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ARTICLE

Wheelchair propulsion fatigue thresholds in electromyographic and ventilatory testing

Liping Qi[1,2] · Li Zhang[1] · Xiang-Bo Lin[3] · Martin Ferguson-Pell[2]

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Abstract

Study design Qualitative study.

Objective The objective of the present study are physiological processes occurring when the intensity of manual wheelchair propulsion approaches levels causing muscular fatigue. In particular, we set out to (1) detect the electromyographic (EMG) and ventilatory fatigue threshold during a single wheelchair incremental test, (2) examine the relationship between EMG threshold (EMGT) and ventilatory threshold (VT), and (3) detect the EMG threshold differences between the propulsive and recovery muscle synergies.

Setting Biomechanics laboratory at the University of Alberta, Canada.

Methods Oxygen uptake and EMG signals from ten wheelchair users (seven males and three females) were recorded as they were each performing an incremental propulsion bout in their own wheelchairs on a wheelchair ergometer. The *V*-slope method was used to identify the VT, and the EMGT of each of the eight muscles (anterior deltoid, middle deltoid, posterior deltoid, infraspinatus, upper trapezius, sternal head of pectoralis major, biceps brachii, and triceps brachii) was determined using the bisegmental linear regression method.

Results For each participant, we were able to determine the EMGT and VT from a single incremental wheelchair propulsion bout. EMGT stands in good agreement with VT, and there was a high similarity in EMGT between push and recovery muscles (intraclass correlation coefficient = 0.91).

Conclusion The EMG fatigue threshold method can serve as a valid and reliable tool for identifying the onset of muscular fatigue during wheelchair propulsion, thus providing a foundation for automated muscle fatigue detection/prediction in wearable technology.

Introduction

Muscle fatigue is defined as a decrease in the force-generating capacity of a muscle that is induced by maximal and even submaximal exercise [1]. As 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 fatigue and imbalance [2]. In addition, manual wheelchair propulsion is an inefficient mode of ambulation. Previous studies have investigated muscle fatigue in relation to stroke velocity and stroke patterns during prolonged wheelchair propulsion [3, 4].

There has been considerable interest in the development of systems that can predict when and how muscles will produce fatigue [5]. The effect of muscle fatigue can be observed by electromyographic (EMG) signals. EMG signals have been used as an index of fatigue. Recently, EMG has also been used to detect the fatigue threshold in groups of both athletes and nonathletes [6–8]. A more conventional method for pinpointing fatigue thresholds is the measurement of ventilation parameters during incremental cycling tests on a cycle ergometer [7]. Several studies have correlated EMG thresholds (EMGTs) with aerobic–anaerobic transitions predicted by ventilatory thresholds (VT) during

✉ Liping Qi
lipingqi@dlut.edu.cn

1 School of Biomedical Engineering, Dalian University of Technology, Dalian 116024 Liaoning, China

2 Faculty of Rehabilitation Medicine, University of Alberta, Edmonton, AB T6G 2G4, Canada

3 School of Information and Communication Engineering, Dalian University of Technology, Dalian 116024 Liaoning, China

incremental cycling tests [9]. It has been suggested that EMGT seems to be more workable than the VT for predicting the onset of muscle fatigue in athletes [10]. The EMGT is derived from the breakpoint in the relation between EMG amplitude and time during incremental cycling test [6]. The nonlinear increase in EMG amplitudes is attributed to changes in the recruitment patterns of different muscle fibers, especially the increased recruitment of type II fibers [11, 12]. The consequence of recruiting additional type II fibers is an accelerated fatigue process of the involved muscles because of increased lactate and H^+ concentration [8, 10]. Regarding the muscles most active in wheelchair propulsion, the proportion of type II muscle fibers appears to be greater in the upper-body muscles than in those of the lower limbs, ~55% in infraspinatus (IS), 58% in biceps brachii (BB), and 67% in triceps brachii (TB) [13]. A nonlinear increase in muscle activity could therefore be more noticeable with respect to the onset of fatigue during repetitive upper limb movements, owing to such a high proportion of type II muscles. Both VT and EMGT have served as a valid and reliable tool for estimating fatigue thresholds [7, 8]. However, the various fatigue thresholds have so far been determined predominantly from able-bodied individuals and lower-extremity exercises.

