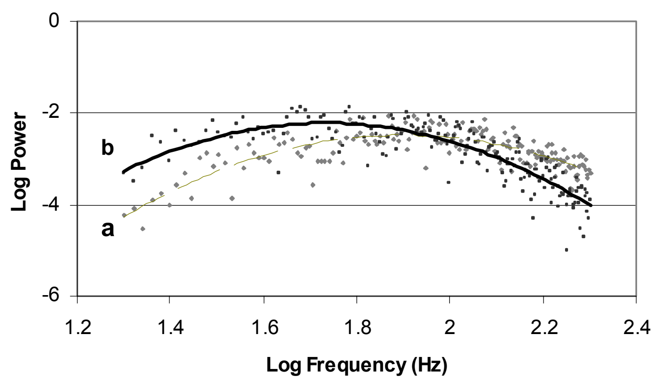

Figure 5.2 Changes on EMG spectral after transformation to logarithmic power-frequency, a) at the beginning, in comparison to b) a few seconds before exhaustion during a prolonged shoulder abduction.

Previous studies have suggested fractal characteristics of the EMG, as another non-linear method to process the EMG signal (e.g. Gupta et al., 1997). In general, fractal signals exhibit long-term correlation and/or "$1/f^{\alpha}$" spectra (Bruce, 2001). Both lower and higher frequency components of the EMG spectrum, after being transformed into a logarithmic function, seem to demonstrate a $1/f^{\alpha}$ pattern (Figures 5.1b and 5.2), indicating the possibility of fractal behavior. Self-similarity at different levels of magnification is one of main properties of fractal signals. As shown in Figure 5.3, a new EMG signal derived by averaging 10-consecutive points (having onetenth of the time resolution) appears similar to the original one. The fractal concept has been proposed as a measure of complexity and heterogeneity for a physiological system that has such self-similarity characteristics (Gitter et al., 1991; Schepers et al., 1992).

a. b.

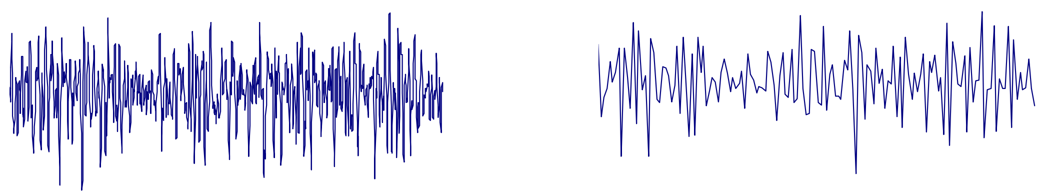

Figure 5.3 EMG signal taken from an isometric exertion shows similarity at different scales a) original signal, and b) averaged signal of 10-consecutive data points.

For general physiological signals, fractal characteristics can be quantified using a 'fractal dimension', which is a measure of the roughness or irregularity of a time series (Meyer and Stiedl, 2003). The fractal dimension of EMG signals has been found to be sensitive to the amount of force exerted, different loading conditions, and rates of flexion-extension (Anmuth et al., 1994; Gitter et al., 1991; Gupta et al. 1997). This dimension can be calculated using the method of Dispersional analysis, which appears to be more robust than other methods such as the Hurst analysis (Schepers et al., 1992; Bassingthwaighte and Raymond, 1995). In Dispersional analysis, the relative dispersion of a sample, an indication of signal variability, is determined by the ratio of the standard deviation to the mean over a range of successive time intervals. The fractal dimension is then computed from the slope of a linear relationship between the log of variance and the log of the data length. Nussbaum and Yassierli (2003) used the Dispersional analysis method to determine changes in fractal dimension during prolonged shoulder abductions. The changes in fractal dimension calculated from raw-EMG (filtered at ~80 to 200

Hz) was found to be more sensitive to fatigue and exertion level in comparison to those calculated from the whole frequency spectrum. This suggests that fractal characteristics of the EMG signals may be limited to a certain frequency range.

In addition to the fractal dimension, fractal scaling has been used to characterize fractal behavior of physiological signals. Fractal scaling provides a measure of complexity of physiological signals by quantifying fluctuation in the data using the method of Detrended fluctuation analysis (DFA; Iyengar et al., 1996). The idea is to analyze changes in a time-frame employing data variance (Varela et al., 2003). DFA has been used to identify age-related disruptions in heartbeat dynamics (Iyengar et al., 1996), and to recognize aging effects on temperature curve decreases (Varela et al., 2003). Alterations of the fractal scaling exponent have also been found to be associated with aging and Huntington's disease in human locomotion (Hausdorff et al., 1997). Further study appears necessary to determine the potential of using fractal dimension and fractal scaling for quantification of localized muscle fatigue.

5.1.4 Criteria for Comparison among EMG-Based Fatigue Indices

What are the criteria for a better EMG-based fatigue index? Unfortunately, authors have used widely different criteria for this evaluation. Ideally, the index should reflect muscle fatigue development within muscles, although the entire physiological fatigue process appears to be too complex to be described by a single index (Hägg, 1991). Thus, sensitivity to fatigue probably can be used as the first criterion of a better EMG index, as suggested by Merletti et al. (1991). Using this criterion, the EMG index would be expected to change following fatigue development, and the rates of change should differ depending on workload. This criterion was employed by Ravier et al. (2005), who compared the sensitivity of their proposed non-linear indices with the commonly used spectral indicators. In addition to sensitivity, another criterion for a better EMG index may be variability, more specifically in terms of the temporal consistency of the changes (e.g. do the data follow an underlying trend). For example, Nussbaum (2001) compared variability among rates of change on MnPF, MdPF, and RMS collected during overhead tasks. Dolan et al. (1995) used this criterion in support of their conclusion that changes in the EMG power spectrum at low frequencies provided for a better index of fatigue (i.e. less variability) than changes in median frequency during isometric back exertions.

Authors have also suggested that a better EMG index should be reliable or repeatable (e.g. Dolan et al., 1995). Reliability (a.k.a. repeatability or reproducibility) is defined as the ability to achieve similar results (e.g. rates of EMG change) on repeated trials (Larsson et al., 2003). However, only few studies have specifically addressed repeatability of fatigue measures during LLEs. In Rainoldi et al. (1999), repeatability for slopes of MnPF during biceps brachii contraction was reported to be lower at 10-30% MVC compared to repeatability at 50-70% MVC, based on intraclass correlation coefficient (ICC).

Since changes in EMG indices with respect to time seem to be linear and a significant correlation between rates of EMG change and endurance time has been reported (Hagberg, 1981; van Dieën et al., 1993), some investigators have employed initial subsets of EMG data to predict and extrapolate the trend of EMG data for the whole endurance test (van Dieën et al., 1993; Maïsetti et al., 2002). This ability, namely predictive ability, could also be considered as an additional criterion for a better EMG index. Predictive ability appears to be beneficial for task evaluations due to various reasons. First, participants can avoid the very strenuous parts of endurance test. Second, it minimizes the influence of motivational factors in voluntary exercises.

Third, it reduces the length of the test. Van Dieën et al. (1998) compared the predictive ability of MnPF over three different workloads: 25, 50, and 75% MVC. At 25% MVC, MnPF typically increased at the beginning of contraction period, resulting in poor predictive ability of a short initial data subset. At higher effort levels, time-dependent changes in MnPF tended to be more consistent (decreasing), producing better accuracy in predicting the trunk extensor endurance.

5.1.5 Purpose of the Study

The main objective of this study was to develop several alternative EMG-based fatigue indices for prolonged static contractions at LLEs, and further to investigate their utility in comparison to existing indices. Based on above discussions, it is clear that the existing EMG parameters such as RMS, MdPF or MnPF demonstrated a lack of utility (e.g., less sensitive to fatigue or less reliable) during sustained isometric contractions at LLEs. In this report, the term 'utility' was used to encompass all the aforementioned criteria: sensitivity, variability, predictive ability, and repeatability. It is worth noting that each of these criteria have been used separately in previous research, but never as a whole. Results of the present study were expected to provide improved EMG-based methods in muscle fatigue assessment for industrial tasks, many of which are predominantly characterized by low-level contractions.

5.2 METHODS

In the present study, two sets of data were analyzed: primary data and secondary data. The first set was obtained from a shoulder abduction experiment consisting of 3 sessions with 12 participants involved; the latter set was taken from a previous study (described in Chapter 2).

5.2.1 Primary Data

Fatigue development was evaluated under two workload conditions, 15 and 30% MVC, as representative of LLEs. To allow for examining repeatability, two replications were conducted at 15% MVC (Retest 15% MVC). The three experimental sessions were performed on separate days with at least 2 days rest. The arrangement of sessions was random across participants; the retest data was taken from the second performance of the 15% MVC condition. In general, procedures used in this experiment were similar to those employed for the shoulder abduction study reported in Chapter 2. The differences between studies were only related to the arm posture; in the present work the arm was abducted at 20º, instead of at horizontal. This posture was intended to minimize the effect of arm mass on contraction levels, since the shoulder effort to hold the right arm in a horizontal position is equal to about 10-15% MVC.

Participants

A total of 12 younger participants (6 males and 6 females) were recruited from the college community (age 18.1 ± 3.6 years old, height 171.1 ± 6.1 cm, mass 67.2 ± 6.2 kg). The participants had an exercise level of 3.5 ± 0.5 according to a self-reported assessment with a 5-point scale, where 1 corresponds to "never" and 5 means "everyday". All participants were right-handed. Participants had no previous shoulder injuries or disorders within the 12 months prior to the experiment. Informed consent, approved by the Virginia Tech Institutional Review Board, was obtained prior to the experiment.

Procedures

A practice session was provided at the beginning of the each session. Each participant was stabilized in a seated position at a dynamometer (Biodex^{TM} System 3 Pro Medical System, Shirley, New York, USA) with shoulder and waist strapped. This dynamometer has been reported to have good reliability for measuring torque and position (Drouin et al. 2004). The dynamometer's center of rotation was aligned with the estimated of the shoulder joint center. A padded strap was used to attach the participant's right arm at the elbow to the dynamometer attachment. The dynamometer attachment was set at 20º from a vertical position. Each participant was instructed to perform MVCs by maximally exerting their right arm in the frontal plane against the padding, while the left arm remained resting at the participant's side. Similar to the procedures for Chapter 2, each recorded MVC value was corrected for gravitational effects from the mass of the participant's arm and the dynamometer attachment. Three trials were conducted with two minutes of rest given between trials. If the torque exerted markedly increased (more than 10%) from the previous trials, an additional trial was carried out. The largest torque obtained was determined as the participant's MVC.

Following a 10-min rest, each participant performed a static endurance test at effort levels of either 15 or 30% MVC with postures consistent with those used in the MVCs. Torque feedback was displayed on a computer screen located in front of the participant, and each participant was instructed to maintain the exertion level within ±5% of the target torque until exhaustion. Immediately after cessation of the exercise, each participant was instructed to perform a post-fatigue MVC. Throughout the tests, efforts were made to ensure that participants were comfortable, and non-threatening verbal encouragement was provided. All postures and fixture configurations were recorded and maintained across days.

Dependent Measures

Dependent measures collected in this study included endurance time, rate of MVC decline, changes in ratings of perceived discomfort (rates of RPD), and modifications in EMG-based fatigue indices. Endurance time was determined as time during which torque was maintained within the target. Rate of MVC decline was computed as percentage reduction in muscular strength, normalized against individual endurance time. Ratings of perceived discomfort were collected throughout the endurance test using the Borg (1990) CR-10 scale for rating perceived exertion. The scale is continuous, ranging from 0-10 in which 0 means "no discomfort at all" and 10 corresponds to "extremely strong (almost maximal)" discomfort. The RPD values were collected every 30 seconds. Modifications in EMG fatigue indices were determined as rates of change (slopes).

To obtain EMG signals, a pair of Ag/AgCl electrodes was placed over the belly of the middle deltoid muscle (according to Hermens et al., 2000), with an inter-electrode distance of 2.5 cm. The skin was shaved, gently abraded, and then cleaned with rubbing alcohol. An inter-electrode resistance below 10kΩ was considered acceptable. The ground electrode was located on the clavicle. Raw signals, collected continuously during the exertions, were preamplified (x100) near the electrode sites, then hardware amplified and band-pass filtered at 10-500 Hz (Measurement System Inc., Ann Arbor, MI, USA). Raw signals were sampled at 2048 Hz. Root-mean-square (RMS) data were obtained using a 110-ms time constant, sampled at 128 Hz, and subsequently low-pass filtered using software (Butterworth, zero phase-lag, 4^{th} order, 3 Hz cut off). The standard processing methods described in Chapter 2 were applied to obtain initial values and slopes of RMS, MdPF, and MnPF. Briefly, for RMS, each 1-sec EMG RMS taken from the endurance tests was averaged and normalized (nRMS) against maximum RMS obtained during the MVC tests. For MdPF and MnPF, raw EMG were divided into 2-sec samples, and subsequently divided into three 1-sec overlapping windows (Luttmann et al., 1996). Hanning window and Fast-fourier transform (FFT) were applied to each sample window. A final power spectral density (PSD) was calculated as an average value across the three windows and was used to determine MnPF and MdPF. In addition to these standard processing methods, some alternative processing methods were applied to the same EMG signals; these methods are explained in section 5.2.3.

5.2.2 Secondary Data

EMG signals collected in Chapter 2 were further analyzed. Briefly, these signals were obtained from static endurance tests of arm abduction and torso extension. Only signals taken from efforts at 30% MVC were analyzed, since this level typically represents an upper limit for LLEs. In chapter 2, rates of change on RMS, MdPF, and MnPF had been obtained for these data. In addition to these indices, the data were processed using alternative methods, which are discussed in the next section.

5.2.3 EMG Processing Methods

The alternative processing methods included the Frequency-band Method, Logarithmic-power Frequency, Dispersional Analysis, DFA, and the Poisson-plot Method. For each processing method, temporal changes in the derived parameters were modeled using linear regression (justified based on data inspection); the resultant slopes of change were used as fatigue measures. The methods used to obtain each alternative index are described below.

5.2.3.1 Frequency-Band Method

Raw EMG were processed to obtain EMG power spectral density (PSD) using standard EMG processing methods (2-sec samples, 1-sec overlapping windows, Hanning window, FFT). The sum of the power within 10-45 Hz bandwidth (LF_{Band}), as done by Allison and Fujiwara (2002), was calculated as percentage of the total power (10-200 Hz). For each participant's data, changes in LF_{Band} over time were fitted into linear regression and the slope was used as a fatigue parameter.

