Changes in surface electromyography signals and kinetics associated with progression of fatigue at two speeds during wheelchair propulsion

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Abstract—The purpose of this study was to determine whether muscle balance is influenced by fatigue in a recordable way, toward creating novel defensive activity strategies for manual wheelchair users (MWUs). Wheelchair propulsion to a point of mild fatigue, level 15 on the Rating of Perceived Exertion scale, was investigated at two different speeds. Surface electromyographic (EMG) activity of 7 muscles was recorded on 14 nondisabled participants. Kinetic variables were measured using a SmartWheel. No significant effect was found of percentage endurance time on kinetic variables for the two propulsion speeds. Fatigue-related changes in the EMG spectra were identified as an increase of EMG intensity and a decrease of mean power frequency as a function of percent endurance time for the tested muscles under both fast and slow speed conditions. The greater increases in activity for propulsive muscles compared with recovery muscles during fast speed wheelchair propulsion indicated muscle imbalance associated with fatiguing wheelchair propulsion. This study shows how kinetic and EMG information might be used as feedback to MWUs to ensure that they conduct activity in ways that do not precipitate injury.

Key words: biomechanical adaptation, mean power frequency, muscle coordination, muscle imbalance, muscle recruitment strategy, muscle synergy, principal component analysis, rehabilitation, shoulder, wavelet analysis.

INTRODUCTION

Most individuals with spinal cord injuries (SCIs) use manual wheelchairs for mobility at home, school, work, and play [1]. Long-term wheelchair use and its consequences on the musculoskeletal system have become an important issue in manual wheelchair research. Since the physiology of the shoulder is not well adapted to the monotonous nature and peak-force requirements of wheelchair use, the necessarily prolonged and often excessive use of the upper limbs leads to muscle imbalance and attrition injuries [2]. Many manual wheelchair users (MWUs) experience upper-limb pain and injuries that interfere with essential activities of daily living involving wheelchair propulsion and transfer [1]. It is important to develop effective strategies to minimize the destructive effects of shoulder pain and injuries. This is

particularly true for senior citizens who are MWUs, in whom problems with upper-limb pain often result in progressively increased dependency.

Muscle imbalance, defined as predominance of one of a synergist pair of muscles during a movement [3], has become an important topic in the etiology of many musculoskeletal disorders [4]. The shoulder muscles do not gain equally in strength during wheelchair propulsion; only those active in the push phase become stronger, whereas those active in the recovery phase remain at the same level of strength. The repetitive movements required to achieve locomotion will, over time, widen the strength discrepancy between push muscles and recovery muscles [5–7]. Previous studies have suggested muscular imbalance in the shoulder as a source of pain and injury in MWUs [5–6]. Ambrosio et al. have shown that weakness exists in the shoulder adductors in people with paraplegia [7]. This weakness could be a cause for rotator-cuff impingement syndrome [5,8–9]. People with paraplegia also demonstrated a significant weakness during external and internal rotation [10]. Strengthening of shoulder internal and external rotators, as well as adductors, has been recommended in clinical guidelines [11]. Successful rehabilitation processes related to muscle imbalance are often accomplished by addressing the cause of the problem rather than symptomatic treatment of the pain [12]. By understanding muscle imbalances associated with wheelchair propulsion, physical therapists can prescribe appropriate exercises for both treatment and prevention.

The present study identified muscle imbalance associated with fatigue and activity of selected upper-limb muscles during wheelchair propulsion. In general, muscle fatigue plays a critical role in the development of muscle imbalance and overuse [13]. According to the published literature on what precipitates musculoskeletal injuries, injuries around a joint may be caused by uneven fatigue among the different joint muscles [14]. The diverse demands on the muscles surrounding a joint may cause them to fatigue at different rates and to different degrees [14]. Rodgers et al. reported that muscle fatigue in MWUs may be responsible for stress and harmful changes in shoulder joints [15]. Since muscle fatigue leads to decrements in muscle force production [16], it is probable that during strenuous activity, each of the muscles around a joint reaches a point of fatigue at a different time; the ensuing imbalance in force production may lead to unnatural joint motions, abnormal joint stresses, and, ultimately, injuries. Rodgers et al. showed that joint power shifts from the shoulder joint to the elbow and wrist joints with the onset of fatigue during wheelchair propulsion [17].

Continuous monitoring of local muscle fatigue during locomotion is possible by measuring myoelectric activity of particular muscles using surface electromyography (sEMG) [18]. Electromyography (EMG) has been widely used in the assessment of musculoskeletal disorders. The myoelectrical manifestations of muscle fatigue can be observed by a decrease in the mean power frequency (MPF) of the power spectrum [19], whereas EMG amplitude can be used to measure muscular activity [20]. In addition, a real-time kinetic measurement device, SmartWheel (Three Rivers Holdings, LLC; Mesa, Arizona), was used at the same time to estimate the forces and moments applied to the pushrim. A combination of kinetics and EMG activity measurement provides complementary information on the timing and intensity of shoulder muscle activity during wheelchair propulsion with respect to the kinetics of the shoulder joint.