Previous studies have shown that the EMG fatigue threshold can be determined from a single incremental cycling and treadmill test [14, 15]. The feasibility of a single-session test to determine the neuromuscular fatigue threshold may have potential applications in various rehabilitative populations such as wheelchair users. During wheelchair propulsion, two muscle synergies have been identified, namely the push-phase synergy and the recovery-phase synergy [16]. The push- phase synergy is dominated by the anterior deltoid (AD), pectoralis major (PM), and BB, whereas upper trapezius, middle deltoid (MD), and posterior deltoid (PD) have their primary activity during the recovery phase [17]. Previous studies reported differences in muscle activation patterns between the push and recovery muscles associated with progression of fatigue during wheelchair propulsion [18]. EMG testing has the advantage of yielding muscle-specific rather than summary threshold data in a minimally invasive manner. In addition to each muscle's activation profile, it should also provide an answer to the question whether the fatigue thresholds of the push muscles vary significantly from those of the recovery muscles, given the differences in their activation.

The purpose of the present study was thus (1) to detect the EMG and ventilatory fatigue thresholds during a single wheelchair incremental test, (2) examine the relationship between EMGT and VT, and (3) to determine whether the fatigue thresholds determined from EMG signals for estimating the onset of neuromuscular fatigue are different between propulsive and recovery muscles.

Methods

Participants

In total, ten wheelchair-dependent participants (seven males and three females) with SCI at the T6 level or below volunteered for the study.

Inclusion criteria: participants experience a spinal cord injury resulting in full-time use of a manual wheelchair. They were recruited at least 6 months after the date of their injury and have been discharged from acute inpatient rehabilitation. Their ASIA impairment score will be within the range AIS A–C (complete to motor incomplete) and have a neurological level in the range T6–T12.

Exclusion criteria: age under 18 or over 50 years, abuse of alcohol, and/or controlled substances, musculoskeletal injury that has affected normal, symmetrical wheelchair ambulation within the last month, uncontrolled exercise-induced asthma, heart disease, obesity- (BMI > 30) known contraindications to maximal exertion exercise, including previous incidents of autonomic dysreflexia during exercise, frequent episodes of autonomic dysreflexia, and visual impairment (participants need to see performance feedback on wheelchair dynamometer system LCD screen).

Table 1 shows the physical characteristics of each participant. They all gave their informed consent in accordance with the procedures approved by the University of Alberta Ethics Committee (ID: Pro00015185). We certify that all applicable institutional and governmental regulations concerning the ethical use of human volunteers were followed during the course of this research.

Instruments

Surface electromyographic (sEMG) activity of upper-extremity muscles was recorded using parallel-bar EMG sensors (DE-3.1 double-differential sensor, three 1-mm-diameter muscle site contacts separated by 10 mm, Bagnoli^{TM}, Delsys Inc., Boston, MA, USA). sEMG sensors were placed on the right shoulder on eight muscles: AD, MD, PD, IS, upper trapezius (UT), sternal head of PM, BB, and TB according to the recommendations of the SENIAM Project [19]. Prior to sensor placement, the skin at each electrode site was dry-shaved and cleaned with alcohol. The EMG signals were sampled at 2000 Hz.

Oxygen uptake (VO_2, l/min) and carbon dioxide output (VCO_2, l/min) were continuously measured using a computerized gas-analyzing system (Oxycon Mobile, CareFusion Respiratory Care, CA, USA). Respiratory gas exchange measurements were obtained every 5 s. System calibration was undertaken before each trial.

The participant's own wheelchair was secured to an instrumented roller ergometer connected to a screen placed

Table 1 Injury and physical characteristics of participants.

Participant	Sex	Age (year)	Weight (kg)	Type of injury	Time since injury (year)	ASIA impairment scale grade
1	M	47	99.2	SCI	12	T6 AIS B
2	M	41	97.2	SCI	18	T6/T7 AIS A
3	M	44	125.4	SCI	10	T11 AIS A
4	M	49	80.5	SCI	2	T11 AIS A
5	M	45	68.3	SCI	2.5	T12/L1 AIS A
6	M	34	68.4	SCI	17	T12 AIS A
7	F	55	73.4	SCI	3.5	T11 AIS A
8	F	51	64.1	SCI	18	T12 AIS A
9	F	29	57.8	SB	12	L2
10	M	33	93.1	SB	18	T10
Mean (SD)		42.1 (8.4)	79.8 (22.0)		10.4 (6.9)	

SCI spinal cord injury, *SB* spina bifida.

in front of the participant for visual speed feedback. The ergometer consisted of two independent cylindrical steel rollers with radius 0.158 m and a mass of 26.4 kg, one for each rear wheel, supported by pillow-block bearings (NSK P208, Japan) within a wooden structure to support the wheelchair. Workload was controlled through friction applied to each roller by a fabric strap attached to pneumatic actuators of a digital pressure controller (FESTO, Esslingen am Neckar, Germany), with a proportional valve to regulate the required air pressure. The desired workload through friction was controlled by a computer program (NI LabVIEW 2012, National Instruments Corporation, Austin, TX, USA).

