Instantaneous quantification of skeletal muscle activation, power production, and fatigue during cycle ergometry

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EXERCISE TOLERANCE IS THE ABILITY TO PRODUCE FORCE OR POWER ADEQUATE TO ACCOMPLISH A TASK. MAINTENANCE OF A HIGH QUALITY OF LIFE IS DEPENDENT ON ADEQUATE EXERCISE TOLERANCE TO PERFORM THE ACTIVITIES OF DAILY LIVING. IN ADDITION TO QUALITY OF LIFE, EXERCISE TOLERANCE IS ALSO STRONGLY PREDICTIVE OF MORTALITY, IN BOTH HEALTH AND DISEASE (5, 25, 26, 37). WHILE WHOLE BODY ENDURANCE EXERCISE TOLERANCE CORRELATES WITH THE MORPHOLOGICAL STRUCTURES AND PHYSIOLOGICAL PROCESSES THAT SUPPORT OXYGEN FLUX BETWEEN THE ATMOSPHERE AND THE MUSCLE MITOCHONDRION, THE INTEGRATED NEUROMUSCULAR MECHANISMS DETERMINING TASK FAILURE REMAIN POORLY UNDERSTOOD. THIS IS LGELY DUE TO THE COMPLEXITY IN MEASURING AND LOCALIZING THE CAUSES OF FATIGUE DURING WHOLE BODY EXERCISE.

Fatigue, an exercise-induced reduction in power or force that is reversible with rest, may be generally categorized as follows: 1) a reduction in efferent activity of the motor cortex, spinal cord, or motoneurons that innervate the skeletal muscles; and/or 2) a disruption to skeletal muscle excitation-contraction coupling from depletion of energy stores or accumulation of metabolites. These are often referred to as “central” and “peripheral” fatigue, respectively. Central and peripheral contributions to fatigue are commonly identified using surface stimulation of the muscle, the peripheral nerve (PNS; magnetic or electrical), or the motor cortex [transcranial magnetic stimulation (TMS)], with or without combined maximal voluntary contraction (MVC; i.e., twitch interpolation). Due to the localized nature of these stimulation techniques, they provide valuable information on corticospinal (motor cortex to spinal nerves), neuromuscular (lower motor nerve to neuromuscular junction), and skeletal muscle fatigue, and knowledge of these can be used to predict patient outcomes (7, 9, 14–16, 19, 21, 27, 34, 38).

The nature of external stimulation necessitates isolated, single-joint isometric muscle contraction. Additionally, the proportion of force that can be generated through surface stimulation, PNS, or TMS alone is a small fraction of MVC force (~10–40%), even under potentiated twitch or when paired stimuli are used (38). Thus these maneuvers have little in common with complex, coordinated, dynamic contractions that provide power for daily tasks.

An idealized approach for studying fatigue, therefore, is to measure power generation during an exercise task that emulates the physiological conditions under which limitation is manifest (9). However, measuring fatigue during a whole body dynamic exercise task (e.g., that achieves maximal rates of ventilation, oxygen uptake, or cardiac output) is extremely challenging, not least because the velocity dependence of muscle power output (3, 4) confounds fatigue assessment made with traditional ergometry at the limit of tolerance (8, 23, 24). The alternative, whole body exercise followed by external stimulation, often introduces a crucial delay, resulting in a loss of information of the fatiguing processes due to the rapid recovery dynamics of both neuromuscular and corticospinal fatigue (14, 16, 30, 36). Consequently, an ideal assessment of fatigue during whole body exercise would 1) be task and...
velocity specific; 2) be able to be applied instantaneously during or following exercise; and 3) quantify the muscle activation and muscle fatigue contributions to the total fatigue process.

We reasoned that a fatigue assessment during cycle ergometry, using a rapid switch from cadence-independent (hyperbolic) to maximal-effort isokinetic cycling with crank power measurement (10), coupled with electromyographic (EMG) measurements of leg muscles activation, would meet these requirements. Isokinetic ergometry allows the velocity-specific decline in power production to be measured instantaneously during exercise, avoiding the confounding variations of contraction velocity present during standard ergometry (3, 30). Muscle EMG allows changes in motor unit action potential trains during a maximal voluntary effort to be measured. Thus, were the relationship between isokinetic power ($P_{\text{iso}}$) and surface EMG activity during cycling to be predictable with known confidence and reproducible in a given subject, then a departure from the baseline (fatigue-free) EMG-$P_{\text{iso}}$ relationship during or after fatiguing exercise would allow for muscle “activation fatigue” (AF; a reduced muscle activity) and “muscle fatigue” (MF; a reduced power at a given muscle activity) to be isolated. In other words, were power during fatigue to be less than expected at the measured muscle activity, then this deficit could be ascribed to MF, with any remaining power reduction being consequent to AF.