METHODS

Subjects with normal upper-limb function who were untrained (not habitual MWUs) provide a suitable baseline of activity for different groups [21]. Also, many temporary MWUs recovering from conditions such as lower-limb injury are "at risk" [22]. Nondisabled participants were therefore recruited to provide a basis for further MWU studies. During occupational and rehabilitation tasks and daily activities, MWUs do not necessarily perform high intensity propulsion and may become only mildly fatigued. In the present study, wheelchair propulsion to a point of mild fatigue, level 15 on the Rating of Perceived Exertion (RPE) scale, was investigated at two different speeds: slow (0.9 m/s) and fast (1.6 m/s). The purpose of this study was to see if muscle balance is influenced by fatigue in a recordable way so that this information could later be used to create novel defensive activity strategies for MWUs.

Participants

Fourteen nondisabled participants (7 males, 7 females, mean age 30 ± 4 years [range 25–40 years], mean weight 65 ± 12 kg [range 46–87 kg]) volunteered to participate in this study. None reported any previous history of

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upper-limb pain or any neuromuscular disorder. Participants were instructed not to perform any exercise during the 48 hours prior to measurement.

Surface Electromyography

We recorded sEMG activity of upper-limb muscles using parallel-bar EMG sensors (DE-3.1 double differential sensor, 1 mm in diameter and separated by 10 mm [Bagnoli, Delsys Inc; Boston, Massachusetts]). We detected sEMG signals on seven muscles: anterior deltoid (AD), middle deltoid (MD), posterior deltoid (PD), pectoralis major (PM), upper trapezius (UT), biceps brachii (BB), and triceps brachii (TB) on the right shoulder (all participants are right side dominant) after prior removal of the hair and cleaning with alcohol swipes. The EMG signals were sampled at 2,000 Hz and recorded with a 16-bit analog-to-digital converter (PCI 6221, National Instruments Corp; Austin, Texas).

Kinetic

The forces applied to the pushrim of the wheel were measured using a SmartWheel. The SmartWheel is a modified alloy wheel capable of measuring three-dimensional forces and moments occurring at the pushrim. The pushrim kinetic data were collected at 240 Hz. Kinetic and EMG recordings were synchronized.

Test Procedure

Maximum Voluntary Isometric Contraction Test

Prior to the propulsion data collection, EMG activity was recorded during maximum voluntary isometric contraction (MVC). A total of four muscle tests were performed following the methods described by Boettcher et al. [23] and Kelly et al. [24]. The test order was block randomized. A force transducer (Model LCCB-1K, OMEGA Engineering; Stamford, Connecticut) was used to measure the force generated from isometric contractions. The force signals were sampled at 2 kHz. Participant feedback and signal recording was achieved through LabVIEW software (Version 8.5, National Instruments).

Wheelchair Propulsion on Ergometer

A test wheelchair (Quickie GP, Sunrise Medical; Longmont, Colorado) with 56 cm (24 in.) diameter rear wheels, 13 cm (5 in.) polyurethane castor wheels, 41 cm (16 in.) seat width, 41 cm (16 in.) seat depth, and 0° camber angle was aligned and secured over the rollers of an ergometer, which connected to a monitor placed in front of the participant to provide visual speed feedback. The SmartWheel was placed on the right side of the test wheelchair with a standard foam cushion. Participants were given several minutes to get used to propelling the wheelchair and establish a comfortable propulsion technique. Then the participants were advised to apply a semicircular propulsion pattern. The semicircular pattern is recognized by the hands falling below the pushrim in the recovery phase [25–26]. Participants were given ample time to acclimate to this pattern prior to data collection.

To facilitate awareness and rating of signs of fatigue during testing, participants were instructed to use a scale of RPE. The RPE was used to evaluate the participants' perceived level of exertion, where 6 = very light exertion and 20 = very hard exertion. Level 15 marks the highest level of effort prior to exhaustion in this study. Participants were asked to perform two trials of wheelchair propulsion with trial order block randomized, one at 0.9 m/s and one at 1.6 m/s. In each case, the participant would continue pushing at the set speed until he or she rated his or her exertion at 15. Participants felt mildly fatigued after testing. At least 10 minutes of rest was given between trials. The slow speed, 0.9 m/s, was selected because it is close to minimally safe speed (1.06 m/s, the speed required to cross a street with a timed light) [27]. Self-selected speeds for propulsion on a level surface have been reported from 0.8 to 1.6 m/s [27–29]. The upper limit, 1.6 m/s, was selected to present a challenging and strenuous situation for the participants.