Test procedure

Participants warmed up for about 5 min while getting used to the ergometer and visual propulsion speed feedback. Participants performed an incremental test at a constant speed of 1 m/s to exhaustion. The workload began at 10 W and increased by 5 W every minute to the point of maximum volitional exhaustion. Two of the participants were practicing paraplegia athletes. For them, the workload was increased in steps of 10 W so that volitional exhaustion could occur within 8–12 min. The endpoint of the test was determined when the participant volitionally stopped owing to fatigue, or the investigators determined that the participant could not maintain the expected speed after three warnings.

Ratings of perceived exertion (RPE) were also recorded using the Borg scale during the incremental tests. Participants were asked every 2 min to give two ratings (RPE_respiration and RPE_arm) of perceived exertion by nodding to the applicable numbers on the Borg scale. The Borg Scale was placed in full view of the participants throughout the exercise trials.

Data analysis

The EMG signals were resolved into intensities in time–frequency space using wavelet techniques, a sensitive method for assessing nonstationary EMG signals. This method has been described in detail in previous papers [20]. The intensities from the wavelets (10–350 Hz, $k = 1$–10) were summed to give the EMG total intensity. The EMG total intensity is a measure of the time-varying power within the signal, and is equivalent to twice the root-mean square.

Estimation of ventilatory and EMG thresholds

VT refers to the breakpoint during exercise at which ventilation starts to increase at a faster rate than oxygen uptake (VO_2). VT was calculated by the *V*-slope method [21].

A bisegmental linear regression algorithm was adopted for estimating sEMG breakpoints for each muscle and each participant. The curve fitting was performed in Origin 2018 (OriginLab Corporation, MA, USA). The parameters were estimated by the least-square method to identify two linear trends in variables over time. The algorithm established two regression lines for all datasets that could be divided into two contiguous groups. A linear regression was applied from the beginning of the EMG signal to a point "n." A second linear regression was applied from the point "$n + 1$" to $n =$ end. The combination of segments that presented the least pooled residual sum of squares was considered the best adjust. The intersection between these two line segments in EMG total intensity versus time was considered as the EMG breakpoint (Fig. 1).

Statistical analysis

All statistical procedures were performed using the statistical software package SPSS 22 (IBM SPSS, Armonk, New York).