5.2.3.2 Logarithmic Power Frequency

PSDs obtained by applying the standard methods to the raw EMG were transformed into a logarithmic function. Three new indices were derived as proposed by Yassierli and Nussbaum (2003): peak frequency (Peak), slope of lower frequency (LFslp), and slope of higher frequency (HFslp). Figure 5.4 illustrates the derivation of these parameters. The first parameter was defined as the frequency with highest log amplitude obtained from a polynomial curve fitted over the 10-200 Hz bandwidth. The last two parameters were derived from slopes of a linear regression fit with two frequency ranges: 10-45 Hz for the lower frequency slope and 90-150 Hz for the higher frequency slope. Although at first glance arbitrary, these selections were based on observation and preliminary data processing in which Peak ranged within 50-85 Hz and data within the ranges selected demonstrated a better linear trend (low variability). Fatigue measures were then determined based on slopes of changes in these three indices after the data were fitted into linear regression.

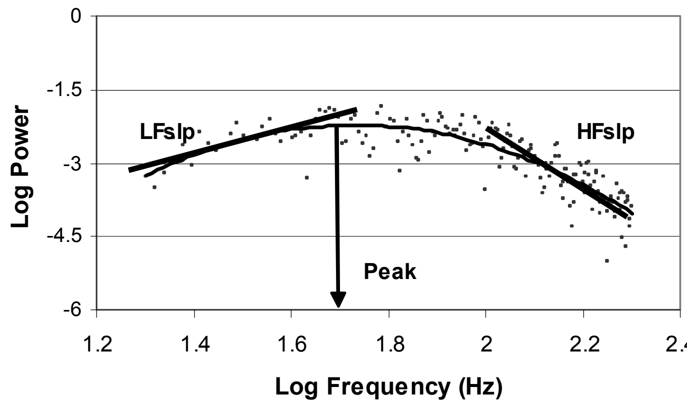

Figure 5.4 Peak, LFslp, and HFslp were derived using linear regression from a portion of frequency bandwidth (in this case, Peak=$10^{1.7}$=50.12)

5.2.3.3 Dispersional Analysis

Fractal dimension (D) was determined using Dispersional Analysis according to the following algorithm (Bassingthwaighte and Raymond, 1995; Chau, 2001):

1. Define the signal, a time series x(t) consisting of N observations.
2. Compute the mean of the signal $\bar{x}$.
3. Divide the signal into n groups composed of m data points (m x n = N).
4. Calculate the mean of the each group.
5. Determine the standard deviation of the group means, SD(m).
6. Compute the relative dispersion, RD(m) = SD(m)/ $\bar{x}$.
7. Repeat steps 3-6 with increasing group size.
8. Plot log RD(m) versus log m. From the slope (S), the fractal dimension can be obtained using D = 1-S.

Raw EMG were divided into 2-second samples (4096 points) with 1-second overlapping windows. For each window, D was computed using the above algorithm for lower and higher frequency components of the signals, namely D_{LF} and D_{HF} respectively. The signals were band-pass filtered using software (Butterworth, zero phase-lag, 4th order) into 10-F_{Peak} Hz and F_{Peak}-200 Hz, respectively intended to cover lower and higher frequency portions of the signal. F_{Peak} was determined as the Peak based on Logarithmic-Power frequency. Prior to applying Dispersional Analysis, the raw EMG signals were full-wave rectified (converted to absolute values). Appropriate group sizes (*m*) were determined based on the sampling rate and filtering used. Group sizes ranged from 10-100 and 5-50 points, respectively, for lower and higher frequency components. Rates of changes of D_{LF} and D_{HF} over time following a linear fit were employed as fatigue measures.

5.2.3.4 DFA

EMG signals obtained from both primary and secondary data were also processed using the DFA method to obtain fractal scaling. Detailed algorithms for this method can be found in Iyengar et al., (1996) or Varela et al., (2003), and are summarized as follows:

1. Define the signal, consisting of N observation of time series x(*t*).
2. Integrate the time series: $y(k) = \sum_{i=1}^{k} (x_i - \overline{x})$
3. Divide the integrated signal into segments of size *n*
4. Determine a regression line for each segment $y_n(k)$
5. Detrend the integrated time series by subtracting each value of $y_n(k)$
6. Calculate the average fluctuation of this integrated and detrended time series as:

$$F(n) = \sqrt{\frac{1}{N} \sum_{k=1}^{N} [y(k) - y_n(k)]^2} \tag{5.1}$$

7. Repeat steps 3-6 over different values of *n*.

The scaling exponent (α) was determined from a plot of log F(n) versus log n which shows a linear relationship (Figure 5.5).

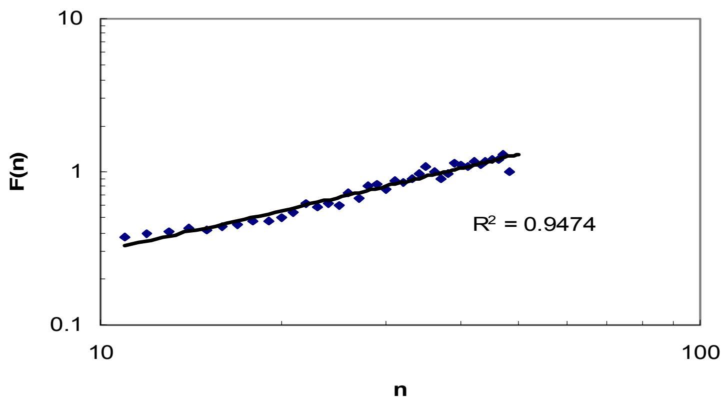

Figure 5.5 An example of F(n) vs. n in logarithmic scales

Similar to the Dispersional analysis method, EMG raw were divided into successive 2-sec windows (4096 points), with a 1-sec overlap between adjacent windows. For each window, α_{LF} and α_{HF} were computed to represent lower and higher frequency parts. The signals were band-pass filtered using software (Butterworth, zero phase-lag, 4th order) into 10-F_{Peak} and F_{Peak}-200 Hz, respectively for α_{LF} and α_{HF}. Correspondingly, n ranged from 10-50 and 5-30 points, chosen based on the sampling rate, filtering used, and data inspection. The temporal changes in α_{LF} and α_{HF} were fitted into linear regression, and the slopes were used as fatigue measures.

5.2.3.5 Poisson-plot Methods

As previously mentioned, changes in the shape of EMG PSD during a fatiguing exercise have received little attention. Assuming that all frequency components provide useful information towards characterizing fatigue development, quantifying changes in the shape of the PSD may be meaningful. The Poisson distribution is a discrete distribution that can be used for describing the distribution of an EMG PSD. This distribution is commonly used to model the number of random occurrences within a given time interval. For a sample of N, the expected frequencies are:

$$m_k = N\,p_\lambda(k) = N\,e^{-\lambda}\lambda^k/k! \qquad k = 0, 1, 2, \ldots \tag{5.2}$$

The parameter λ represents the average value. Hoaglin (1980) introduced a Poissonness plot, a graphical method to diagnose how well a set of data can be fitted using a Poisson model. Taking the logarithm on both sides of the equation, and by assuming some fixed values of λ, and that observed frequency (x_k) equals the expected frequency (m_k), equation 5.2 can be manipulated as follows:

$$\log(x_k) = N(\log) - \lambda + k\log(\lambda) - \log(k!) \quad \text{or}$$

$$\log(x_k) + \log(k!) = k\log(\lambda) + N(\log) - \lambda \tag{5.3}$$

Using this method, λ can be estimated as the slope of $\log(x_k)+\log(k!)$ versus k if a relatively linear relationship exists.

To apply this method to EMG signals, raw data were processed using the standard EMG processing method (2-sec samples, 1-sec overlapping windows, Hanning window, FFT). The resultant PSD was divided into segments of 10 Hz frequency bands, and the sum of the power within each band was computed (Figure 5.6). Formula 5.3 was then applied to obtain a Poissonness plot. Finally, λ was calculated from the slope (for example in Figure 5.7, $\lambda = 10^{0.854} = 7.14$). Rates of changes in λ with respect to time fitted into a linear regression were used as a fatigue measure.

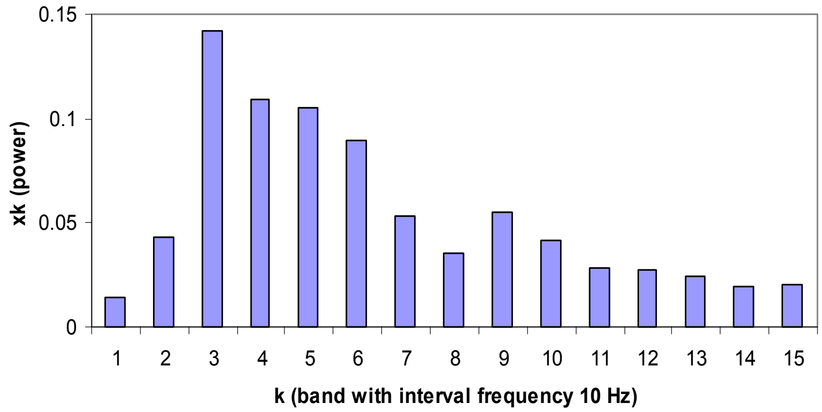

Figure 5.6 EMG power spectral density after grouping with a 10-Hz interval.

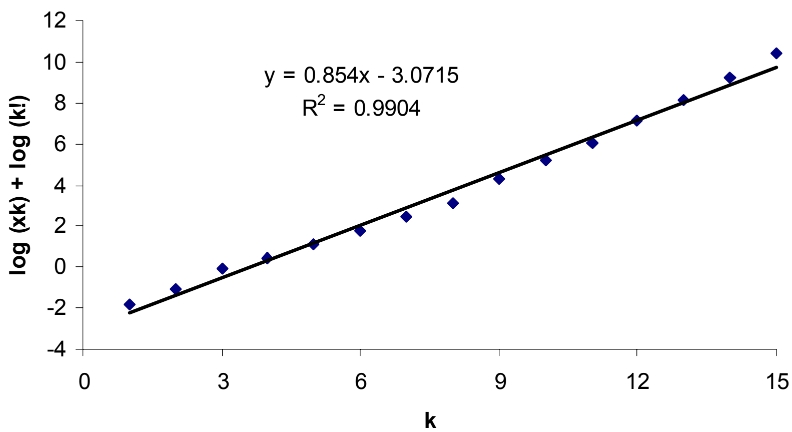

Figure 5.7 A Poissoness plot taken from subsets of dynamic EMG.

5.2.4 Analysis

The primary data were analyzed using procedures similar to those in Chapter 2. Independent variables were gender and effort level. Dependent variables were grouped into EMG and non-EMG measures since analyses of EMG signals were simultaneously conducted for primary and secondary data. Non-EMG measures included endurance time, rates of MVC decline, and changes in rating of perceived discomfort. The presence of main and interaction effects between independent variables on these measures collected during exercises at 15% and 30% MVC were tested via a two-way repeated measures analysis of variance (ANOVA).

Repeatability of non-EMG measures were determined by comparing data collected during exercises at 15% and Retest 15% MVC. Intra-class correlation coefficient (ICC), standard error of measurement (SEM), and coefficient of variation (CV) were employed as repeatability indices. These indiced are commonly used given their higher sensitivity to changes in means and standard deviation than Pearson's correlation coefficient (Sleivert and Wenger, 1994; Keller et al. 2001, Elfving et al., 1999). For all dependent variables, associations among the measures were tested using correlations. Significance for all tests was determined at $p<0.05$.

EMG data were categorized into 3 data sets: A, B, and C, respectively for primary data, secondary data of shoulder abduction, and secondary data of torso extension. The latter were collected bilaterally from the longissimus thoracis and multifidus muscles (C1 and C2, respectively). For each data set, the utility of the derived EMG parameters were compared based on sensitivity, variability, repeatability, and predictive ability using the following computation procedures.

1. Sensitivity to fatigue development was determined based on the proportion of data which produced significant linear changes over time (increasing or decreasing trend) across participants and trials. In addition, sensitivity to fatigue at different effort levels was also established by comparing slopes of change of EMG indices during A-15% and A-30% MVC. It is noteworthy that sensitivity or responsiveness can be defined as the ability to detect differences in effort level (Fagarasanu and Kumar, 2002), and is here represented by omega squared (ω^2) using Equation 5.4 (Keppel, 1991). This index, ranging from 0 to 1, has advantages over the significance level given by an F-test due to its independence from sample size.

$$\omega^2 = \frac{SS_A - (a-1)(MS_{S/A})}{SS_T + MS_{S/A}} = \frac{(a-1)(F-1)}{(a-1)(F-1) + (a)(n)} \tag{5.4}$$

where: a = levels of treatment,
n = number of subject assigned to each treatment level
Note that the F-value of interest corresponds to effort level obtained from ANOVA results.

2. Variability was computed as the residual (root-mean-square) error from linear regression. Since the units of measure should be comparable, two types of data analysis were conducted. First, data were normalized to intercept values, as conducted by Merletti et al. (1990) and Nussbaum (2001). Second, instead of using normalization, a transformation was done to yield centered and standardized data (Newsom et al., 2003). Centering allows data to have a midpoint at zero, while standardizing results in a standard deviation equal to one. Differences among indices on variability were tested using ANOVA; separate ANOVAs were conducted for each data set.
3. Repeatability of EMG slopes was determined based on exercises at 15% MVC and Retest 15% MVC (only available for data set A). Repeatability was represented by ICC and CV, which have relative units. Both parameters seem to be most appropriate than SEM, which has absolute unit.
4. Predictive ability was tested by comparing EMG slopes estimated from shorter fixed periods, calculated over half and one-fourth of the mean endurance time across participants, to slopes over the full exercise period. This ability was determined using two measures: ICC and correlation values (r) between predicted and actual slopes.