Endurance time differed according to each participant's fitness level and strength. Results were normalized by setting the actual endurance time in each case as 100 percent and dividing it into five equal windows expressed as percent endurance time. For each participant, the average of the last 10 propulsion cycles of each endurance time window (% endurance time) at the fast and the slow speed were selected for data analysis. The first 20 percent endurance time was defined as fresh state, whereas the 100 percent endurance time was defined as fatigue state.

DATA ANALYSIS

Kinetics Data Analysis

A complete propulsion cycle was defined as palm strike to palm strike. The push phase is defined as palm strike to palm off [30], and the recovery phase is defined

as palm off to palm strike at the pushrim for next cycle. For this study, the onset of propulsion was defined as the point at which a propulsive moment (M_z) was applied to the SmartWheel, and the end of propulsion was defined as the point at which the moment returned to zero. The recovery phase was defined as the end of propulsion to the next onset of propulsion, when the SmartWheel moment was zero.

For each speed condition, variables from all the cycles collected at steady state were determined. The key kinetic variables calculated were peak resultant force (peak F_{tot}), peak tangential force (peak F_t), average resultant force (Ave F_{tot}), average tangential force (Ave F_t). The resultant force (F_{tot}) is the total force applied to the pushrim. The tangential force (F_t) is the force directed tangentially to the pushrim. Mechanical effectiveness was calculated by F_t/F_{tot}. In addition, by using the output of the SmartWheel, the push frequency, push length in degrees, and push time were determined.

Wavelet Analysis of Electromyography Signal

When dynamic contractions are considered, the spectral estimation technique used to describe them must be chosen carefully, taking into account the specific type of nonstationarity exertion affecting the signal of interest. We used wavelet analysis with well-defined time and frequency resolution, which has shown to provide a highly sensitive method of assessing nonstationary EMG [31–32]. All signal processing was performed using custom programs written in Mathematica (version 6.0, Wolfram Inc; Champaign, Illinois). The time axis of the EMG data was normalized to percentage of propulsion cycle time and synchronized with the kinetic data. The EMG signals were resolved into intensities in time-frequency space using wavelet techniques [31]. The method has been described in detail in previous articles [32–34]. A filter bank of 11 nonlinearly scaled wavelets was used, index by k, with center frequency, f_c, ranging from 7 Hz (wavelet 0) to 395 Hz (wavelet 10). The first wavelet of EMG covered a frequency band of 0 to 10 Hz, which is typically associated with movement artifacts. We reduced the effects of movement caused by dynamic contractions by removing the first wavelet from the spectra. The mean intensity for each participant for the MVC was calculated and used to normalize the spectra for the respective participant. Total intensity (i_k) was given by summing the intensities over the selected wavelets (10–350 Hz, $k = 1$–9). Total intensity is a measure of the time-varying power within the signal

and is equivalent to twice the square of the root-mean-square. The instantaneous MPF was calculated by

$$\text{MPF} = \frac{\sum_k f_c(k) i_k}{\sum_k i_k} \quad , \quad \textbf{(1)}$$

with f_c representing the center frequency of each wavelet and the mean frequency calculated as the mean of the MPF values taken from whole propulsion cycle.

To determine the duration of EMG activity, a threshold (10%–15% MVC) was computed for each muscle and each participant [18]. The onset of EMG activity was defined as the time when the EMG total intensity remained above a threshold. The cessation of the EMG activity was defined as the time when EMG total activity remained below the threshold level. The duration of EMG activity was then calculated as the time difference between the onset and cessation of the EMG activity. The results are reported as percentage of propulsion cycle.

Principal Component Analysis

Principal component analysis (PCA) employed the techniques previously reported [32]. The data set consists of a $p \times N$ matrix, where $p = 9$ wavelet domain and $N = 700$ (participants × 10 cycles × time windows) for each muscle. A PCA was carried out on this matrix using the customized programs in Mathematica. The principal components (PCs) were calculated from the covariance matrix of the data set matrix with no prior subtraction of the mean data, so the PCs describe the components of the entire signal [33].

The PC weighting is given by the eigenvector and the PC loading score is given by the eigenvalue, a scalar value that describes the amount of each eigenvector in each measured spectrum [35]. A quantitative measure of the contribution of high and low frequency content within the signal is given by the angle θ formed between the first and second PC (PCI and PCII) loading scores [33,35–36]. A higher value of θ represents relatively more low frequency signal content and a smaller θ value is associated with relatively more high-frequency content in the EMG signal.