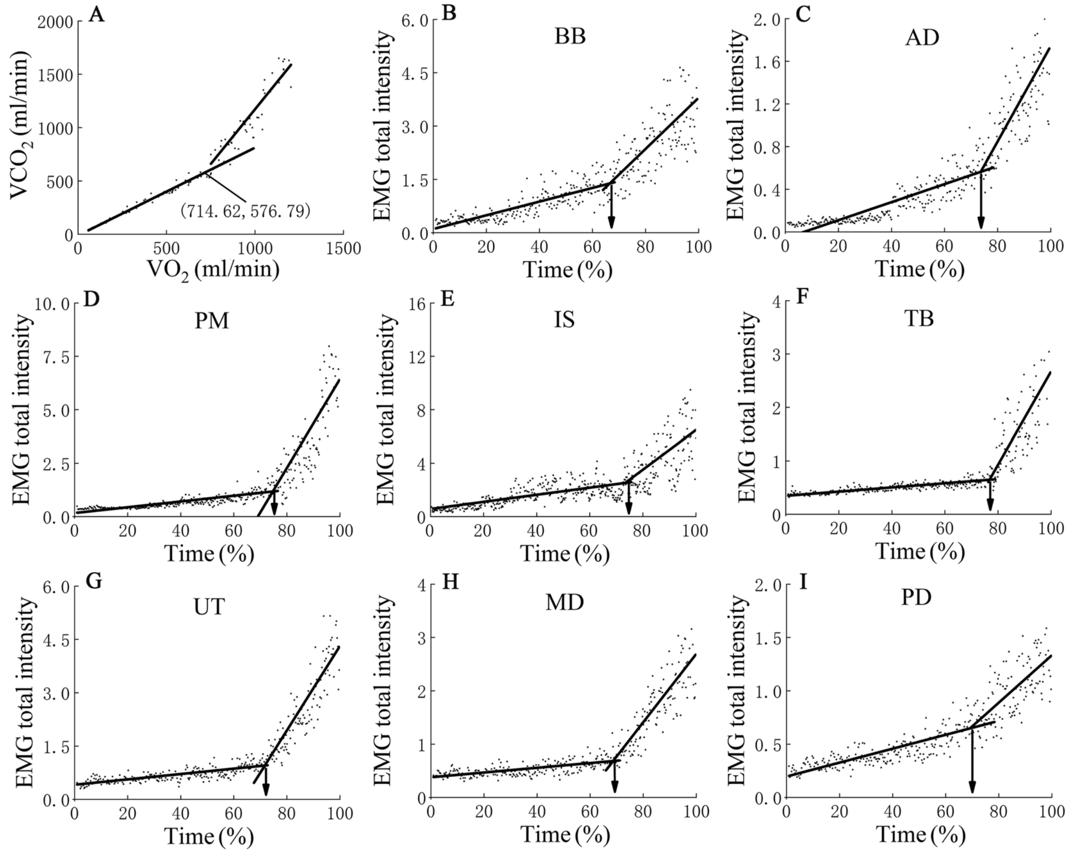

Fig. 1 Determination of ventilatory threshold (VT) and EMG threshold (EMGT) from a representative participant. The VT is the point at which the slope of the relative rate of the increase in VCO_2 to VO_2 changes (**a**, top left). EMGT is the intersection between two line segments in EMG total intensity. **b** biceps brachii, **c** upper trapezius, **d** anterior deltoid, **e** middle deltoid, **f** posterior deltoid, **g** pectoralis major, **h** triceps brachii, **i** infraspinatus.

Shapiro–Wilk test was performed first to confirm data normality. Intraclass correlation coefficient (ICC) was calculated to estimate the VT and EMGT intraclass reliability. Meanwhile, Bland and Altman plot [22] was used to assess the agreement between EMGT and VT measurements. The Bland and Altman plot analysis is a simple way to evaluate a bias between the mean differences, and to estimate an agreement between two quantitative methods of measurements. The quantification of the agreement of two measurements (EMGT vs. VT) was examined by the mean and standard deviation (SD) of the differences between two methods. Statistical significance was set at $p < 0.05$ for the tests.

Results

Table 2 shows the values of RPE_respiration and RPE_arm at different % VO_2 peak levels. The average time to maximum volitional exhaustion is 620 ± 172 s during the incremental tests.

The occurrence of the VT and individual breakpoints (EMGT) for each muscle and participant is reported in Tables 3 and 4. The EMG fatigue threshold (EMGT) represents the maximum force (% VO_2 peak) or power (% maximum workload) that produces an EMG amplitude breakpoint. For a few muscles and participants, we were unable to detect the occurrence of the EMG breakpoint, namely PM of participant B, MD of participant G, and AD of participant I. Those gaps were treated as missing data in the subsequent statistical analysis. The ICC (Cronbach's Alpha) is 0.91, which indicates a high similarity between VT and EMGT (from the eight tested muscles).

Figure 2 illustrates how the EMGT and VT differ in terms of their average % value in relation to VO_2 peak and maximum workload. As for the agreement analysis proposed by Bland and Altman, all individual values of eight

Table 2 The values of RPE_respiration and RPE_arm at different % VO_2 peak levels during the incremental tests.

	40% VO_2 peak	60% VO_2 peak	80% VO_2 peak	100% VO_2 peak
RPE_respiration	9.4 ± 0.7	11.2 ± 1.8	13.7 ± 2.1	17.3 ± 2.0
RPE_arm	9.2 ± 0.9	10.9 ± 1.8	13.6 ± 2.3	17.1 ± 2.2

Table 3 The occurrence of ventilatory threshold (VT) and individual EMG breakpoints (EMGT) for each muscle and participant. The occurrences of VT and EMGT were normalized in percentage of peak oxygen uptake (% VO_2 peak).