We hypothesized that the baseline (nonfatigued) EMG-$P_{\text{iso}}$ relationship would be linear, velocity dependent, reproducible, and sufficiently precise to provide a reference for comparison with instantaneous EMG-$P_{\text{iso}}$ during cycle ergometry. This would allow an individual’s muscle AF and MF to be measured during, or following, whole body exercise to the limit of tolerance.

MATERIALS AND METHODS

Ethical Approval and Participants

The Institutional Review Board, Los Angeles Biomedical Research Institute at Harbor-UCLA Medical Center, approved this study, and all procedures complied with the latest revision of the Declaration of Helsinki and Belmont Report. Written, informed consent was obtained from healthy, active volunteers [$n = 13$, $42 \pm 14$ yr (range 29–72 yr), $171 \pm 8$ cm, $75 \pm 14$ kg, peak rate of pulmonary O$_2$ uptake ($V\dot{O}_2_{\text{peak}}$) $3.2 \pm 0.7$ l/min] before their participation in the study. Volunteers were screened for cardiovascular disease risk with a resting ECG and a health history questionnaire.

Exercise Protocol

Volunteers visited the laboratory on two occasions for duplicate measures of an identical protocol. On each visit, the participants completed two phases: 1) short ($\sim 5$ s) bouts of variable effort isokinetic cycling at three pedaling frequencies; and 2) a ramp-incremental exercise test, followed by a short ($\sim 5$ s) maximal effort isokinetic bout performed immediately at the limit of tolerance and at each minute of recovery.

Measurement of baseline EMG-$P_{\text{iso}}$. Volunteers cycled on an ergometer (Lode BV) with pedaling rate constrained at 50, 70, or 100 rpm (isokinetic). An example of the protocol at 70 rpm is shown in Fig. 1. Participants were asked to give four to five variable efforts at $\sim 25$, 50, 75, and 100% of maximum effort. Each of the efforts lasted for $\sim 3$ to 5 s and were separated by $\sim 0.5$ to 5 min of unloaded cycling (longer recovery following the maximal efforts). The process was repeated two to three times at each pedaling frequency (50, 70, and 100 rpm).
100 rpm). The mean EMG (Fig. 1A) and isokinetic crank power (Fig. 1B) at each pedaling frequency (Fig. 1C) were determined from three consecutive isokinetic crank revolutions and used to model the baseline (fatigue-free) velocity-dependent EMG-Piso relationship (Fig. 1D) (see EMG-Piso, Relationship and Fatigue Characterization for details of the analysis).

**Measurement of EMG-Piso following ramp exercise.** Following a rest period of ~20–30 min, participants returned to the ergometer to complete a ramp incremental test. This test consisted of ~2 min at rest, ~4 min of 0 W cycling, a ramp phase of 15–30 W/min until the limit of tolerance, followed by recovery at 0 W. During the ramp phase, the ergometer power was cadence independent (hyperbolic). The limit of tolerance was defined as being unable to maintain a frequency above 55 rpm, despite strong encouragement. At the limit of tolerance, the ergometer was switched instantaneously to the isokinetic mode constrained to 70 rpm. This instantaneously reduced the flywheel breaking to zero (because cadence was below 70 rpm at the limit of tolerance), and volunteers were strongly encouraged to give a maximal effort lasting ~5 s; thus the flywheel was rapidly (within ~1–2 s) accelerated to, and held at, 70 rpm while crank power was measured for 4–5 revolutions. This maneuver is similar to the baseline maximal effort isokinetic bout during Measurement of baseline EMG-Piso, with which the participants were well familiarized. A maximal ~5 s isokinetic bout at 70 rpm was repeated each minute during recovery until Piso was similar to the baseline.

Eleven of thirteen participants completed duplicate laboratory visits separated by at least 48 h. This duplicate visit included both phases for Measurement of baseline EMG-Piso and Measurement of EMG-Piso following ramp exercise.