STATISTICS

Mean values of the kinetic and EMG variables were calculated from each propulsion cycle. Statistical analysis

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was performed using SPSS (SPSS 16, SPSS, Inc; Chicago, Illinois). The Kolmogorov-Smirnov test and Shapiro-Wilk test were performed first to test data normality. The tests were not significant, so the data were normal distributed. Mixed model analysis of variance (ANOVA) was then used for statistical analysis. The between-subject factor was speed. The within-subject factor was percent endurance time. Repeated measures ANOVAs were used to analyze EMG total intensity, EMG duration, θ, and kinetic variables. Multiple comparisons between percent endurance time ($N = 5$) and speeds ($N = 2$) were made according to Bonferroni's method with a significance level of $p < 0.05$. All data are reported in the text as mean ± standard deviation, while the results are graphed in each figure as mean ± standard error of mean. Significant level was set at $p < 0.05$ for all statistical procedures.

RESULTS

The highest level of exertion was set at RPE 15 (out of a maximum of 20) to allow for a sufficiently challenging level of exertion while minimizing the risk of injury or overuse syndromes. The average propulsion duration at the fast speed (1.6 m/s) was 154 ± 74 s; the average number of push cycles at the fast speed was 172 ± 111 cycles. The average propulsion duration at the slow speed (0.9 m/s) was 334 ± 139 s; the average number of push cycles for the slow speed was 307 ± 146 cycles. The average RPE was 15 ± 1 for both the slow and fast speeds. An RPE of 15 has been deemed an adequate fatigue level, representing 75 to 90 percent maximum oxygen consumption [37–38].

Propulsion Kinetics

Figure 1 displays the kinetic mean values calculated at 20, 40, 60, 80, and 100 percent of the endurance time during prolonged wheelchair propulsion at the slow and fast speeds. Peak F_{tot}, Peak F_t, Ave F_{tot}, Ave F_t, push time, and push frequency were significantly different between the two speeds, whereas no significant effect was found for percent endurance time on these kinetic variables for the two propulsion speeds. Push length at 20 percent endurance time was significantly longer than at 100 percent endurance time for both speeds, while the main effect of percent endurance time had no statistically significant ($p = 0.052$) effect on push length for either speeds.

Electromyography Characteristics

The EMG intensity, MPF, and θ at 20, 40, 60, 80, and 100 percent of the endurance time are shown in **Figure 2**. A significant effect of percent endurance time on the EMG intensity was present for the tested muscles ($p < 0.05$), except BB ($p = 0.08$); EMG intensity increased continuously throughout the propulsion duration for both speed conditions. The EMG intensities for AD, PM, BB, and TB were significantly greater at the fast speed than at the slow speed ($p < 0.05$), while no significant differences were present in EMG intensity for UT, MD, and PD between the two speeds. Mixed model ANOVA revealed that the EMG MPF decreased ($p < 0.05$) with percent endurance time in all seven muscles for both speed conditions, while no significant difference was present in MPF between the two speeds. No significant differences were present in θ between the two speeds. Significant differences ($p < 0.05$) did exist in θ along the scale of percent endurance time, showing that θ differed significantly over time for AD ($p < 0.001$), PM ($p < 0.001$), TB ($p = 0.002$), UT ($p < 0.001$), and MD ($p < 0.001$), except in BB ($p = 0.16$) and PD ($p = 0.08$).

Although shoulder muscle activity increased significantly during fatigue (**Figure 3**), no significant effect of the percent endurance time on the EMG duration existed for all the tested muscles. The significant differences in EMG duration did, however, exist between the two speeds, showing that muscle activity differed significantly at different speeds.

DISCUSSION

Effect of Muscle Fatigue on Electromyography Spectra

Fatigue-related changes in the EMG spectra were identified as an increase of EMG intensity and a decrease of MPF as a function of percent endurance time for the tested muscles (except BB) under both fast and slow speed conditions. Our findings were in agreement with previous investigations showing that a compression of the power spectrum to lower frequencies is typically observed as muscle fatigue [39–40].