Participant	Push synergy					Recovery synergy			
	VT	BB	AD	PM	IS	TB	UT	MD	PD
10	41	55	50	36	40	38	39	48	47
7	45	31	37	39	34	51	32	NaN	36
9	46	76	NaN	63	57	65	71	78	60
6	48	56	60	50	54	58	60	54	50
5	51	40	52	71	49	53	56	63	58
4	51	39	49	54	43	54	35	65	63
2	51	67	50	NaN	83	61	59	63	70
8	58	53	60	64	82	68	60	74	74
1	63	52	56	59	47	60	57	54	58
3	65	62	81	64	81	75	55	69	66
Mean	52	53	55	56	57	58	52	63	58
SD	8	14	12	12	18	10	13	10	11

The order is sorted by VT (low to high). *NaN* not a number, breakpoint was not identified.

VT ventilatory threshold, *BB* biceps brachii, *AD* anterior deltoid, *PM* pectoralis major, *IS* infraspinatus, *TB* triceps brachii, *UT* upper trapezius, *MD* middle deltoid, *PD* posterior deltoid.

Table 4 The occurrences of ventilatory threshold (VT) and EMG threshold (EMGT) were normalized in percentage of maximum workload (% maximum workload).

Participant	VT	Push synergy				Recovery synergy			
		BB	AD	PM	IS	TB	UT	MD	PD
8	33	33	33	56	56	56	33	56	56
5	38	25	38	62	38	38	50	50	50
7	38	23	38	38	38	54	38	NaN	38
2	46	54	46	NaN	73	54	54	54	64
6	46	46	46	46	46	46	46	46	46
10	46	54	54	38	38	38	38	46	46
3	50	50	70	50	70	60	40	50	50
9	50	75	NaN	62	50	62	62	75	62
4	54	38	46	54	46	54	38	62	62
1	78	67	67	67	56	67	67	67	67
Mean	48	46	49	53	51	53	47	56	54
SD	12	17	13	10	13	10	11	10	9

The order is sorted by VT (low to high). *NaN* not a number, breakpoint was not identified.

VT ventilatory threshold, *BB* biceps brachii, *AD* anterior deltoid, *PM* pectoralis major, *IS* infraspinatus, *TB* triceps brachii, *UT* upper trapezius, *MD* middle deltoid, *PD* posterior deltoid.

muscles were within the limits of 1.96 SD between EMGT and VT (Fig. 2). The Bland and Altman procedure thus confirmed a good concordance between EMGT and VT.

Discussion

The major finding of this study was that bisegmental linear regression as used in this study is a viable method for determining the EMG fatigue threshold (EMGT). We were able to determine the EMGT from a single incremental wheelchair propulsion bout. Moreover, EMGT largely agrees with VT, and there was a high similarity in EMGT readings between the major muscle groups around the shoulder.

EMGT has been shown to successfully predict muscular fatigue during cycling and treadmill running in able-bodied subjects [9, 10, 14, 15]. The analysis of the EMG-intensity curve in the present study has shown a nonlinear increase in EMG intensity during incremental wheelchair propulsion. Previous studies with able-bodied subjects have shown the occurrence of a breakpoint (the "EMGT") at which the increase in EMG of lower-extremity muscles becomes nonlinear during exercise protocols comparable with that used in the present study. The EMGT has been shown to occur during the onset of muscular fatigue, at about 70–80% of VO_2 max in elite cyclists [9, 23], about 65–70% of VO_2 max in able-body participants [24], and about 60% of VO_2 max in cardiac transplant patients [25]. In the present study, the EMGT occurred at about 52–58% VO_2 peak in wheelchair users, and the corresponding RPE values (both RPE_respiration and RPE_arm) were about 11 (Table 2). This might be attributed to the smaller muscle mass activated during arm exercise compared with leg cycling [26]. Individuals with spinal cord injury or spinal cord disease are generally less physically active than their able-bodied counterparts. Furthermore, possible differences in muscle fiber composition cannot be excluded, the upper-extremity muscles comprising a greater percentage of fatigable type II fibers [13, 27].