5.3 RESULTS

5.3.1 Non-EMG Data (Primary Data)

Male participants had significantly higher MVC (~43%) than females. Initial MVC was found to have high repeatability (ICC=0.95) with fairly small variability (SEM = 4.13 Nm and

CV= 7.44%). Mean endurance times were 7.8 and 2.4 min for efforts at 15 and 30% MVC, respectively. Gender effects on endurance time were not significant (p=0.17), but the gender x effort level interaction effect approached significance (p=0.09), with slightly longer endurance times for males at the higher effort level (30% MVC). As expected, effects of effort level were significant for endurance time, rates of MVC decline, and rates of RPD. For the latter two dependent measures, no significant effects of gender and gender x effort level were present (p>0.35). Rates of RPD ranged from 0.6 - 3.1/min and 1.7 - 6.0/min, for 15 and 30% MVC, respectively. Mean rates of MVC decline were 4.2 and 7.9%/min, for the lower and higher effort levels, respectively. Repeatability at 15% MVC was high for endurance times (ICC=0.86, SEM=0.99 min, CV=12.96%) and ratings of RPD (ICC=0.93, SEM=0.18/min, CV=13.27%), but seemed to be poor for rates of MVC decline (ICC=0.1, SEM=2.88%/min, CV=68.15%).

5.3.2 EMG Data

5.3.2.1 Example Data

Figure 5.8 demonstrates examples of temporal changes of several EMG parameters. The data were taken from data set A, from the same participant. MnPF, MdPF, Peak, LFslp, λ, and D_{HF} tended to decrease with fatigue, whereas RMS, LF_{Band}, α_{LF} demonstrated an opposing tendency.

(a)

(b)

(c)

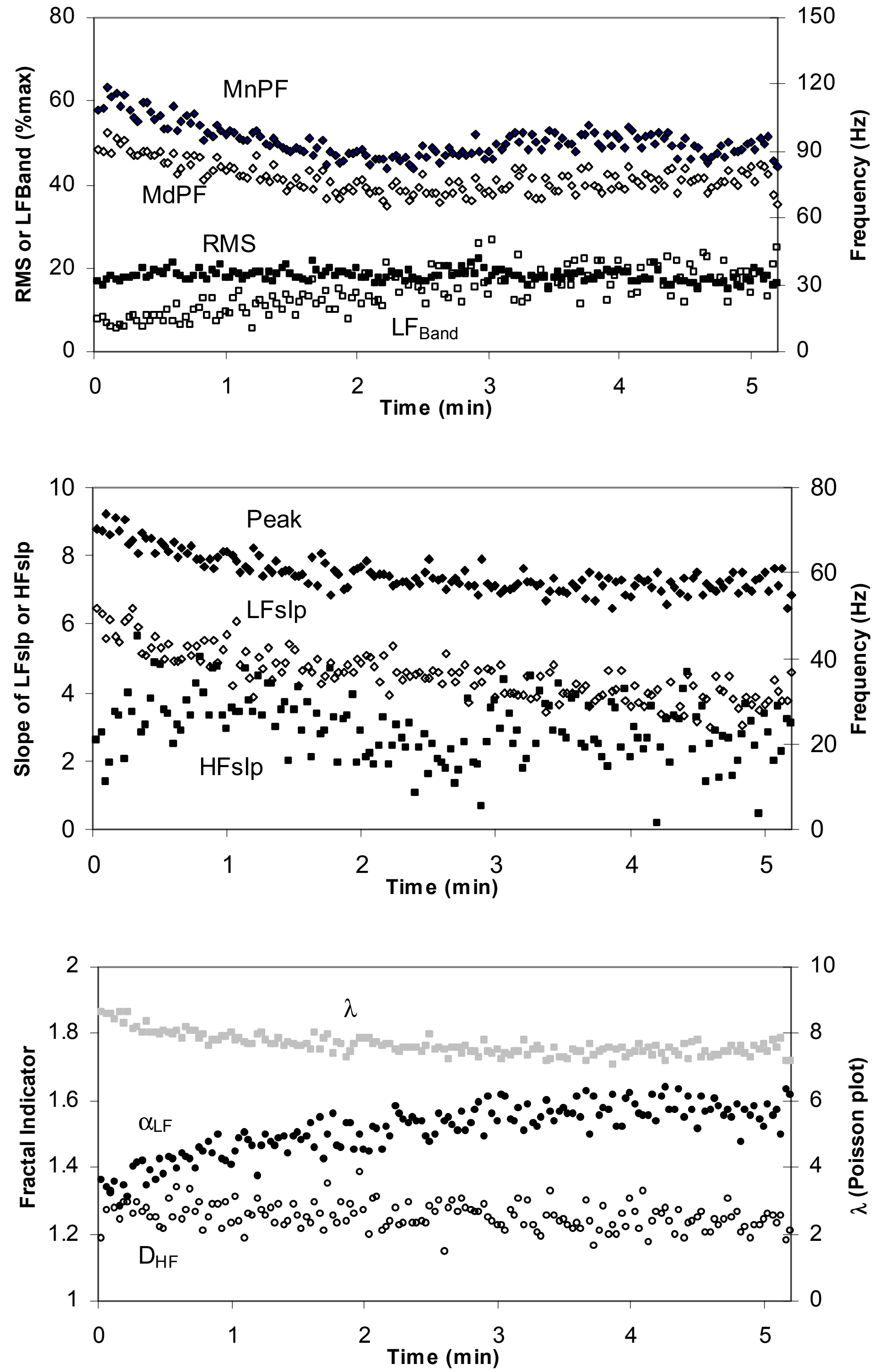

Figure 5.8 Sample time-dependent changes of different EMG indices taken from a same subject, in order from the top: MnPF, MdPF, RMS, LF_{Band}, Peak, LFslp, HFslp, λ, α_{LF}, and D_{HF}. The data are only illustrative, not necessarily represent all participants.

5.3.2.2 Sensitivity

Sensitivity to fatigue for the EMG indices varied depending on the muscles investigated (Table 5.1). For shoulder exercises (data sets A and B), significantly positive linear changes were observed for α_{LF} and LF_{Band} across all participants and trials. While decreasing trends were expected for MnPF and MdPF changes, some of the data showed non-significant changes or increasing trends. For data set C, the common spectral indicators (MnPF and MdPF) appeared to be the most sensitive parameter to fatigue, followed by λ, α_{HF}, Peak. Sensitivity of these indices was higher for the multifidus (C2) than the longissimus thoracis muscle (C1). The least sensitive index across data sets was RMS.

Table 5.1 Sensitivity of EMG indices.

	Proportion of time-dependent changes (-/0/+)				ω^2
Index	A	B	C1	C2	(A)
RMS	33/3/64	46/8/46	36/11/52	51/18/31	0
MnPF	94/6/0	98/0/2	88/10/2	92/8/0	0.71
MdPF	94/3/3	100/0/0	89/9/2	91/7/2	0.76
LF_{Band}	0/0/100	0/0/100	7/15/78	4/19/77	0.66
λ	94/3/3	100/0/0	83/11/5	88/9/3	0.64
Peak	97/3/0	100/0/0	79/16/5	85/11/3	0.64
LFslp	97/3/0	100/0/0	44/36/20	56/36/7	0.26
HFslp	31/11/58	13/2/85	8/42/50	4/42/54	0.03
D_{LF}	72/3/25	77/23/0	47/50/3	40/59/1	0.36
D_{HF}	73/8/19	77/23/0	39/54/7	42/48/10	0.22
α_{LF}	0/0/100	0/0/100	6/17/77	3/17/80	0.74
α_{HF}	3/6/91	0/0/100	6/9/84	4/9/86	0.64

Note: A, B, and C represent data sets (see 5.2.4). The proportion was determined as the percentages of slopes that were significantly positive (+), negative (-), or non-significant trend (0).

Sensitivity to fatigue at different workloads (ω^2) also differed between measures, with fractal scaling at low-frequency components (α_{LF}) the most sensitive index to fatigue among the new proposed indices. In comparison to the existing parameters, ω^2 was considerably higher than RMS, somewhat greater than MnPF, but slightly lower than MdPF.

5.3.2.3 Variability

Variability was significantly different among EMG indices (Table 5.2). Post-hoc analyses indicated that the two computational methods used (Method1 and Method2) produced different results (Table 5.3). For example, for data set A the first method demonstrated 4 clusters while the latter showed 6 clusters of indices. Variability was relatively higher for data set C, consistently observed across EMG indices. Overall, variability was comparable among α_{LF}, α_{HF}, MnPF, MdPF, and λ. Among the existing measures, MnPF had slightly and significantly lower variability than MdPF and RMS, respectively. Among the new indices, higher variability was observed for those derived from the logarithmic power frequency and dispersional analysis. For data set C, variability was significantly higher for C2 (multifidus) than C1 (longissimus muscle).

Table 5.2 Variability among EMG indices.

	Method1					Method2				
Index	A-15%	A-30%	B	C1	C2	A-15%	A-30%	B	C1	C2
RMS	0.18	0.17	0.11	0.08	0.07	0.72	0.68	0.72	0.80	0.83
MnPF	0.05	0.05	0.03	0.04	0.03	0.58	0.23	0.20	0.63	0.59
MdPF	0.07	0.05	0.04	0.06	0.05	0.66	0.20	0.26	0.73	0.68
LF_{Band}	0.34	0.59	0.42	0.21	0.21	0.61	0.22	0.36	0.82	0.83
λ	0.03	0.03	0.02	0.02	0.02	0.55	0.27	0.20	0.72	0.72
Peak	0.01	0.01	0.04	0.06	0.05	0.51	0.20	0.22	0.76	0.78
LFslp	0.08	0.08	0.07	0.10	0.07	0.53	0.59	0.70	0.93	0.93
HFslp	0.30	0.31	0.41	0.47	0.68	0.91	0.78	0.69	0.94	0.93
D_{LF}	0.04	0.04	0.04	0.02	0.02	0.96	0.85	0.86	0.96	0.97
D_{HF}	0.03	0.03	0.03	0.03	0.03	0.90	0.80	0.85	0.97	0.96
α_{LF}	0.03	0.02	0.03	0.03	0.04	0.43	0.16	0.23	0.78	0.82
α_{HF}	0.02	0.03	0.03	0.02	0.03	0.64	0.40	0.25	0.73	0.72

Note: Variability was calculated as residual error from normalized data against intercepts (Method1) and from centered-standardized data (Method2). For C1 and C2, the results were averaged values from bilateral multifidus and longissimus muscles, respectively.

Table 5.3 Results of post-hoc analysis on variability among EMG indices.

	Method1			Method2		
	A	B	C	A	B	C
HFslp	3	2	5	6 5	3	5
LFslp	1	2	4	4 3	3	4 3
RMS	2	1	3	5 4	3	4 3 2
LF_{Band}	4	1	4	3 2 1	2	4
D_{LF}	1	1	4	6	4	4 3
D_{HF}	1	1	4	6 5	4 1	4 3
Peak	1	1	3 2	2 1	1	3 2
MdPF	1	1	3 2	3 2 1	1	2 1
MnPF	1	1	1	2 1	1	1
λ	1	1	2 1	3 2 1	1	2 1
α_{HF}	1	1	2 1	3 2	1	2 1
α_{LF}	1	1	3 2	1	1	4 3 2

Note: A, B, and C represent different data sets. Indices that do not share a number are significantly different. Lower numbers corresponds to lower variability.

5.3.2.4 Repeatability

Higher ICC values were generally associated with lower CV (Table 5.4). ICC was highest for LF_{band}, followed by D_{HF} and LFslp. These three measures also showed relatively low CV values. The least repeatable measure was HFslp, followed by RMS and D_{LF}. Based on CV magnitudes, repeatability of α_{LF} was better than MnPF and MdPF.

Table 5.4 Repeatability of the rates of changes on different EMG measures (Data set A: isometric shoulder abduction at 15% MVC).

Measures	ICC	CV (%)
RMS	0.69	76.36
MnPF	0.87	44.88
MdPF	0.90	42.41
LF_{Band}	0.98	20.2
λ	0.83	47.74
Peak	0.87	38.86
LFslp	0.91	20.8
HFslp	0.70	171.9
D_{LF}	0.60	70.83
D_{HF}	0.93	31.49
α_{LF}	0.88	31.88
α_{HF}	0.76	67.98

5.3.2.5 Predictive Ability

Mean values at the 50% endurance times for data set A were 4.0 and 1.2 minutes for effort levels at 15 and 30% MVC, respectively. For data sets B and C the corresponding values were were 0.8 and 2.3 minutes, respectively. ICC magnitudes were in agreement with correlation values (r), as shown in Tables 5.5 and 5.6. Predictive ability using half of the estimated endurance time was typically better than that using one-fourth, for which ICC was substantially reduced on average almost 30, 50, and 50% respectively for data sets A, B, and C. Overall, RMS, LF_{Band} and D_{HF} demonstrated the highest predictive ability. The lowest values of ICC and r were observed for HFslp. LF_{Band} appeared to be the most predictive measure with ICC>0.36 for both prediction calculation methods across data sets. MdPF had a better predictive ability than MnPF.

Table 5.5 Predictive ability among EMG indices using data subsets of half the estimated endurance time.

	ICC					r				
Index	A-15%	A-30%	B	C1	C2	A-15%	A-30%	B	C1	C2
RMS	0.91	0.87	0.65	0.69	0.81	0.94	0.87	0.69	0.73	0.83
MnPF	0.68	0.22	0.72	0.83	0.88	0.83	*ns*	0.79	0.86	0.90
MdPF	0.64	0.45	0.73	0.76	0.85	0.83	0.66	0.80	0.84	0.89
LF_{Band}	0.84	0.57	0.78	0.83	0.89	0.93	0.57	0.78	0.85	0.89
λ	0.73	0.42	0.67	0.81	0.85	0.74	0.60	0.76	0.84	0.88
Peak	0.68	0.40	0.69	0.81	0.82	0.84	0.59	0.76	0.85	0.87
LFslp	0.72	0.52	0.60	0.75	0.76	0.86	0.61	0.70	0.77	0.80
HFslp	0.55	0.33	0.29	0.66	0.78	0.65	*ns*	0.38	0.78	0.82
D_{LF}	0.61	0.62	0.43	0.60	0.64	0.73	0.78	0.52	0.70	0.70
D_{HF}	0.89	0.63	0.29	0.59	0.75	0.89	0.75	0.35	0.67	0.78
α_{LF}	0.65	0.21	0.72	0.80	0.79	0.85	*ns*	0.82	0.84	0.84
α_{HF}	0.80	0.20	0.62	0.76	0.85	0.87	*ns*	0.72	0.84	0.88

Note: *ns* = non-significant correlation.