**Ergometry**

All exercise tests were undertaken on a computer-controlled electromagnetically braked cycle ergometer (Excalibur Sport PFM, Lode BV, Groningen, the Netherlands). In addition to the standard application and measurement of power at the electromagnetically braked flywheel, the ergometer was instrumented with force transducers in the bottom bracket spindle. Left and right torque (Nm) was measured independently (peak force 2,000 N, < 0.5 N resolution, and measurement uncertainty of < 3%); Instantaneous angular velocity of the crank (rad/s) was measured with a resolution of 2° using three independent sensors sampling in series (measurement uncertainty of < 1%). Power (W) was calculated every 2° from torque and angular velocity measurements. There was no systematic difference in the power production between the left and right cranks, and therefore power from the right crank only was used (10) and averaged over three crank revolutions to provide a paired datum with EMG from the same leg (described below).

**EMG**

Surface EMG was measured in five muscles of the right leg: vastus lateralis, rectus femoris, vastus medialis, biceps femoris, and gastrocnemius lateralis. Placement sites were shaved, abraded with gauze, and cleaned with 70% volume isopropyl alcohol. Wireless transmitting Ag bipolar parallel-bar surface electrodes (Trigno Wireless System, Delsys, Boston, MA) were placed over the muscle belly using Surface Electromyography for the Non-Invasive Assessment of Muscles (SNIAM) recommendations. Electrodes were placed over the vastus lateralis two-thirds of the distance from the anterior superior iliac spine to the lateral side of the patella, over the rectus femoris halfway between the anterior superior iliac spine and the superior border of the patella, over the vastus medialis eight-tenths of the distance from the anterior superior iliac spine to the joint space in front of the anterior border of the medial ligament, over the biceps femoris halfway between the ischial tuberosity and lateral epicondyle of theibia, and over the gastrocnemius lateralis one-third the distance between the head of the fibula and the calcaneus. The longitudinal axis of the electrode was aligned parallel to the long axis of the muscle. EMG signals were differentially amplified and sampled at 2 kHz with 16-bit resolution. Each sensor had a signal bandwidth of 20–450 Hz and common mode rejection ratio of >80 dB. During postprocessing, signals were filtered with a second-order Butterworth bandpass filter (3 dB, 10–500 Hz) and smoothed via root mean square (RMS) with a 100-ms moving window with no overlap. The peak activity (µV; from the 100-ms RMS) from each crank revolution was used as an estimate of muscle activity. The earliest three consecutive isokinetic crank revolutions that were appropriately constrained at the desired angular velocity were identified in the output from the cycle ergometer, and the peak RMS EMG from these were ensemble averaged for each muscle; these were typically the second, third, and fourth crank revolutions after switching to isokinetic cycling. The RMS EMG values from the five muscles of the right leg were averaged to provide an EMG datum to pair with Piso produced at the crank from the same leg. The muscle selection reflected the weighted power contributions from knee extension/flexion and plantar flexion (12).

**EMG-Piso, Relationship and Fatigue Characterization**

The characterization of the EMG-Piso relationship is specific to the electrode placement (skin preparation, conduction, etc.), and therefore the baseline “physiological normalization” (e.g., Fig. 1D) was repeated at each laboratory visit, and the RMS EMG values were normalized to the visit maximum. The baseline relationship between power production and EMG activity (Measurement of baseline EMG-Piso) was characterized using linear regression at each angular velocity.

**Pulmonary Gas Exchange**

Respired gases were measured breath by breath with a commercial metabolic measurement system (Vmax Spectra, Sensormedics, Yorba Linda, CA). The system was calibrated immediately before each testing session. A 3-liter syringe (Hans Rudolph, Shawnee, KS) was used to calibrate the flow sensor (hot-wire anemometer) from ~0.2 to 8 l/s, mimicking flow rates expected at rest and during exercise. The CO2 and O2 analyzers were calibrated using gases of known concentrations (O2 26.0 and 16.0%; CO2 0.0 and 4.0%). ECG (Cardiosoft, GE Healthcare, Little Chalfont, UK) and finger pulse oximetry (Radical-7, Masimo, Irvine, CA) were monitored throughout exercise.

**Statistical Analyses**

Means were compared, where appropriate, with t-tests, ANOVA, or repeated-measures ANOVA. Analysis of covariance (ANCOVA) was used to assess the velocity dependence of each individual’s baseline EMG-Piso relationship. The standard error of the estimate (SEE) of the baseline EMG-Piso regression was calculated as an index of sensitivity of MF measurement. Bland-Altman plots were generated for an index of reproducibility (6). Statistical significance was determined at P < 0.05. Statistics were computed using IBM SPSS v20. Data are presented as means ± SD.
fatigue measurement is shown in Fig. 3. During the ramp incremental test, peak power measured at the flywheel was 261 ± 58 W, and $V_{O2}$ peak was 3.2 ± 0.7 l/min. As a group, $P_{iso}$ at the limit of tolerance (termed “R0”, for recovery at 0 min; 335 ± 88 W, two times one leg) was greater than flywheel power [261 ± 58 W, confidence interval difference (CI_Difference) 28, 120 W, $P < 0.05$], but only 55 ± 14% of baseline $P_{iso}$ (630 ± 156 W, CI_Difference 211, 380 W, $P < 0.05$).