Activation patterns were assessed for each muscle individually since the response to fatigue might reflect the muscle's particular anatomy and biomechanical function within the synergy. In previous studies, activity of propulsive and recovery muscles increased with the progression of fatigue during wheelchair propulsion, but it increased more in propulsive muscles [20]. Musculoskeletal modeling and forward dynamics simulations of wheelchair propulsion also suggested compensatory strategies to overcome weakness in individual muscle groups during extended wheelchair propulsion [18]. However, the results of the present study show a high similarity (ICC = 0.91) in the occurrence of EMGT among AD, MD, PD, UT, PM, BB, TB, and IS. The onset of fatigue detected by EMGT reflects the increased firing rate and the progressive recruitment of additional faster motor units over time [6]. The present results may indicate that the relation between rates of increase in firing rate and/or motor unit recruitment and the

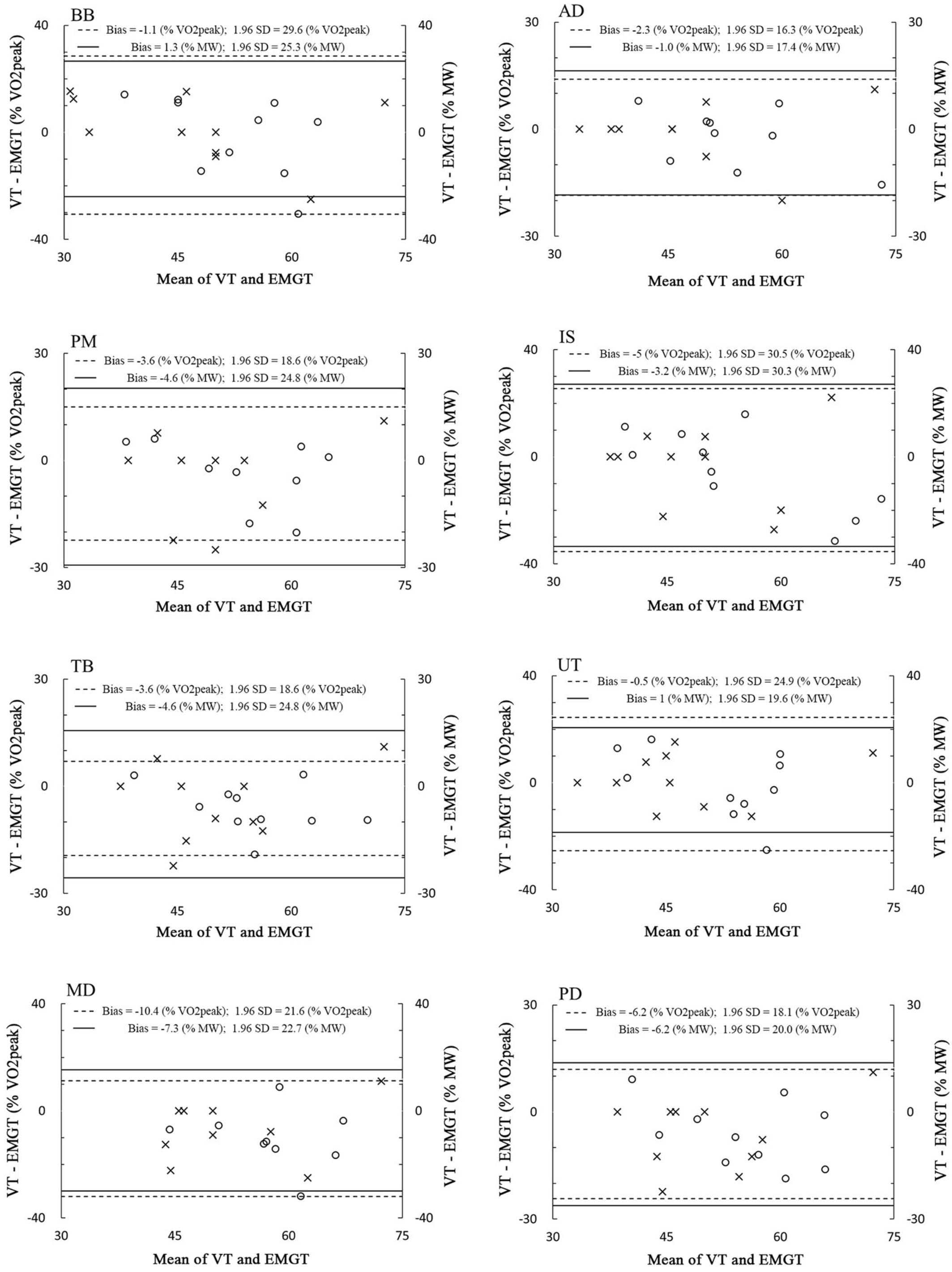

Fig. 2 Graphical analysis of EMG threshold corresponding to the ventilatory threshold (VT). *Y* axis shows the difference between the two paired measurements (VT–EMGT), and the *X* axis represents the average of these measurements (mean of VT and EMGT). In total, 95% of the data points lie within ±1.96 standard deviation (SD), which shows a good agreement between VT and EMGT. Double *Y* axes are used; the occurrence of VT and EMGT was normalized as % VO_2 peak (VT and EMGT are shown as circles, left *Y* axis) and % maximum workload (VT and EMGT are shown as *x*, right *Y* axis). The dotted line represents the 1.96 SD of the differences between the two methods (EMGT vs. VT, normalized as % VO_2 peak). The solid line represents the 1.96 SD of the differences between the two methods (EMGT vs. VT, normalized as % VO_2 peak).

onset of muscle fatigue may be similar between propulsive and recovery muscles during the wheelchair incremental test. Since an EMG signal combines impulses from the central and the peripheral nervous systems, it would be difficult to ascribe the near uniformity of shoulder muscle EMGT values directly to peripheral impulses originating in the individual muscles around the shoulder joint. Housh et al. reported that the EMGT occurred at the same power output for the superficial muscles of the quadriceps during incremental cycle ergometry [28]. Lucía et al. also reported that the EMGT of vastus lateralis and rectus femoris occurred at a similar exercise intensity [23]. In the present study, EMGT occurred at similar % VO_2 peak and % maximum workload in both propulsive and recovery muscles during the wheelchair incremental test, which means a single muscle EMG recording would be able to detect the onset of muscle fatigue in an eventual wearable autonomous fatigue detection system.