The angle θ is formed by the PCI and PCII of the spectra. PCA extracts the important features in the signal, so some variables, such as movement artifacts [41], were given lower-order components and were excluded. Therefore, the main PCs contain a significant proportion of the spectra without being skewed by confounding
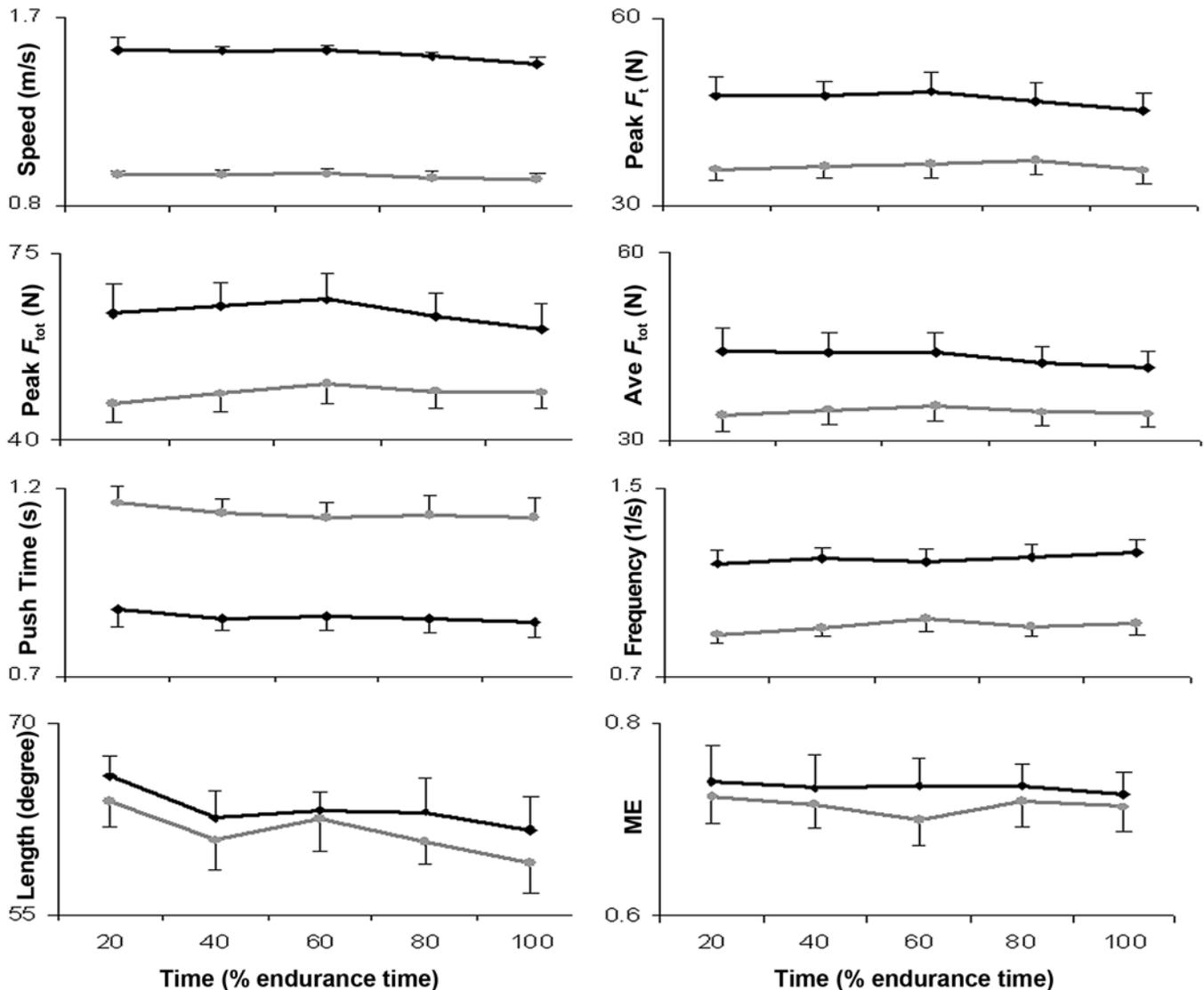
Figure 1.
Changes in pushrim kinetics parameters: speed, peak total force (peak F_{tot}), peak tangential force (peak F_t), average total force (Ave F_{tot}), push time, push frequency, push length in degrees, and mechanical effectiveness (ME; F_t/F_{tot}) as function of time (expressed as percentage of endurance time) during fast speed (black line) and slow speed (gray line) wheelchair propulsion. Values graphed as mean ± standard error of mean.

effects. In the present study, the major source of variation in the EMG signal was attributed to fatigue, and so the first two PCs (PCI and PCII), that account for more than 85 percent of the EMG signal, should be sensitive to fatigue. The contribution of PCII loading scores relative to PCI loading scores (quantified by angle θ) accounts for frequency shifts in the EMG intensity spectra. Thus, in this fatiguing study, the decreases in MPF that occur over time concur with increases in θ.