Table 5.6 Predictive ability among EMG measures using data subsets of one-fourth the estimated endurance time.

	ICC					r				
Index	A-15%	A-30%	B	C1	C2	A-15%	A-30%	B	C1	C2
RMS	0.86	0.49	0.25	0.28	0.38	0.89	0.75	0.34	0.35	0.43
MnPF	0.36	0.18	0.37	0.40	0.53	0.69	*ns*	0.56	0.52	0.70
MdPF	0.39	0.22	0.33	0.38	0.52	0.73	*ns*	0.51	0.50	0.68
LF_{Band}	0.83	0.55	0.51	0.36	0.40	0.89	0.58	0.52	0.40	0.49
λ	0.46	0.28	0.29	0.48	0.50	0.74	0.68	0.39	0.58	0.62
Peak	0.36	0.23	0.23	0.50	0.38	0.70	0.56	0.34	0.67	0.51
LFslp	0.29	0.17	0.18	0.27	0.38	0.63	*ns*	0.45	0.38	0.60
HFslp	0.23	0.10	0.13	0.33	0.33	*ns*	*ns*	*ns*	0.55	0.51
D_{LF}	0.27	0.33	0.05	0.09	0.21	0.65	0.81	*ns*	*ns*	0.38
D_{HF}	0.47	0.10	0.04	0.16	0.24	0.74	*ns*	*ns*	0.28	0.37
α_{LF}	0.50	0.15	0.32	0.53	0.39	0.69	*ns*	0.46	0.67	0.52
α_{HF}	0.49	0.23	0.26	0.40	0.45	0.81	*ns*	0.40	0.54	0.57

Note: *ns* = non-significant correlation.

5.3.3 Correlations among Measures

Coefficients of correlation (r) between EMG parameters, endurance time, and rates of MVC decline are compiled across data sets in Table 5.7. Note that correlations between the EMG parameters and rates of RPD were only available for data set A. For the remaining measures, correlation values were taken as an average over all data sets (A, B, and C). With the exception of HFslp and RMS, EMG-based fatigue indices had good correspondence with endurance time and rates of RPD (absolute r greater than ~0.5). Correlations for MnPF, MdPF, peak, λ, α_{LF}, and α_{HF} appeared to be comparable. Similar good correlations were observed between the majority of EMG indices and rates of muscle strength decline which has been considered as a 'gold-standard' indicator of localized fatigue; high correlations were observed for MnPF, MdPF, peak, λ, α_{LF}, and α_{HF}. All of these indices had an absolute r>0.93.

Table 5.7 Correlations among EMG parameters as well as with endurance time, rates of MVC decline, and rates of RPD.

	RMS	MnPF	MdPF	LF_{Band}	λ	Peak	LFslp	HFslp	D_{LF}	D_{HF}	α_{LF}	α_{HF}
Endurance time	*ns*	0.63	0.64	-0.50	0.62	0.64	0.48	-0.19	0.51	0.49	-0.62	-0.61
MVC decline	0.26	-0.66	-0.66	0.48	-0.66	-0.66	-0.43	0.26	-0.52	-0.51	0.64	0.66
RPD decline	0.42	-0.82	-0.84	0.76	-0.84	-0.84	-0.5	*ns*	-0.65	-0.71	0.84	0.83
RMS		-0.27	-0.21	0.26	-0.21	-0.20	*ns*	*ns*	*ns*	-0.33	0.20	0.20
MnPF			0.98	-0.77	0.94	0.95	0.60	-0.35	0.74	0.72	-0.94	-0.96
MdPF				-0.76	0.93	0.96	0.60	-0.33	0.78	0.69	-0.94	-0.95
LF_{Band}					-0.80	-0.72	-0.56	0.03	-0.61	-0.60	0.77	0.72
λ						0.97	0.61	-0.37	0.77	0.77	-0.97	-0.97
Peak							0.59	-0.38	0.77	0.71	-0.99	-0.98
LFslp								*ns*	0.51	0.46	-0.60	-0.54
HFslp									-0.21	-0.29	0.35	0.41
D_{LF}										0.59	-0.77	-0.71
D_{HF}											-0.68	-0.74
α_{LF}												0.96

Note: *r values were calculated only from data set A; *ns*=non significant.

5.4 DISCUSSION

The main purpose of this study was to investigate the utility of different EMG-based fatigue parameters for isometric contractions at LLEs. The parameters included existing (RMS, MnPF, MdPF, and LF_{band}) and alternative (Peak, LFslp, HFslp, λ, DLH, DHF, α_{LF}, and α_{HF}) indices. Note that although LF_{band} is less commonly used than the first three existing indices, it has been proposed by Bigland-Ritchie et al. (1981), Stulen and De Luca (1981) and Dolan et al. (1995). The utility of the indices was evaluated based on several criteria which have been employed separately in previous studies. Assuming that localized fatigue can be used as a valid indicator for risk of injury, a reliable EMG-based fatigue index is clearly needed for the design and evaluation of occupational tasks at LLEs (Hagberg, 1981; Nussbaum, 2001), since WMSDs may occur even for tasks at LLEs. Results of this study suggested that several alternative EMG indices can be used to monitor local fatigue development during LLEs. Among the alternative indices, parameters derived from fractal analysis were found to be more sensitive to fatigue and exhibited less variability in terms of their linear changes over time.

5.4.1 Existing Parameters

Overall, utility of MnPF and MdPF seemed to be similar. Rates of change for MnPF were typically slightly higher than for MdPF, probably due to PDS shifts being accompanied with changes in its shape (Merletti et al., 1992). MdPF was found to be more sensitive and more repeatable, but had higher variability than MnPF. This result is in agreement with previous reports such as Stulen and De Luca (1981) and Nussbaum (2001), and supports the suggestion that MdPF seems to be less affected by noise and signal aliasing than MnPF (De Luca, 1997).

Previous studies have shown that these common spectral parameters can be insensitive to fatigue during LLEs (e.g. Oberg et al., 1994; Thorn et al., 2002), and a phenomenon was observed here. Some trials showed non significant linear changes (i.e., initial decrease followed by an increasing trend) and even increasing linear changes. A reason for this, as postulated by Hägg and Ojok (1997), might be that these spectral indicators seem to be too simple to signal fatigue during LLEs. Note that motor unit behaviors while maintaining LLEs have been shown to be fairly complex, including combinations of decreasing firing rate, de-recruitment of motor units, motor unit rotation, and recruitment of larger motor units with larger action potentials (Kamo, 2002, Jensen et al., 2000). Increasing trends in MnPF or MdPF may be due to the effect of this last behavior (i.e., recruitment of larger motor units) which tends to be more dominant than the first three behaviors (Hägg and Ojok, 1997).

RMS seemed to be inferior compared to other parameters and demonstrated inconsistent performance as a fatigue index. This result is in agreement with previous findings which showed increasing, almost unchanged, and decreasing RMS data throughout contractions (e.g. Dimitrova and Dimitrov, 2003). The relatively high predictive ability associated with this index should be taken with caution since its sensitivity and variability were poor. Moreover, no significant correspondence was observed between this parameter and endurance time. LF_{Band} was found to be the best index in terms of repeatability and predictive ability, but its sensitivity was slightly lower than the sensitivity for MdPF, MnPF and $\alpha_{LF.}$ This finding agrees with Maisetti et al. (2002), who found that changes in a lower frequency band (6 and 30 Hz) computed from the first 15-30 second of knee extensor contractions had a better correspondence with endurance time than RMS, MnPF and MdPF. Note that these authors examined exercise at a higher effort level (50% MVC), suggesting that its utility may fairly general over a range of effort levels.

5.4.2 Fractal Analysis

A number of techniques have been proposed to characterize the fractal dimension of physiological signals (e.g. Schepers et al., 1992). If the EMG signal is assumed to be a roughly Gaussian random process (Stulen and De Luca, 1981), then dispersional analysis seems to be more appropriate in characterizing its fractal dimension (Caccia et al., 1997; Eke et al., 2000). However, this method computes the relative dispersion of the signals by determining the ratio of the standard deviation over the mean. Since raw EMG signals tend to have a zero mean, it was necessary to rectify the raw EMG signal to apply this method. As a result, the $1/f^{\alpha}$ pattern of power spectra might have been disturbed. To address this possible error-inducing artifact, alternative methods were also applied in our preliminary analyses (not reported here) by shifting the signal by 10 volts (i.e., mean = 10 volts) or by computing the relative dispersion as RD(m) = SD(m), instead of RD(m) = SD(m)/$\bar{x}$. However, both approaches produced similar results in which inconsistent plots of log RD(m) versus log m. were obtained.

In this work, another computation method (i.e. DFA) was employed. Our literature review produced no indication that this method had been applied in the past for EMG signals. Using DFA, the original raw EMG signal can be maintained. This method was initially developed by Iyengar et al. (1996) for heartbeat signals and was found to be able to identify age-related disruptions in heartbeat. Results of this study showed DFA produced a more sensitive, more repeatable, and less variable fatigue index than dispersional analysis.

Fractal characteristics of EMG signals have recently been documented (Nieminen and Takala, 1996; Gupta et al., 1997), and similar phenomena were observed in this study. In general, changes over time in fractal indicators (the fractal dimension and fractal scaling) were consistently found in this study during all fatiguing exercises. It has been suggested that a fractal dimension in the range 1.0 < D < 1.5 designates the presence of long-range positive correlations, while a dimension in the range 1.5 < D < 2.0 shows the existence of long-range negative correlations, and D=1.5 suggests random, uncorrelated signals (Bassingthwaighte et al., 1994). The observation of decreasing D associated with fatigue (starting value ~1.2-1.3) can be interpreted as evidence of increasing long-range positive correlations (vs. pure randomness). This can also be inferred as lowered dimensionality of the signals with fatigue as suggested by Nieminen and Takala (1996). They speculate that this may reflect a decreased ability of the physiological control system to perform motor control functions as fatigue increases. Pertaining to the fractal scaling, an α value of 0.5 indicates white noise, and an α of 1.5 corresponds to Brownian noise or a random walk (Iyengar et al., 1996). The increases in α associated with fatigue (starting value ~1.2-1.4) in this study can be explained as an increasing short term correlation in EMG signals, probably due to increasing motor unit synchronization.

Previous research has noted that fractal characteristics may be limited to a certain frequency range (Nussbaum and Yassierli, 2003). Their findings are supported by the profile of power spectral density; if the PSD is plotted using log-log axes, the higher frequency components show a $1/f^{\alpha}$ pattern, a common sign of fractal behavior. In the present work, both components of lower and higher frequency of signals were investigated since a similar, though inverse, pattern was also observed for the lower frequency components. The results of this study suggested that both lower and higher frequency components of the EMG signal demonstrated fractal characteristics. In fact, fractal scaling computed from the lower frequency components (α_{LF}) seemed to have better utility than that of higher frequency component (α_{HF}), particularly for the shoulder exercises. However, for the torso exercise the utility of α_{HF} was in general slightly better than α_{LF}. This may suggest a muscle-dependency for the frequency range showing fractal behavior since both muscles (shoulder and torso) are morphologically different (Manta et al. 1996; Mannion et al. 1997). This hypothesis remains to be investigated in future studies.

It is not clear why fractal indicator (i.e. α_{LF} and α_{HF}) resulted in relatively better sensitivity and variability than the existing EMG parameters, especially for the shoulder exercise. Several reasons are plausible. First, one advantage of this method is that it focuses on either lower or higher frequency signal components. Changes due to a decrease in conduction velocity and an increase in motor unit synchronization mainly affect the shape of PSD in the lower-frequency range, and recruitment or decruitment of motor units will, to some extent, influence power in middle to higher frequency band (Bigland-Ritchie et al. 1981; Hägg, 1991). These behaviors are expected to occur mainly during sustained contractions at LLEs. Thus, observations on either extreme of the frequency spectrum may be advantageous if one behavior can balance out others (Hägg and Ojok, 1997). Second, fractal methods such as DFA view signals using a completely different approach in comparison to 'normal' PSD measures, such as MnPF, MdPF, Peak, or LF_{Band}. Using fractals, complexity and dimensionality of the signals were analyzed, and theoretically, fatigue is associated with an increase in signal complexity and a reduction in signal dimensionality. Further study is warranted to investigate whether fractal approaches are similarly applicable for different exercise types (e.g. dynamic effort) and why the fractal indicators demonstrated lower utility for back muscles.

5.4.3 Logarithmic-Power Frequency and Poisson-plot

The methods of Logarithmic-power frequency and Poisson-plot were inspired from a common observation that fatigue is associated with compressions of the PSD toward lower frequencies (Lindstrom et al., 1977; Kranz et al., 1983; De Luca, 1984; Hägg, 1992). Both methods were expected to facilitate detecting changes in PSD shape using simple geometrical approaches. Although these methods appeared to be less sensitive than existing measures (e.g. MnPF and MdPF), they indeed provided additional information on changes in EMG signals due to fatigue.

Logarithmic-power frequency was initially proposed by Yassierli and Nussbaum (2003) with an expectation that the parameters resulting from this method would be less sensitive to noise, sampling rate, and frequency resolution than other existing fatigue indices. Among the three parameters proposed using this method, Peak appeared to have a better utility. Note that Peak may be more reliable than mode-frequency (discussed in Schweitzer et al. (1979) and De Luca (1984)), since Peak is derived from a "smooth" PSD in a log-log scale, while mode-frequency is more dependent on the nature of EMG signals which are stochastic and unsmooth near the peak value. Another advantage of this transformation is the possibility to obtain the best Peak estimation even for poor signal-to-noise ratios. In general, Peak was found to decrease with fatigue as PSD shifted toward lower frequencies. As suggested by Hägg (1991), an increase in Peak, although the overall trend was to decrease, may be a sign of newly recruited motor units. This phenomenon might not be detected by MnPF or MdPF.