Muscle fatigue, particularly when it goes undetected, can put users of assistive rehabilitation technologies at risk of injury. It is important to include detection of muscle fatigue when designing appropriate rehabilitation programs for wheelchair users. By identifying EMGTs, it is possible to predict when muscle fatigue will occur, which provides the foundation of an automated system. Automating muscle fatigue detection/prediction through wearable devices could play an essential role in the design of assistive rehabilitation technologies. Real-time muscle monitoring and fatigue prediction can be realized through computational modeling based on the muscle fatigue threshold. Future studies will focus on developing automated fatigue detection/prediction systems using machine-learning algorithms to improve wheelchair users' daily performance and to avoid injury.

Limitations

The present study has shown that the EMGT can be detected from the upper-extremity muscles during a single incremental test in the wheelchair ergometer. While the timing of the onset of fatigue may be similar between muscles, there may be differences in the slopes themselves. It would be interesting to investigate the slope changes of EMG activity between muscles in the future studies. Extrapolation of our findings to other populations would, however, be hampered by the small size and homogeneous makeup of our participants. The strenuous incremental test in particular made it necessary to recruit generally fit participants. To see then whether more wheelchair users with different levels of fitness could benefit from these findings, validation within each additional group would be desirable.

Conclusion

In wheelchair users, the EMG activities of eight shoulder muscles (AD, MD, PD, IS, UT, PM, BB, and TB) show similar patterns, with a distinct breakpoint, EMGT, occurring at an exercise intensity of 52–58% VO_2 peak and 46–56% maximum workload. The EMG fatigue threshold method seems to be both valid and reliable for identifying the onset of muscular fatigue during wheelchair propulsion exercise. Future studies might focus on automating muscle fatigue detection/prediction in wearable devices (with a single EMG sensor). An automated muscle fatigue detection/prediction system would be used to enhance daily-life wheelchair performance and to avoid injury.

Data availability

The datasets generated during and/or analyzed during the current study are available in Mandelay at LQ (2019), "Fatigue threshold," Mendeley Data, v1 https://doi.org/10.17632/ccpyhysw7h.1.

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Author contributions LQ was responsible for designing the experimental protocol, conducting the search, screening potentially eligible studies, extracting and analyzing data, interpreting results, and writing the paper. LZ was responsible for screening potentially eligible studies. She contributed to writing the report, extracting and analyzing data, and interpreting the results. XBL contributed to extracting, analyzing data, and interpreting the results. MFP was responsible for designing the experimental protocol, conducting the search, screening potentially eligible studies, and interpreting the results.

Compliance with ethical standards

Conflict of interest The authors declare that they have no conflict of interest.

Ethical approval We certify that all applicable institutional and governmental regulations concerning the ethical use of human volunteers were followed during the course of this research.

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