As shown in **Figure 2**, the EMG intensity lines become progressively steeper in the fast speed condition than in slow speed condition as a function of percent endurance time, which may indicate the recruitment of additional motor units (MUs), particularly the faster MUs, to generate higher force at the fast speed. The fast speed requires a higher propulsive force on the pushrim than the slow speed; therefore, both the number and size of recruited MUs increased for the higher mechanical

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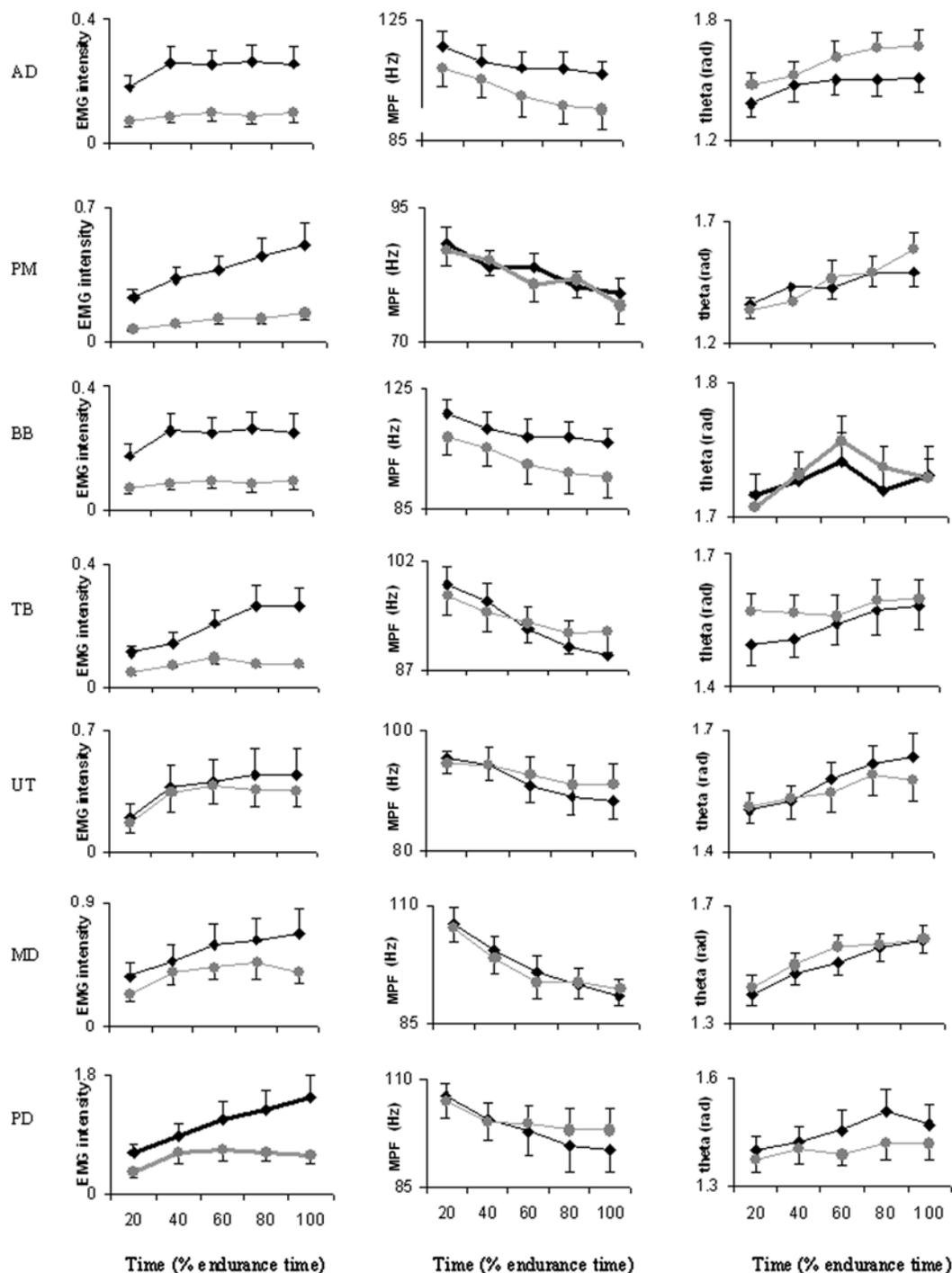
Figure 2.
Changes in electromyography (EMG) intensity, mean power frequency (MPF), and theta as function of time (expressed as percentage of endurance time) during fast speed (black line) and slow speed (gray line) wheelchair propulsion for anterior deltoid (AD), pectoralis