As reported by Yassierli and Nussbaum (2003), LFslp and HFslp tend to have with high variability. Similar results were obtained in this study. Overall, LFslp seemed to have better utility than HFslp, but their overall utilities were less than those for MnPF and MdPF. Lower sensitivity may be due to complex motor unit behaviors associated with fatigue, as previously mentioned, that affect power in either of both the low or high frequency components. Although the data for both low- and high-frequency components can visually be fitted to a line, this approach needs further justification through further study. Also, recall that the chosen frequency bands were 10-45 and 90-150 Hz. Note that the cut-off ranges of the lower and higher frequency band varied across previous studies. For example, for lower frequency content: 5-30 Hz (Dolan et al. 1995), 20-40 Hz (Bigland-Ritchie et al., 1981; Haäg, 1991), 15-45 Hz (Allison and Fujiwara, 2002); for higher frequency: 130-238 Hz (Bigland-Ritchie et al., 1981) or above 95 Hz (Allison and Fujiwara, 2002). Physiological interpretations for each bandwidth are still unknown. It is worth noting that the same logarithmic-power frequency method used here was also recently documented by Ravier et al. (2005). In their study, and in line with the results in this investigation, the resulted parameters in comparison to MdPF were found to be less sensitive to fatigue, but more sensitive to force level.

The utility of the measure derived from the Poisson-plot method was found to be comparable to MnPF and MdPF. It was expected that the measure, λ, would be able to detect changes in PSD shape that can be associated with fatigue. However, as mentioned by Merletti et al. (1992), fatigue may result in compressions in PSD with or without an associated shape change. This may explain why the utility of λ was not superior. In further study (Chapter 6), this method will be examined for subsets of dynamic signals, leveraging the advantage that this method can be used for poor-resolution signals due to short sample windows since the signals were grouped into frequency intervals.

5.4.4 Limitations

A main limitation of this study is related to the model used to fit time-dependent changes in EMG parameters during the contractions. In this study, linear trend analysis was taken, an approach that has been suggested by others (van Dieën et al., 1998; Masuda et al., 1999; Nussbaum, 2001). However, fatigue-related data have also been reported to follow an exponential trend (Lindstrom et al., 1977; Merletti et al., 1991) or a sequence of rapid change and plateau (Gerdle and Fugl-Meyer, 1992). Here, a linear model greatly simplified the analysis and seemed to be applicable for all parameters. Another limitation pertains to the exercises chosen for this study, which were limited to shoulder abduction and torso extension. Furthermore, the secondary data were only available at 30% MVC which may be relatively higher than average low level efforts. Further study is needed that employs similar methods on lower exercise levels and for different muscles, considering that motor unit behaviors may vary depending on contraction level and muscle morphology.

5.4.5 Conclusions

Fatigue is a complex process and multiple EMG indices may be needed to characterize the development of local fatigue. Previous studies have suggested that motor units demonstrate complex behaviors while contracting the muscle to maintain a low-effort contraction. Since the commonly used spectral indicators (MnPF and MdPF) have been considered to be too simple to accurately represent these behaviors, several alternative indices were employed in this investigation and the utility among these indices was compared. In general, parameters derived from fractal analysis demonstrated high utility, suggesting a potential application of this method. Comparisons of MnPF and MdPF with the proposed measures indicated that LF_{Band} was better in terms of repeatability and predictive ability and α_{LF} was found to be more sensitive and less variable for shoulder exercise. Pertaining to the research methodology, this study suggested that similar utility criteria can be used in future studies that explore other alternative EMG-based fatigue indices.

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CHAPTER VI
EVALUATION OF EMG-BASED FATIGUE MEASURES DURING INTERMITTENT ISOKINETIC EFFORTS

Abstract

The present study evaluated alternative electromyographic (EMG)-based fatigue measures intended for dynamic efforts, using experimental data obtained in shoulder abduction and torso extension. These alternative measures were derived from the frequency band method, a Poisson-plot method, logarithmic-power frequency, and fractal scaling. Data from two prior experiments (Chapter 3 and Chapter 4) were employed, and both the alternative and traditional fatigue measures were analyzed for 'utility', in terms of sensitivity, variability, repeatability, and predictive ability. Each experiment consisted of four experimental conditions with 48 participants involved. Additional experiments were conducted that replicated a selected condition with 10 and 12 participants, respectively, for shoulder abduction and torso extension. For shoulder abduction, an index derived from a Poisson-plot had relatively higher sensitivity than other alternative indices, but its sensitivity was comparable to mean power frequency. Fractal indicators had higher predictive ability, and initial values of parameters derived from logarithmic-power frequency demonstrated higher repeatability. For torso extension, all measures demonstrated inconsistent changes with time. Overall, this study suggests the potential utility of alternative EMG-based fatigue measures for dynamic contractions, in particular shoulder abduction.

6.1 INTRODUCTION

Assessment of localized muscle fatigue, based on standard electromyographic (EMG) indices derived from changes in amplitude (root-mean-square; RMS) and mean or median frequency of spectral distribution, is well-established for controlled isometric contractions, particularly at moderate-high effort levels. Isometric contraction, however, is less common in daily and occupational activities, which are instead characterized by dynamic contractions. When applied to dynamic contractions, performance of such standard indices appears inconsistent, with reports showing both sensitivity and insensitivity to muscle fatigue. Similarly, conflicting results were obtained in our previous studies (Chapter 3 and 4) concerning these indices. Whereas findings from Chapter 3 suggested a possible use of the indices for dynamic shoulder abductions, results from Chapter 4 indicated that the indices were questionable if employed for dynamic torso extensions. Differences in the muscles investigated and experimental protocols may account for the inconsistency, however further investigation appears necessary to find more sensitive and reliable EMG-based fatigue indices for dynamic conditions.

6.1.1 EMG-based Evaluation of Dynamic Tasks

Several factors may impair or confound the sensitivity of EMG measures during dynamic contractions. During such contractions, muscles can slide relative to the skin and the detecting electrodes, resulting in geometric artifacts (Rainoldi et al., 2000). Changes in joint angle and muscle force during movement can alter magnitudes of mean and median power frequencies (Farina et al., 2001; Shankar et al., 1989). The latter effect (due to muscle force variation) would be more apparent if muscle fiber types with different conduction velocities are recruited during movement (MacIsaac et al., 2001). Additionally, speed of movement may also mask sensitivity of EMG measures (Duchene & Goubel, 1993; Hagg, 1992). Increased EMG signal magnitude could be obtained during tasks with higher speeds as a result of increased muscle activation (Laursen et al., 1998). However, a non-conclusive effect of movement speed on EMG-based indices has been reported. For example, Shankar et al. (1989) found a dependency of median power frequency (MdPF) on movement velocity; but on the contrary, Masuda et al. (2001) suggested no correlation between movement speed and EMG indices. The discrepancy may be due to interactions between changes in muscle length and velocity that influence the magnitude of EMG signals (Potvin, 1997).

Dynamic EMG can be separated into concentric and eccentric components. Concentric actions result from shortening muscle fibers, while eccentric contractions control or resiss a movement by lengthening muscle fibers. The magnitude of EMG signals obtained may differ between both contractions due to differences in active motor units and mean firing rates (Linnamo et al., 2003; Christensen et al., 1995). Depending on the joint angle and the movement speed, the effects of this effort type (concentric or eccentric) on EMG power spectra may be marginal (Potvin, 1997) or substantial (Komi et al., 2000). It has been suggested that muscle fiber conduction velocity may be slower during eccentric actions (Komi, et al., 2000) and mean power frequency (MnPF) seems to be greater in concentric effort (Moritani et al., 1988). This indicates that eccentric contraction may be associated with much less pronounced motor unit recruitment (Komi et al., 2000). In short, both effort types need to be differentiated in characterizing dynamic EMG signals.

6.1.2 EMG-Based Fatigue Assessment during Dynamic Efforts

Stationarity of signals has been considered an important issue in processing dynamic EMG. Standard processing methods have commonly used Fast-Fourier Transform (FFT) in yielding EMG spectral distribution, by decomposing temporal signals into frequency components without preserving information about time. Thus, the assumption used for this method is that the signal should be wide-sense stationary, with a constant mean and variance across segments (Bruce, 2001). In general, these stationary conditions can be satisfied for most constant-force isometric contraction for epochs of 0.5-2.0 sec (Roy et al., 1998).

It has been suggested that EMG signals obtained during dynamic contractions might violate stationarity assumptions (Knaflitz and Bonato, 1999). Changes in muscle force, muscle length, and the location of electrodes relative to the active muscle fibers may result in non-stationary signals, which exclude the use of FFT. For instance, Bazzy et al. (1986) showed an alteration in EMG frequency content due to a change in length of the bicep brachii muscle. To address this issue, new methods which require more complex algorithms such as Time-Frequency Analysis (TFA) and wavelet transform, have been proposed (Roy et al., 1998; Knaflitz and Bonato, 1999; Karlsson et al. 2001).

To enable implementation of standard EMG processing methods during dynamic exercise, investigators have incorporated repeated isometric tests at submaximal efforts within an ongoing exercise (so-called test contractions). Petrofsky (1979), for example, applied such an approach to assess fatigue development throughout a bicycle exercise. Though this could possibly disrupt the observed task and may contribute to additional fatigue (Roy et al., 1998), the approach meets the assumption of stationarity so that FFT can be applied easily for analysis.

In contrast to the approach of using isometric tests within a dynamic exercise, investigators have applied the standard EMG processing methods to dynamic EMG. Potvin and Bent (1997) showed that subsets of dynamic EMG (250 ms epochs) could be assumed to be at least weakly stationary. This argument was based on non-parametric tests conducted by Shankar et al. (1989). No effects of dynamic contractions on EMG spectral distribution were observed when the efforts were performed at low speed and low-force levels, based on comparisons between intramuscular and surface EMG methods (Christensen et al., 1995). The results of the latter study also suggested that standard methods may still be applicable for evaluating occupational tasks, though another limiting factor with this approach is that short data subsets of dynamic EMG would result in poor spectral resolution (Karlsson, et al., 2003). Interestingly, a recent study (Beck et al., 2005) demonstrated that the standard processing method (i.e. FFT) and the wavelet transform provided similar information on rates of change during isometric and dynamic contractions of the biceps brachii.

While controversy exists pertaining to the applicability of the standard processing methods for dynamic EMG, researchers have recently analyzed physiological signals (such as heart rate variability) using non-linear methods (Akay, 2001). This approach is inspired by the observation that such biosignals seem to be complex and high-dimensional. Therefore, linear statistical measures (such as mean or median value) may not adequately describe system complexity (Meyer and Stiedl, 2003). In dealing with dynamic signals, a number of non-linear methods have been proposed including detrended fluctuation analysis (DFA), approximate entropy, Lyapunov exponent, and recurrence plot analysis (Akay, 2001; Vaillancourt and Newel, 2002). DFA, for example, provides a measure of complexity by quantifying self-similar properties of non-stationary data where data fluctuation can be characterized by a (fractal) scaling exponent (Iyengar, et al., 1996). The study in Chapter 5 explored the possibility of using such a method to monitor fatigue development, though it was limited to low levels of isometric effort. This method may be applicable for dynamic EMG, and further study is needed with a goal of finding a more sensitive EMG-based fatigue measure for dynamic conditions.

6.1.3 Repeatability of EMG Measures

In addition to sensitivity, repeatability has been used as a criterion for determining the utility of EMG-based fatigue measures. Repeatability or reproducibility is defined as the ability to achieve similar results on repeated tests, or the degree to which an instrument provides consistent measures under the same experimental conditions (Larsson et al., 2003). Most studies on repeatability have used the test-retest method for estimating the variability of measurements in repeated trials. Authors have employed several indices, such as Intra-Class Correlation of coefficient (ICC) which is considered to be more sensitive to the changes in means and standard deviation than Pearson's correlation coefficient (Sleivert & Wenger 1994; Keller et al. 2001).

A number of studies have been conducted to investigate the repeatability of the standard EMG-based fatigue measures during isometric exercises. Excellent repeatability for initial values and slopes of MdPF was reported during isometric knee extensions (Kollmitzer et al., 1999). Good repeatability for MdPF was also documented for isometric tests of lower back muscles (Dedering et al., 2000). In contrast, Falla et al. (2002) reported poor repeatability for slopes of MnPF during isometric cervical flexion, while better repeatability was found for the initial values of MnPF. The results of Falla et al. (2002) are consistent with work done by Rainoldi et al. (1999) who investigated the repeatability of isometric contractions of the biceps brachii muscle. In general, EMG root-mean-square (RMS) has lower repeatability than MdPF (Kollmitzer et al., 1999; Larivière et al. 2002).

Evidence on the repeatability of standard EMG-based fatigue measures during dynamic exercise is more limited. Fair to good repeatability for RMS was obtained during isokinetic maximum voluntary contraction (MVC) tests of lower extremity movement (Sleivert and Wenger, 1994). Good repeatability for RMS and MnPF was obtained from 100 repeated MVC's of knee extensions (Larsson et al., 2003). During repetitive lifting, good repeatability of EMG measures of fatigue was reported for the lower back and vastus lateralis muscles (Ebenbichler et al., 2002). Different results, though, were obtained by Hager (2003) during an overhead tapping task. He found low repeatability for slopes of RMS, MnPF, and MdPF, but the intercept showed high repeatability (ICC ≥ 0.6). Note that analysis of repeatability seems to be critical for dynamic contractions due to the possibility of the aforementioned artifacts. Therefore, a better EMG-based fatigue measure should be sensitive as well as repeatable.

6.1.4 Purpose of the Study

The main purpose of this study was to develop alternative EMG measures that were sensitive and repeatable in monitoring fatigue during dynamic efforts. In addition to these two criteria, variability and predictive ability were included for utility comparisons that have similarly been applied in Chapter 5. As previously mentioned, there are conflicting arguments concerning appropriate EMG-based fatigue measures for dynamic contractions. It was expected that results of this study could provide an alternative approach in analyzing dynamic EMG, and contribute to improving EMG-based fatigue assessment methods for industrial tasks.

6.2 METHODS

6.2.1 Secondary Analysis of Existing Data

EMG data obtained from two previous studies (dynamic shoulder abductions, Chapter 3 and dynamic torso extensions, Chapter 4) were further analyzed. The data were recorded from repeated cyclic efforts under four different conditions. For shoulder abduction, the conditions were composed of 30%/10s, 30%/20s, 40%/10s, 40%/20s, representing combinations of effort level and cycle duration. For torso extension, similar combinations were used of 30%/15s, 30%/30s, 40%/15s, 40%/30s. Recall that the recorded EMG data were grouped as static and dynamic EMG. The latter, which are the focus in this investigation, were obtained from the last exertion cycle (prior to the sub-maximal static contractions) from each effort set. Within this cycle, a window was identified (using positional data) near the maximum joint angle.