major (PM), biceps brachii (BB), triceps brachii (TB), upper trapezius (UT), middle deltoid (MD), and posterior deltoid (PD). Each point is average value (mean ± standard error of mean, n = 14) of 10 cycles of each time window.
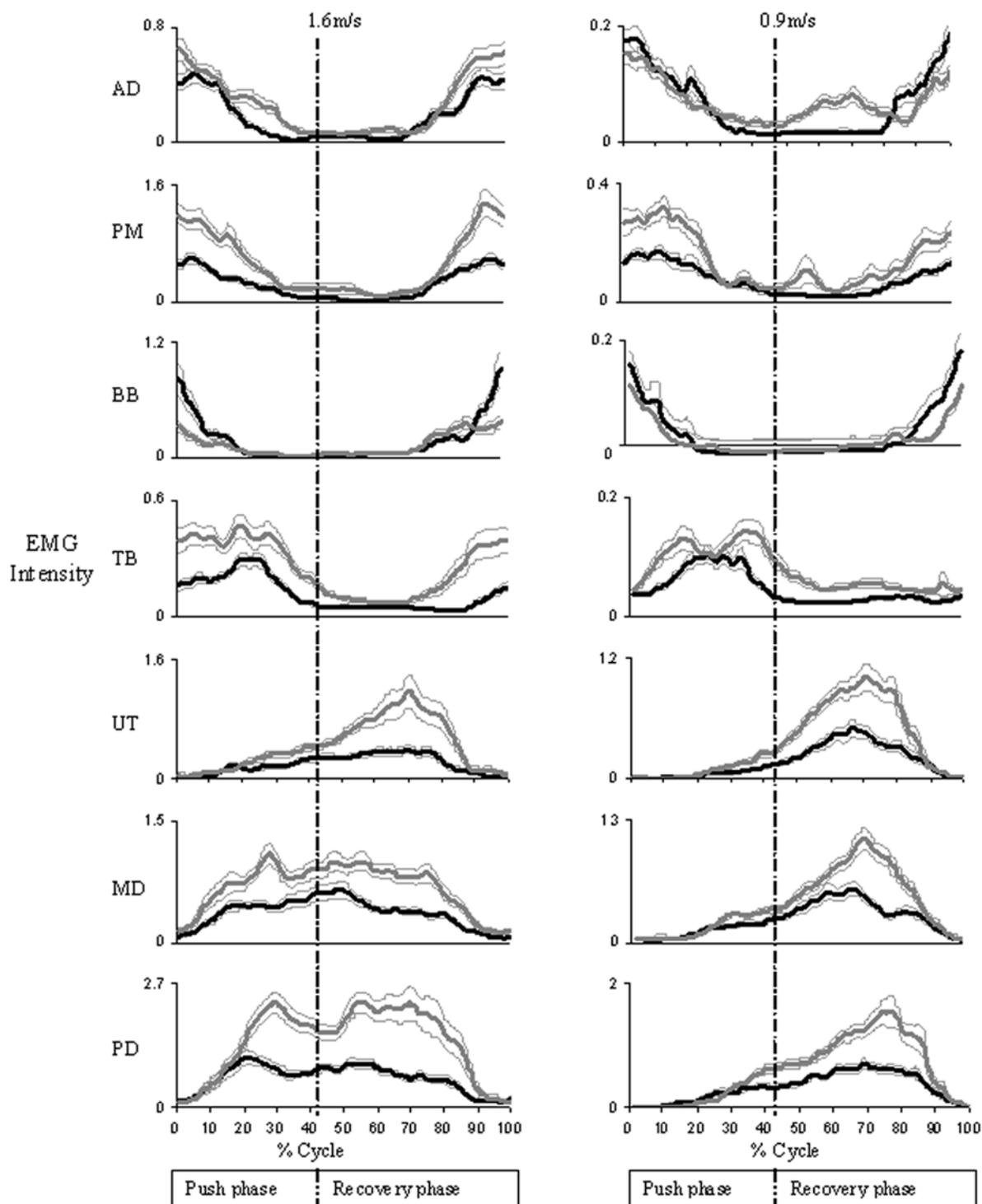
Figure 3.
Electromyography (EMG) intensity for 7 shoulder muscles obtained at 20 percent endurance time window (black lines) and at 100 percent endurance time window (gray lines) of two speeds of wheelchair propulsion. Time zero indicates hands on pushrim. Each profile represents mean (thick line) ± standard error of mean (thin lines) obtained from averaging individual data across 10 consecutive propulsion cycles of each time window. AD = anterior deltoid, BB = biceps brachii, MD = middle deltoid, PD = posterior deltoid, PM = pectoralis major, TB = triceps brachii, UT = upper trapezius.

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requirement within individual muscles. On the other hand, the increase in θ values and decline in MPF were in general greater for the fast speed than for the slow speed.

Effect of Muscle Fatigue on Wheelchair Kinetics

Previous studies have investigated changes in wheelchair biomechanics caused by fatigue. Rice et al. reported an increase in push time during an extended period of propulsion, while stroke frequency remained static [42]. In the present study, no significant effects of kinetic variables on the endurance time of wheelchair propulsion were found. This might be because the participants in the study were nondisabled individuals and they only became mildly fatigued. It has been reported that experienced MWUs compensate for fatigue differently than nonusers. Rodgers et al. reported a power shift from the shoulder to the elbow and wrist joints during fatiguing wheelchair propulsion [17]. Compared to the increased activities of the other shoulder muscles during fatigue (**Figure 2**), the decrease in BB activity during strenuous wheelchair propulsion may be related to reduction of push length (**Figure 1**). The initiation of contact angle was closer to the top dead center in the fatigue stage than in the fresh stage as recorded from the kinetic system, which reduced the activation level of BB. The reduction of push length associated with a decrease in the elbow and shoulder range of motion indicates upper-limb biomechanical adaptations to fatigue in an effort to maintain the target velocity at a higher push frequency.

Effect of Muscle Fatigue on Muscle Recruitment Pattern

It has been reported that force production around the joint becomes unbalanced in relation to the fatigue state of each individual muscle during lifting tasks [14] and cycling [43]. In the present study, EMG intensities for the propulsive muscles—AD, PM, BB, and TB—were significantly greater at the fast speed than at the slow speed, while no significant differences were present in EMG intensity for recovery muscles—UT, MD, and PD—between the two speeds. This leads to the following recommendations:

1. Avoid muscle imbalance. The selective recruitment of the propulsive muscles and task dependency of muscle fatigue is likely to contribute to muscle imbalances. This situation has the potential to cause abnormal or unnatural motions of the joints, creating significant abnormal stress distributions and possibly leading to injury. Indeed, it has been reported that long-term use of the manual wheelchair leads to muscle imbalance, overdevelopment, strengthening and shortening of the AD and pectoralis, and weakening and lengthening of the opposing muscle groups [5,7,44–46], which should be avoided.

2. Strengthen posterior musculature. Because of the nature of a wheelchair user's movement patterns, special attention should be made to strengthening the posterior musculature to ensure muscle balance between the anterior and posterior sides of the body. Proper muscle balance will help maintain efficient movement and allow for maximum power production, as well as help prevent shoulder impingement and capsular problems that may arise because of this type of muscle imbalance.

3. Develop protective habits. It is essential for new MWUs to establish "protective habits" from the start of using a wheelchair and that this knowledge be used to assist both the process of biofeedback and the eventual development of supporting technologies that will provide direct feedback to users on the level of their activity.

4. Provide a positive influence. To effectively develop a system that can distinguish well-coordinated patterns from aberrant ones, movement patterns must be communicated to the user in a way that influences his or her activity positively; this study demonstrates that EMG and kinetic measurements can be used to make such distinctions during wheelchair propulsion. This approach requires validation in a patient population.

5. Re-evaluate early. If the subject is not improving, a period of EMG monitoring to enhance biofeedback may help the subject retrain the correct muscle patterning.

LIMITATIONS

In the present study, the general pattern of EMG activity of the seven muscles was similar to the results of most previous studies. In wheelchair users, the timing of EMG activity may happen in a slightly different way because of compensation for muscle impairment and poor trunk control. Indeed, Mulroy et al. reported that the level of SCI significantly affects the shoulder muscle recruitment patterns during wheelchair propulsion [47].

This study investigated the effect of mild fatigue on wheelchair propulsion in nondisabled individuals as a pilot study in a stimulated environment (static ergometer).

The wheelchair was not measured to the participants; the same test wheelchair was used for data collection. The axle position of the rear wheel was set as default for most wheelchairs, which may have resulted in varied seating positions among participants [26,48–49]. Future study should focus on using the personal wheelchairs of users who have shoulder pain and injuries. The physical condition of the wheelchair users, such as level of injury, pain history, and fitness level, should also be considered.

CONCLUSIONS

The purpose of this study was to develop algorithms for muscle-fatigue detection and muscle-recruitment patterns in routine wheelchair propulsion scenarios, e.g., daily practice. Our long-term interest is to develop an inexpensive, real-time, and noninvasive approach that can be used in wearable devices.

The results indicate that an association existed between biomechanical adaptation to fatigue and fatigue-related changes in the frequency content of the surface EMG from upper-limb muscles during fatiguing wheelchair propulsion. MPF and θ, determined by time-frequency analysis and PCA, showed sensitive and consistent changes in terms of muscle fatigue at low to moderate levels of wheelchair propulsion. During faster speed wheelchair propulsion, the activities of propulsive muscles and recovery muscles had both increased, but propulsive muscles increased more. This could imply a muscle imbalance associated with prolonged wheelchair propulsion. The ability to measure the rate of fatigue and muscle recruitment patterns can enhance our understanding of shoulder muscle function and potentially provide a tool for fatigue assessment and imbalance evaluation.

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