6.2.2 Repeatability Test

In addition to analyzing secondary data, a test-retest study was conducted for determining the repeatability of the indices. A total of 10 and 12 participants were asked to repeat dynamic shoulder abductions and torso extension, respectively, at the lower effort level and the shorter cycle duration (30%/10s for shoulder and 30%/15s for torso). Based on a pilot study, this workload produced the lowest variability in endurance time. A minimum of 2 days and a maximum of 14 days rest were allowed between experimental sessions as recommended by Larivière et al. (2002). The same experimental procedures described in Chapter 3 and 4 were applied. In brief, the procedures consisted of pre-fatigue MVCs, endurance testing, and post-fatigue MVC. During the endurance test, each participant performed intermittent-isokinetic efforts either to their limit of endurance or up to one hour, whichever came first.

For EMG signal acquisition, a pair of Ag/AgCl electrodes (inter-electrode distance of 2.5 cm) placed over the muscle belly of the middle deltoid of the right arm during shoulder abductions (Hermens et al., 2000). During torso extensions, EMG signals were recorded bilaterally from the multifidus muscle at the L4/L5 level (Biedermann et al., 1990) and from the longissimus thoracis muscle at the L1 level (Larivière et al., 2002). For both types of exercise, electrode locations were located based on a record noted from the prior session to ensure consistent placement in subsequent experimental sessions. Prior to data collection, the skin was shaved, gently abraded, and cleaned with rubbing alcohol, with an inter-electrode resistance less than 10kΩ considered acceptable. The clavicle or the C7 vertebral process was used for grounding.

Raw signals were preamplified (x100) near the electrode sites, then hardware (Measurement Systems Inc., Ann Arbor, MI, USA) amplified and band pass filtered between 10-

500 Hz. Raw signals were sampled at 2048 Hz. Root-mean-square (RMS) data were obtained using a 110-ms time constant, sampled at 128 Hz, and subsequently low-pass filtered using a software-based Butterworth filter (zero phase-lag, 4^{th} order, 3 Hz cut off). Although EMG signals were collected continuously throughout contractions, only windows of dynamic EMG were processed. The windows were taken using similar procedures to those provided in Chapters 3 and 4.

6.2.3 EMG Processing Methods

The obtained windows of dynamic EMG were separated into Eccentric (Ecc), Concentric (Conc), and All (Eccentric + Concentric). The effects of concentric and eccentric actions were considered due to the possibility of variations in the activation patterns that can influence recorded signals (Linnamo et al. 2003; Komi et al., 2000). RMS EMG within these windows were averaged and expressed as normalized values against maximum RMS values from MVC trials ($nRMS_{Ecc}$, $nRMS_{Conc}$, and $nRMS_{All}$). Raw EMG were processed using Hanning window, FFTs, and zero padding to determine median ($MdPF_{Ecc}$, $MdPF_{Conc}$, and $MdPF_{All}$) and mean frequencies ($MnPF_{Ecc}$, $MnPF_{Conc}$, and $MnPF_{All}$). Temporal changes of these indices were used as fatigue measures.

Along with the standard EMG parameters, different processing methods were applied to the same windows to derive alternative EMG indices. The methods included Frequency–band method, Logarithmic Power Frequency, DFA (detrended fluctuation analysis), Poisson-plot, as previously described in Chapter 5. A brief explanation of each processing method is provided below.

Frequency-Band Method

From the EMG PSDs, the sums of the power within 10-45 Hz bandwidth (based on Allison and Fujiwara, 2002) were determined as LF_{Band}. This parameter was computed for each type of data (Eccentric, Concentric, All) to obtained $LF_{Band\text{-}Ecc}$, $LF_{Band\text{-}Conc}$, $LF_{Band\text{-}All}$ respectively. Changes in these indices over time were separately fitted into linear regression and the slopes were used as fatigue parameters.

Logarithmic Power Frequency

The same PSDs were transformed using a logarithmic representation. Three new indices were derived as proposed by Yassierli and Nussbaum (2003): peak frequency ($Peak_{Ecc}$, $Peak_{Conc}$, $Peak_{All}$), slope of lower frequency ($LFslp_{Ecc}$, $LFslp_{Conc}$, $LFslp_{All}$), and slope of higher frequency ($HFslp_{Ecc}$, $HFslp_{Conc}$, $HFslp_{All}$), in order corresponding to windows of Eccentric, Concentric, and All. Recall that the first index (Peak) was defined as the frequency with highest log amplitude obtained from a polynomial curve fitted over the 10-200 Hz bandwidth. The last two parameters were derived from the slopes of linear regression fit models for lower and higher frequency contents with frequency ranges: 10-45 Hz for the lower frequency and 90-150 Hz for the higher frequency. Based on preliminary data processing, these frequency bands demonstrated linear trends and seemed to have low variability. Fatigue measures were then determined based on slopes of changes in these three indices with respect to exercise time.

Detrended Fluctuation Analysis

For each data window, the following DFA algorithm was applied according to Iyengar et al., (1996) and Varela et al., (2003).

1. Define the signal, consisting of N observation of time series x(t).
2. Integrate the time series: $y(k) = \sum_{i=1}^{k} (x_i - \overline{x})$
3. Divide the integrated signal into segments of size n
4. Determine a regression line for each segment $y_n(k)$
5. Detrend the integrated time series by subtracting each value of $y_n(k)$
6. Calculate the average fluctuation of this integrated and detrended time series as:

$$F(n) = \sqrt{\frac{1}{N} \sum_{k=1}^{N} [y(k) - y_n(k)]^2} \quad (6.1)$$

7. Repeat steps 3-6 over different values of n.

The scaling exponent (α) was determined from a plot of log F(n) versus log n which showed a linear relationship. As a result, $\alpha_{LF\text{-}Ecc}$, $\alpha_{LF\text{-}Conc}$, $\alpha_{LF\text{-}All}$ and $\alpha_{HF\text{-}Ecc}$, $\alpha_{HF\text{-}Conc}$, $\alpha_{HF\text{-}All}$ were obtained to represent lower and higher frequency parts respectively for Eccentric, Concentric, and All. The signals were band-pass filtered using software (Butterworth, zero phase-lag, 4th order) into 10-F_{Peak} for lower frequency and F_{Peak}-200 Hz for higher frequency. F_{Peak} was obtained from logarithmic power frequency. Correspondingly for the lower and higher frequency parts, n ranged from 10-50 and 5-30 points, chosen based on the sampling rate, filtering used, and data inspection. The temporal changes in the indices were fitted using linear regression, and the slopes were used as fatigue measures.

Poisson-plot Methods

For each data type, the PSDs were divided into segments of 10 Hz frequency bands, and the sums of the power within each band were computed. Formula 6.2 (Hoaglin, 1980) was applied to estimate λ from the slope of $\log(x_k)+\log(k!)$ versus k assuming that a linear relationship exists, where λ represents the average value of a sample of N and x_k denotes the observed frequency.

$$\log(x_k) + \log(k!) = k\log(\lambda) + N(\log) - \lambda \quad \text{for } k=0,1,2,\ldots \qquad (6.2)$$

Then, λ_{Ecc}, λ_{Conc}, and λ_{All} were calculated from the slopes of each data type. Rates of changes in each λ with respect to time fitted using linear regression and used as the final set of fatigue measures.

6.2.4 Analysis

The utility of the derived EMG parameters (the standard and the alternative indices) was compared based on sensitivity, variability, repeatability, and predictive ability as follows.

1. Sensitivity was established based on the proportion of data which produced significant linear changes over time (increasing or decreasing trend) across participants and trials. In addition, sensitivity to fatigue at different effort levels was determined by comparing rates change of EMG indices (slopes from linear regression) between effort levels at 30% and 40% MVC. This sensitivity was represented by ω^2 according to Keppel (1991).
2. Variability was computed as the residual (root-mean-square) error from linear regression. This analysis was conducted if indices typically showed linear changes throughout contractions. Similar to the approaches used in Chapter 5, two types of data analysis were conducted. First (Method1), data were normalized to the intercept (Merletti et al., 1990; Nussbaum, 2001). Second (Method2), instead of using normalization, a transformation was done to yield centered and standardized data (Newsom et al., 2003), resulting in a midpoint at zero and a standard deviation of one. Differences between indices in variability were tested using separate ANOVAs for each data set.

3. Repeatability of initial values and rates of change were determined using ICC. The following formula was used to calculate ICC (Shrout and Fleiss, 1979; Denegar and Ball, 1993):

$$ICC(2,1) = \frac{BMS - EMS}{BMS + (k-1)EMS + k(TMS - EMS)/n} \tag{6.3}$$

where: BMS = between-subject mean square
EMS = error mean square
TMS = trial mean square
k = number of trials
n = number of subjects

4. Predictive ability was tested by comparing EMG slopes estimated from shorter fixed periods, specifically half and one-fourth of the mean endurance time across participants, to slopes obtained over the full exercise period. Two measures quantified predictive ability: ICC and correlation values (r) between predicted and actual slopes. The analysis was conducted only for trials at the 30% effort level, as it yielded relatively longer endurance times than 40%. Similar to variability, this criterion was applied only if linear changes were observed over time.

6.3 RESULTS

6.3.1 Dynamic Shoulder Abduction

6.3.1.1 Sensitivity

Table 6.1 compiles the proportion of temporal indices demonstrating linear changes for each data type during shoulder abduction. MnPF, MdPF, Peak, and λ demonstrated a typical decreasing trend with fatigue, whereas RMS, LF_{Band}, and α showed an opposite tendency. A qualitative comparison indicated that subsets of dynamic EMG from All (All EMG) typically resulted in higher sensitivity than Concentric EMG or Eccentric EMG, except for RMS and

LF_{Band}. For these two latter measures, Eccentric EMG seemed to be more sensitive to fatigue than Concentric EMG. For All EMG, MnPF, λ, and LF_{Band} appeared to be among the most sensitive indices to fatigue, with more than 70% of trials having significantly linear trends (either increasing or decreasing). The least sensitive index was LFslp.

Table 6.1 Sensitivity of EMG indices for shoulder abduction.

	Proportion of time-dependent changes (-/0/+)					ω^2	
	30%/10s	30%/20s	40%/10s	40%/20s	Average	10s	20s
RMS_{Conc}	21/19/60	50/33/17	23/23/33	44/23/33	34/24/41	0.05	0
RMS_{Ecc}	10/17/73	19/27/54	2/6/92	2/10/88	8/15/77	0.06	0
RMS_{All}	17/21/63	27/40/33	8/23/69	19/35/46	18/30/53	0.05	0
$MnPF_{Conc}$	75/17/8	75/17/8	77/19/4	69/29/2	74/20/6	0.13	0.12
$MnPF_{Ecc}$	79/15/6	81/15/4	69/27/4	71/25/4	75/20/5	0.10	0.15
$MnPF_{All}$	88/10/2	79/17/4	75/19/6	73/21/6	79/16/5	0.11	0.13
$MdPF_{Conc}$	58/35/6	69/27/4	71/27/2	60/35/4	65/31/4	0.11	0.12
$MdPF_{Ecc}$	65/33/2	63/33/4	58/33/8	63/33/4	62/33/5	0.08	0.16
$MdPF_{All}$	81/19/0	75/21/4	71/23/6	67/29/4	73/23/4	0.08	0.12
$LF_{Band-Conc}$	21/27/52	10/44/46	13/46/42	10/52/38	14/42/44	0.08	0.02
$LF_{Band-Ecc}$	0/23/77	0/29/71	2/19/79	0/23/77	1/23/76	0.09	0.08
$LF_{Band-All}$	6/17/77	0/31/69	6/21/73	2/33/65	4/25/71	0.14	0.08
λ_{Conc}	63/35/2	69/25/6	75/21/4	67/31/2	68/28/4	0.11	0.13
λ_{Ecc}	65/29/6	65/29/6	60/33/6	73/21/6	66/28/6	0.06	0.19
λ_{All}	77/21/2	69/29/2	63/31/6	77/19/4	71/25/4	0.08	0.17
$Peak_{Conc}$	54/40/6	65/33/2	56/42/2	56/40/4	58/39/4	0.05	0.07
$Peak_{Ecc}$	56/33/10	60/38/2	52/42/6	65/29/6	58/36/6	0.03	0.11
$Peak_{All}$	73/23/4	71/27/2	65/33/2	69/29/2	69/28/3	0.04	0.11
$LFslp_{Conc}$	13/52/35	10/69/21	6/65/29	8/69/23	9/64/27	0	0
$LFslp_{Ecc}$	33/56/10	25/73/2	17/73/10	17/73/10	23/69/8	0	0
$LFslp_{All}$	35/58/6	40/52/8	23/65/13	19/71/10	29/62/9	0	0
$HFslp_{Conc}$	56/40/4	35/58/6	40/50/10	27/63/10	40/52/8	0	0
$HFslp_{Ecc}$	44/50/6	33/63/4	35/54/10	33/56/10	36/56/8	0	0.08
$HFslp_{All}$	75/25/0	46/50/4	52/44/4	48/44/8	55/41/4	0	0.03
$\alpha_{LF-Conc}$	20/23/57	7/26/67	9/36/55	6/38/56	10/31/59	0.08	0.04
α_{LF-Ecc}	16/31/53	4/40/56	13/31/56	6/35/58	10/34/56	0.08	0.12
α_{LF-All}	4/25/71	6/25/69	0/35/65	4/25/71	4/27/69	0.10	0.10
$\alpha_{HF-Conc}$	32/3/66	19/12/69	23/20/57	4/42/54	19/20/61	0.09	0.06
α_{HF-Ecc}	32/3/65	19/21/60	16/21/63	4/31/65	18/19/63	0.04	0.16
α_{HF-All}	9/20/72	11/22/67	6/23/70	4/27/69	8/23/69	0.08	0.12

Note: The proportion was determined as the percentages of slopes that were significantly positive (+), negative (-), or non-significant trend (0).

6.3.1.2 Variability

For both computational methods (Method1 and Method2), variability was significantly different among EMG indices (Table 6.2). For Method1, variability was lowest for $\alpha_{HF\text{-}All}$ and λ_{All}, while for Method2, the lowest variability was associated with $MnPF_{All}$ followed by RMS_{Ecc} and $\lambda_{All.}$ Overall, variability was lower for All than Concentric or Eccentric. Comparing the last two data types, Concentric had slightly lower variability if computed using Method1, but contradictory results were obtained for Method2. Across data types, variability among MnPF, MdPF, and RMS seemed to be comparable for Method1, but the former was slightly lower than RMS and was significantly better than MdPF based on Method2. Among the alternative indices, the lowest variability was produced by α_{HF} and λ.

Table 6.2 Variability of EMG indices for shoulder abduction.

	Method1					Method2				
	30%/10s	30%/20s	40%/10s	40%/20s	Average	30%/10s	30%/20s	40%/10s	40%/20s	Average
RMS_{Conc}	0.14	0.12	0.14	0.11	0.13	0.79	0.83	0.86	0.86	0.83
RMS_{Ecc}	0.17	0.17	0.15	0.13	0.15	0.70	0.78	0.68	0.74	0.73
RMS_{All}	0.12	0.11	0.12	0.09	0.11	0.72	0.80	0.76	0.83	0.78
$MnPF_{Conc}$	0.08	0.07	0.07	0.07	0.07	0.84	0.79	0.80	0.71	0.79
$MnPF_{Ecc}$	0.07	0.06	0.06	0.06	0.06	0.86	0.78	0.78	0.68	0.78
$MnPF_{All}$	0.05	0.05	0.05	0.05	0.05	0.79	0.73	0.72	0.61	0.71
$MdPF_{Conc}$	0.11	0.09	0.10	0.09	0.10	0.91	0.87	0.88	0.79	0.86
$MdPF_{Ecc}$	0.10	0.09	0.09	0.09	0.09	0.92	0.88	0.86	0.77	0.86
$MdPF_{All}$	0.08	0.07	0.07	0.07	0.07	0.88	0.81	0.82	0.70	0.80
$LF_{Band\text{-}Conc}$	0.73	0.72	0.71	0.17	0.58	0.94	0.91	0.91	0.87	0.91
$LF_{Band\text{-}Ecc}$	0.81	1.17	1.08	0.70	0.94	0.89	0.83	0.82	0.75	0.82
$LF_{Band\text{-}All}$	0.59	0.73	0.79	0.77	0.72	0.87	0.80	0.81	0.72	0.80
λ_{Conc}	0.05	0.05	0.05	0.05	0.05	0.88	0.84	0.84	0.77	0.83
λ_{Ecc}	0.05	0.04	0.04	0.04	0.04	0.91	0.87	0.85	0.71	0.83
λ_{All}	0.03	0.03	0.03	0.03	0.03	0.85	0.79	0.79	0.66	0.77
$Peak_{Conc}$	0.21	0.14	0.14	0.10	0.14	0.93	0.89	0.87	0.81	0.87
$Peak_{Ecc}$	0.20	0.14	0.15	0.09	0.14	0.94	0.90	0.87	0.76	0.87
$Peak_{All}$	0.11	0.09	0.09	0.07	0.09	0.89	0.84	0.83	0.72	0.82
$LFslp_{Conc}$	0.27	0.27	0.27	0.27	0.27	0.97	0.96	0.96	0.90	0.95
$LFslp_{Ecc}$	0.29	0.28	0.28	0.26	0.28	0.98	0.97	0.96	0.91	0.95
$LFslp_{All}$	0.22	0.21	0.22	0.21	0.22	0.98	0.95	0.95	0.91	0.95
$HFslp_{Conc}$	1.08	0.94	0.85	0.45	0.83	0.96	0.94	0.94	0.89	0.93
$HFslp_{Ecc}$	1.56	0.74	0.77	0.64	0.93	0.97	0.95	0.95	0.87	0.93
$HFslp_{All}$	1.35	0.57	0.58	0.63	0.78	0.95	0.94	0.92	0.84	0.91
$\alpha_{LF\text{-}Conc}$	0.08	0.07	0.07	0.06	0.07	0.92	0.89	0.87	0.81	0.87
$\alpha_{LF\text{-}Ecc}$	0.08	0.07	0.07	0.05	0.07	0.93	0.90	0.88	0.80	0.88
$\alpha_{LF\text{-}All}$	0.06	0.05	0.06	0.04	0.05	0.89	0.85	0.84	0.75	0.83
$\alpha_{HF\text{-}Conc}$	0.04	0.04	0.03	0.03	0.03	0.92	0.85	0.87	0.78	0.86
$\alpha_{HF\text{-}Ecc}$	0.03	0.03	0.03	0.03	0.03	0.92	0.86	0.84	0.76	0.85
$\alpha_{HF\text{-}All}$	0.03	0.02	0.03	0.02	0.02	0.89	0.82	0.82	0.72	0.81

Note: Method1 was calculated as residual error from normalized data against intercepts, Method2 was computed from centered-standardized data.

6.3.1.3 Repeatability

Repeatability of rates of fatigue (represented by ICC of slope) was highest for MdPF, followed by MdPF and LF_{Band} (Table 6.3). RMS seemed to be the least repeatable measure. In general, All EMG resulted in higher repeatability than Concentric or Eccentric EMG. No trends were noted between the latter two types of data. Measures derived from Logarithmic power frequency appeared to have higher repeatability of initial values (ICC intercept) than others. Similar to results for repeatability of slope, a conclusion could not be made concerning whether a particular data type was associated with higher repeatability.

Table 6.3 Repeatability of EMG indices (measured using ICC) for shoulder abduction.

	Slope			Intercept		
Index	Conc	Ecc	All	Conc	Ecc	All
RMS	0	0	0	0.37	0.56	0.61
MnPF	0.54	0.67	0.82	0.49	0.38	0.46
MdPF	0.60	0.82	0.89	0.58	0.27	0.44
LF_{Band}	0.52	0.61	0.71	0.30	0.25	0.23
λ	0.61	0.14	0.56	0.73	0.49	0.59
Peak	0.07	0	0.28	0.04	0.22	0.41
LFslp	0.10	0.11	0.04	0.69	0.86	0.43
HFslp	0.71	0	0.45	0.71	0.49	0.79
α_{LF}	0.59	0.41	0.25	0.45	0.43	0.39
α_{HF}	0	0.34	0.70	0.61	0.38	0.60

6.3.1.4 Predictive Ability

Mean values of the 50% endurance times were 27, 24, 15, and 12 minutes respectively for conditions of 30%/10s, 30%/20s, 40%/10s, 40%/20s. ICC magnitudes appeared to be in agreement with correlation values (*r*), as shown in Table 6.4 and 6.5. For predictions based on half the estimated endurance time, effect of cycle duration (10s or 20 s) seemed to be minor since both resulted in comparable ICC and *r* values. In contrast, for predictions based on one-fourth the estimated endurance time, longer cycle duration typically produced better predictive ability. While predictions using half the estimated endurance time appeared to be reasonable for all indices, several parameters were not predictive at one-fourth the estimated endurance time as demonstrated by non-significant *r* and low ICC values. Overall, the predictive ability based on one-fourth the estimated endurance time appeared to be comparable across EMG indices and across data type. Across cycle durations, $\alpha_{HF\text{-}Ecc}$ seemed to be the most predictive measure with ICC>0.30. Eccentric EMG typically resulted in a better predictive ability than Concentric EMG.

Table 6.4 Predictive ability of EMG indices using data subsets of half the estimated endurance time for shoulder abduction.

	ICC		r	
Index	30%/10s	30%/20s	30%/10s	30%/20s
RMS_{Conc}	0.81	0.76	0.84	0.78
RMS_{Ecc}	0.75	0.87	0.79	0.92
RMS_{All}	0.75	0.84	0.79	0.87
$MnPF_{Conc}$	0.75	0.44	0.81	0.50
$MnPF_{Ecc}$	0.73	0.69	0.74	0.71
$MnPF_{All}$	0.75	0.76	0.76	0.62
$MdPF_{Conc}$	0.78	0.52	0.80	0.62
$MdPF_{Ecc}$	0.73	0.74	0.74	0.78
$MdPF_{All}$	0.76	0.65	0.78	0.73
$LF_{Band\text{-}Conc}$	0.46	0.95	0.61	0.96
$LF_{Band\text{-}Ecc}$	0.58	0.99	0.77	0.98
$LF_{Band\text{-}All}$	0.49	0.99	0.78	0.98
λ_{Conc}	0.78	0.72	0.79	0.76
λ_{Ecc}	0.70	0.67	0.72	0.74
λ_{All}	0.71	0.71	0.72	0.77
$Peak_{Conc}$	0.48	0.40	0.54	0.45
$Peak_{Ecc}$	0.75	0.03	0.79	ns
$Peak_{All}$	0.92	0.64	0.92	0.71
$LFslp_{Conc}$	0.50	0.28	0.56	0.33
$LFslp_{Ecc}$	0.38	0.30	0.45	0.47
$LFslp_{All}$	0.34	0.57	0.50	0.66
$HFslp_{Conc}$	0.62	0.68	0.66	0.71
$HFslp_{Ecc}$	0.38	0.56	0.45	0.60
$HFslp_{All}$	0.68	0.79	0.72	0.58
$\alpha_{LF\text{-}Conc}$	0.77	0.60	0.80	0.71
$\alpha_{LF\text{-}Ecc}$	0.91	0.67	0.89	0.72
$\alpha_{LF\text{-}All}$	0.75	0.74	0.76	0.71
$\alpha_{HF\text{-}Conc}$	0.74	0.91	0.81	0.89
$\alpha_{HF\text{-}Ecc}$	0.73	0.72	0.76	0.76
$\alpha_{HF\text{-}All}$	0.76	0.70	0.79	0.78

Note: *ns* = non-significant correlation.

Table 6.5 Predictive ability of EMG indices using data subsets of one-fourth the estimated endurance time for shoulder abduction.

	ICC		r	
Index	30%/10s	30%/20s	30%/10s	30%/20s
RMS_{Conc}	0.28	0.15	0.34	ns
RMS_{Ecc}	0.50	0.26	0.56	0.39
RMS_{All}	0.38	0.17	0.42	ns
$MnPF_{Conc}$	0.20	0.08	0.30	ns
$MnPF_{Ecc}$	0.26	0.18	0.44	0.35
$MnPF_{All}$	0.25	0.12	0.41	ns
$MdPF_{Conc}$	0.08	0.13	ns	0.31
$MdPF_{Ecc}$	0.15	0.28	ns	0.52
$MdPF_{All}$	0.11	0.24	ns	0.45
$LF_{Band-Conc}$	0.07	0.73	ns	0.87
$LF_{Band-Ecc}$	0.28	0.88	0.77	0.93
$LF_{Band-All}$	0.00	0.83	ns	0.88
λ_{Conc}	0.20	0.38	0.34	0.70
λ_{Ecc}	0.11	0.25	ns	0.54
λ_{All}	0.18	0.37	0.29	0.71
$Peak_{Conc}$	0.35	0.18	ns	0.39
$Peak_{Ecc}$	0.16	0.03	ns	ns
$Peak_{All}$	0.26	0.21	0.32	0.37
$LFslp_{Conc}$	0.06	0.02	ns	ns
$LFslp_{Ecc}$	0.11	0.09	0.30	0.39
$LFslp_{All}$	0.09	0.09	0.34	ns
$HFslp_{Conc}$	0.17	0.25	0.38	0.50
$HFslp_{Ecc}$	0.14	0.09	0.36	ns
$HFslp_{All}$	0.13	0.13	0.29	0.32
$\alpha_{LF-Conc}$	0.38	0.28	0.55	0.54
α_{LF-Ecc}	0.18	0.22	ns	0.44
α_{LF-All}	0.00	0.25	ns	0.45
$\alpha_{HF-Conc}$	0.20	0.52	ns	0.69
α_{HF-Ecc}	0.33	0.39	0.52	0.56
α_{HF-All}	0.17	0.20	ns	0.47

Note: *ns* = non-significant correlation.

6.3.2 Dynamic Torso Extension

6.3.2.1 Sensitivity

In contrast to shoulder abduction data, less than 50% of trials across EMG indices demonstrated significant linear trends during torso abduction (Table 6.6). Results for subsets of Concentric, Eccentric, and All seemed to be comparable. RMS produced the highest proportion of linear trends, followed by LF_{Band} and α_{HF}. Since linear fits were not consistently obtained, rates of EMG changes were determined based on means of the three cycles taken from the beginning and end of each trial. As suggested by results in Chapter 4, rates of change were computed as the average values of the left and right muscles for the longissimus and multifidus. Based on ω^2 across muscles investigated, λ seemed to be more sensitive to fatigue than MnPF and MdPF (Table 6.7). The most sensitive index was RMS and λ for the longissimus and multifidus, respectively. The least sensitive index across data sets was HFslp.

Table 6.6 Sensitivity of EMG indices in terms of proportion of time-dependent changes (-/0/+) for torso extension.

Index	Conc	Ecc	All
RMS	21/39/40	17/37/47	22/31/47
MnPF	28/55/16	24/55/21	29/51/20
MdPF	21/65/14	18/63/19	22/59/19
LF_{Band}	15/53/32	13/57/30	15/50/35
λ	19/64/17	15/63/22	19/59/21
Peak	13/68/19	12/68/21	15/64/21
LFslp	18/69/13	18/72/10	19/71/9
HFslp	12/77/11	14/76/11	13/77/10
α_{LF}	21/66/13	15/66/19	15/62/23
α_{HF}	13/68/19	13/57/30	15/50/35

Note: The proportion was determined as the percentages of slopes that were significantly positive (+), negative (-), or non-significant trend (0).

Table 6.7 Sensitivity of EMG indices measured using ω^2 for torso extension.

Muscle	Index	15 sec Conc	15 sec Ecc	15 sec All	30 sec Conc	30 sec Ecc	30 sec All	Average
Longissimus	RMS	0.10	0.07	0.09	0.18	0.16	0.14	0.12
	MnPF	0.07	0.04	0.02	0.05	0	0.04	0.04
	MdPF	0.11	0.04	0.04	0.04	0.04	0.05	0.05
	LF_{Band}	0.01	0.05	0.02	0.14	0.04	0.07	0.06
	λ	0.06	0.10	0.09	0.05	0.04	0.07	0.07
	Peak	0.02	0.01	0.04	0.09	0.02	0.09	0.05
	LFslp	0.11	0.02	0.08	0.18	0.08	0.04	0.09
	HFslp	0.04	0	0	0.00	0	0.03	0.01
	α_{LF}	0.04	0.01	0	0.11	0	0.14	0.05
	α_{HF}	0.02	0.09	0.03	0.06	0.03	0.03	0.05
Multifidus	RMS	0.07	0	0	0.03	0.00	0.04	0.02
	MnPF	0.08	0.04	0.07	0.06	0.09	0.03	0.06
	MdPF	0.11	0.07	0.05	0.02	0.07	0.01	0.06
	LF_{Band}	0.01	0.06	0.04	0.01	0.02	0.00	0.02
	λ	0.08	0.08	0.04	0.07	0.12	0.05	0.07
	Peak	0.07	0	0.01	0.02	0.14	0.04	0.05
	LFslp	0.05	0.02	0.04	0.08	0.03	0.06	0.05
	HFslp	0.00	0.04	0.03	0.11	0	0.03	0.03
	α_{LF}	0.00	0.02	0.11	0.05	0.01	0	0.03
	α_{HF}	0.04	0.03	0.12	0.02	0	0.01	0.04

6.3.2.2 Repeatability

Repeatability was better for intercepts compared to rates of change (Table 6.8). Subsets of All EMG were typically associated with higher repeatability than Concentric or Eccentric. For All EMG, ICCs computed for rates of change were highest for RMS and MdPF, respectively, for the longissimus and multifidus muscles. For initial value, ICC values of the three data types seemed to comparable with the most repeatable measure shown by λ.

Table 6.8 Repeatability based on ICC values for torso extension.

Muscle	Index	Rate of change			Initial value		
		Conc	Ecc	All	Conc	Ecc	All
Longissimus	RMS	-0.04	-0.06	0.62	0.62	0.31	0.53
	MnPF	-0.11	0.24	-0.18	0.71	0.60	0.71
	MdPF	-0.35	-0.30	0.03	0.77	0.56	0.74
	LF_{Band}	0.07	-0.01	0.03	0.11	0.20	0.11
	λ	0.00	0.00	-0.06	0.86	0.79	0.94
	Peak	0.04	0.26	0.28	0.81	0.66	0.89
	LFslp	0.00	0.25	0.21	0.44	0.85	0.46
	HFslp	0.51	-0.08	0.11	0.42	0.77	0.36
	α_{LF}	0.07	-0.09	-0.17	-0.19	-0.19	-0.16
	α_{HF}	0.00	0.07	0.03	-0.12	-0.12	-0.11
Multifidus	RMS	-0.04	0.00	-0.01	0.43	0.47	0.50
	MnPF	-0.08	0.12	0.34	0.84	0.84	0.81
	MdPF	-0.30	0.07	0.43	0.76	0.64	0.64
	LF_{Band}	0.08	-0.07	-0.01	0.03	0.05	0.04
	λ	-0.53	0.40	0.35	0.78	0.88	0.85
	Peak	-0.39	-0.05	0.25	0.83	0.67	0.86
	LFslp	0.13	0.24	-0.09	0.20	0.08	0.17
	HFslp	0.13	-0.01	0.40	0.58	0.27	0.65
	α_{LF}	-0.02	0.56	-0.05	-0.09	-0.09	-0.10
	α_{HF}	-0.09	-0.02	-0.01	-0.11	-0.20	-0.11

6.4 DISCUSSION

The present investigation sought to develop alternative EMG-based fatigue parameters for dynamic contractions. The EMG parameters investigated included existing (RMS, MnPF, MdPF, and LF_{band}) and alternative (Peak, LFslp, HFslp, λ, α_{LF}, and α_{HF}) indices. Utility of the indices was mainly evaluated based on sensitivity and repeatability which have been employed separately in previous studies. Variability and predictive ability were used as additional utility criteria if changes in indices demonstrated linear trends. This work is motivated by the fact that fatigue assessment using EMG is preferable to other methods such as endurance time or changes in muscle strength, as previously discussed. Earlier reports suggest that reliability and sensitivity of EMG is questionable during dynamic contraction, a condition which is more relevant to daily and occupational tasks. If reliable and sensitive, EMG-based fatigue assessment would have great practical value for task design, physical therapy, and clinical application (e.g. assessment of diaphragm fatigue; De Luca, 1984).

6.4.1 Shoulder Abduction vs. Torso Extension

Results of this study suggest that the utility of EMG indices during dynamic efforts is dependent on the muscle investigated and effort type. Changes in EMG indices demonstrated relatively consistent linear trends during shoulder abduction. However, similar results were not obtained during torso extension, which showed less than 50% linear trends. Inconsistency observed during the latter exercise could be explained by a combination of the complexity of the low back musculature and dynamic exertions that lead to artifacts. Note that during isometric efforts, EMG may be still sensitive to fatigue as reported in Chapter 2. During dynamic contractions, however, the signals recorded could be confounded by cross-talk (Stokes et al., 2003) and geometric artifacts (Rainoldi et al., 2000) resulting from skin movement. Load sharing among low back and hip muscles reducting from fatigue also may have contributed to the inconsistency of EMG indices during the exertions. As discussed in Chapter 4, earlier studies have shown that dynamic torso exercise may lead to a de-recruitment of the lumbar muscle and then trigger activation of other muscles including hip muscles (Kankaanpää et al., 1998; Leinonen et al., 2000, and Clark et al. 2002).

Contrary to torso extension, effects of load sharing which may randomly occur are probably marginal during shoulder abduction. The middle deltoid, which was selected in the investigation, seems to be dominantly active throughout the exercise. As a result, the recorded signals presumably provide information on changes in muscle states from non-fatigue to fatigue conditions. These changes include an increase in motor unit recruitment and firing rate, and a decline in the muscle fiber conduction velocity.

6.4.2 Utility of EMG Indices

Findings from this study suggest that the utility of an EMG index varied depending on the criteria used. Note that the concept of multi-indices of fatigue has previously been suggested by Merletti et al. (1991), given that fatigue is a complex physiological process. For repetitive shoulder abduction, MnPF and λ (derived from the Poisson-plot) were associated with higher sensitivity and lower variability. Both EMG parameters characterize and were derived from the EMG spectral distribution. It was expected that λ would be more sensitive to fatigue, considering that this index is computed based on an average of consecutive interval bandwidth and then fitted using a Poisson distribution. Both methods may minimize effects of noise and signal aliasing. During shoulder abduction, noise in the signals recorded may be marginal due to superficial location of the middle deltoid muscle. This may explain comparable results in terms of sensitivity between λ and MnPF. Though associations between fatigue and a Poisson model is not established, a discrete Poisson distribution seems to be graphically well-fitted with a common distribution of EMG spectra. Indeed, this is the main reason of adopting this index as an alternative EMG-based fatigue measure, though further development on this method is still needed.

For shoulder abduction, predictive ability appeared to be higher for fractal indicators (α_{LF} and α_{LF}). Note that the predictive ability is affected by linearity of index changes with time (van

Dieën et al., 1998). This implies that the fractal indicators demonstrated more consistent linear changes with fatigue than other indices. The consistency was supported by relatively lower variability associated with these parameters. With regard to other criteria, this measure showed moderate sensitivity and repeatability. This may be due to short sample windows of dynamic EMG. Note that these parameters were derived from either lower or higher frequency components using a band-pass filter and these procedures may be less reliable for such dynamic windows. Though in general the utility of these indicators were not superior, the results confirm the possibility of using fractal methods for EMG-based muscle fatigue assessment. The observation of increasing trends associated with fatigue can be interpreted as evidence of increasing complexity, probably due to increasing motor unit synchronization. It is worth noting that several alternative fractal algorithms are available (e.g. Raymond & Bassingthwaghte, 1999; Akay, 2001), and it is still of interest in further study to employ such non-linear methods for dynamic EMG.

For shoulder abduction, repeatability of initial values was highest for LFslp and HFslp, while MdPF seemed to have higher repeatability in fatigue rate (slope of change over time). A higher repeatability associated with LFslp and HFslp supports a report by Ravier et al. (2005) who found that parameters derived from a Logarithmic-power frequency were more sensitive to force level, but less sensitive to fatigue. As initially reported by Yassierli and Nussbaum (2003), both indices tend to have high variability. This may be due to complex muscle behaviors associated with fatigue that require a more complex model than a simple regression fit of either lower of higher frequency components of the PSD.

Repeatability of fatigue rate may be more meaningful for application. Results from the shoulder abduction studies indicated that MdPF seems to be the most repeatable or reliable parameter. Earlier studies have compared reliability and sensitivity between MnPF and MdPF (Stulen & De Luca, 1981; Nussbaum, 2001). Higher reliability of MdPF supports the suggestion that this index seems to be less affected by noise and signal aliasing than MnPF (De Luca, 1997).

For repetitive torso extension, RMS and λ produced relatively higher sensitivity than other indices, although repeatability of RMS was poor. Previous studies have suggested that RMS has lower repeatability than MdPF (Kollmitzer et al., 1999; Larivière et al. 2002). This may be due to normalizing procedures using maximum RMS values obtained from dynamic MVC test. It can be expected that the maximum RMS of each torso muscle would also have poor reliability, given that contribution of each muscle in producing maximum torque may vary across MVC trials. Though perhaps affected by noise, increasing RMS values can be associated with an increase in muscle activation such as motor unit recruitment and firing rate (De Luca, 1979; Suzuki et al., 2002). That is why previous work has utilized RMS in detecting fatigue development during different torso efforts (Kankaanpää et al., 1998; Clark et al. 2003), though changes of this index with time could not be fitted using linear regression.

6.4.3 Concentric vs. Eccentric

It was expected that concentric and eccentric actions may influence sensitivity of an EMG index due to variations in motor unit activation patterns (Linnamo et al. 2003; Komi et al., 2000). However, inconsistent trends were observed in this investigation. In some cases, eccentric EMG seems to be more sensitive and more predictive than concentric EMG, but the overall trend was conflicting across indices or across utility criteria. In contrast to conflicting results between concentric and eccentric data, observations on this study suggest that All EMG (concentric + eccentric) typically yield better utility than eccentric or concentric. This could be explained by the longer sample window used, resulting in better signal resolution. Further investigation is needed to examine whether the categorization of concentric or eccentric is necessary for muscle fatigue assessment.

6.4.4 Limitations

A major limitation of this investigation is the assumption of linear trends of time-dependent changes in EMG indices during exercise. Though such linear analysis has been suggested in earlier reports (van Dieën et al., 1998; Masuda et al., 1999; Nussbaum, 2001), some investigators also report an exponential trend or a sequence of rapid change and plateau trends in fatigue-related data (Gerdle and Fugl-Meyer, 1992; Lindstrom et al., 1977; Merletti et al., 1991). Another limitation relates to the dynamic effort investigated. The secondary data were taken from the earlier studies, which were limited to repetitive shoulder abductions and torso extensions at 30% and 40% effort levels with two cycle durations. Further study appears necessary that employs similar methods for different dynamic efforts and for different muscles, considering that motor unit behaviors may vary depending on muscle morphology and exercise condition.

6.4.5 Conclusions

The concept of multi-indices of fatigue has been suggested. Since procedures and methods for muscle fatigue assessment during dynamic contraction are still inconclusive, the present study evaluated the utility of different EMG-based fatigue indices. Utility was analyzed in terms of sensitivity, variability, repeatability, and predictive ability. In general, all EMG indices investigated were sensitive to fatigue during shoulder abduction, but not during torso extension. Among alternative indices, an index derived from a Poisson-plot seems to have relatively higher sensitivity, and fractal indicators were associated with higher predictive ability.

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CHAPTER VII
CONCLUSION

7.1 SUMMARY AND STUDY CONTRIBUTION

Two main issues were addressed in the present work, including effects of age on muscle fatigue during isometric and dynamic exercises, and alternative EMG-based fatigue measures for low-level isometric and intermittent dynamic contractions. Findings from this study suggested that both issues were dependent on the muscles investigated. Aging was found to be a critical factor affecting muscle fatigue, while differences due to gender appeared to be marginal. Aging was associated with slower progressions of local fatigue, and the age effects were more consistent for the shoulder than for the torso muscles during both isometric and dynamic exercises. Age effects on fatigue were observed to be typically moderated by effort level, but not by cycle duration. These results, overall, strongly support a basic premise in job design that task and individual factors should be considered simultaneously. The task factors include effort level, effort type, and muscle required, while age can be regarded as a major individual factor.

Muscle fatigue is a complex process, and the current work suggests that multiple indices may be needed for assessment of muscle fatigue using EMG. Four criteria of utility were proposed for comparisons among EMG indices, including sensitivity, variability, repeatability, and predictive ability. Several alternative EMG indices were introduced that derived from logarithmic transformation of EMG power spectra, fractal analysis, and parameter estimation based on a Poisson distribution. In a number of cases, these alternative indices had better utility than the commonly used EMG indices (mean or median of spectral distribution).

Similar to the age effects, a muscle dependency was also observed in evaluation of EMG-based fatigue measures during dynamic efforts. During repetitive shoulder abduction, results of this study support the efficacy of EMG measures for muscle fatigue assessment. However, less temporal consistency in EMG changes was obtained during repetitive torso extension, was and this discrepancy is likely due to an interaction of the complexity of exercise procedures and the muscle involved.

The present work also provides quantitative data concerning on muscle strength, muscle endurance, rates of fatigue, and rates of recovery from the shoulder abduction and torso extension at different effort levels. Though they may be only applicable for similar experimental protocols and comparable age groups, these data, along with existing reports, contribute to a better understanding of age and gender effects on work capacity during static and intermittent dynamic efforts.

7.2 DIRECTIONS FOR FUTURE RESEARCH

Several research issues, which are not completely addressed in this study, remain for future investigations. It can be speculated that load sharing among hip and back muscles is a main reason for the inconsistency of temporal EMG changes during repetitive torso extensions. This needs to be confirmed, and further studies are suggested that investigate multiple muscles simultaneously (agonists and antagonists). It is expected that results obtained from such work would also provide information on age- and gender-related differences in patterns of muscular coordination. A similar study could also be expanded to more realistically simulate occupational tasks, such as lifting, carrying or pulling.

The present work is an initial step toward developing EMG-based fatigue measures that are more sensitive and more reliable for low-level isometric contractions and intermittent dynamic exertions. The alternative measures proposed, while promising, need further development. Several parameters (such as frequency bandwidth) were chosen based on preliminary results from a pilot study, yet at present their physiological meanings are not entirely clear yet. A number of non-linear methods are also available that warrant further evaluation. Eventually, such work is likely to lead to improved EMG-based evaluations and interventions for industrial and clinical areas.

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