A representative example of the baseline EMG-$P_{iso}$ relationship and $P_{iso}$ measured during fatigue is shown in Fig. 4. The difference in $P_{iso}$ between baseline and R0 determined PF (158 ± 80 W, one leg): on the EMG-$P_{iso}$ plot (Fig. 4A), PF is the total displacement in the power (y) dimension. There was a significant inverse relationship between $V_{O2}$ peak and PF normalized to baseline peak $P_{iso}$ ($r^2 = 0.49$; $P < 0.05$), i.e., a greater oxidative capacity associated with lesser relative PF at the limit of ramp incremental exercise (Fig. 4C).

The baseline EMG-$P_{iso}$ relationship was then used to characterize AF and MF in each participant at the limit of tolerance (R0) and the first 3 min during recovery (R1, R2, R3). A representative example is shown in Fig. 4B, and group data in Fig. 5. PF is rank-ordered in Fig. 5A, with AF and MF in the

**RESULTS**

**Measurement of Baseline EMG-$P_{iso}$**

As expected, participants’ peak single-leg $P_{iso}$ (measured at the right crank arm, mean over 3 revolutions) during baseline assessment was strongly cadence dependent: 232 ± 56 W at 50 rpm; 307 ± 75 W at 70 rpm; and 373 ± 114 W at 100 rpm ($n = 13$; $F(1.1,13.3) = 49.9$; $P < 0.05$; $\eta^2 = 0.81$; all Bonferroni post hoc comparisons $P < 0.05$; Fig. 2). Importantly, individual EMG-$P_{iso}$ relationships were strongly linear over the ranges measured (between ~25 and 100% effort, $r^2 = 0.95 ± 0.04$). Individual EMG-$P_{iso}$ slopes differed across the three pedaling cadences ($P < 0.01$; ANCOVA) (Fig. 2) in all but one participant ($P = 0.07$; ANCOVA). The precision of the EMG-$P_{iso}$ regression at 70 rpm was determined using SEE. These values were used to estimate the “sensitivity” to detect a difference from baseline of a single EMG-$P_{iso}$ measurement during fatigue. The SEE was not different among cadences [$F(2,24) = 0.2$; $P > 0.05$; $\eta^2 = 0.02$], and averaged 4.2 ± 2.0, 4.5 ± 2.2, and 4.2 ± 2.6% of peak $P_{iso}$ at baseline, at 50, 70, and 100 rpm respectively.

**Measurement of EMG-$P_{iso}$ Following Ramp Exercise**

A representative example of crank power and cadence during the ramp incremental test and instantaneous isokinetic

**Fig. 2.** The velocity dependence of the isokinetic EMG-$P_{iso}$ relationship. A: representative participant values for EMG-power relationship at 50 (●), 70 (○), and 100 rpm (▲). B: group mean values for the isokinetic EMG-$P_{iso}$ relationship at 50, 70, and 100 rpm. Error bars are SD. *Significant ($P < 0.05$) difference for slope across pedaling frequencies via analysis of covariance.

**Fig. 3.** Representative participant data during a ramp incremental test terminated with a maximal effort, 5-4, isokinetic bout. The dotted lines indicate that data are continuous, where the x-axis is expanded to better visualize power. Gray shading along the x-axis represents hyperbolic ergometry, or cadence independent power. Black shading along the x-axis represents isokinetic ergometry at 70 rpm. A: average crank arm power measured during ramp and isokinetic exercise. B: pedaling frequency (velocity) measured throughout the experiment. During the isokinetic phase, cadence is constrained at the target velocity.
hatched and solid bars, respectively. AF and MF were 97/11006 and 60/11006 W, respectively. Fatigue assessment during recovery from intolerance revealed that P iso varied across time [n = 13; F(1.7, 19.9) = 44.1; P < 0.05; η² = 0.79] and was reduced (P < 0.05, Bonferroni) vs. baseline measures at R0 (CIDifference 80, 201 W) and R1 (CIDifference 13, 80 W) (Fig. 5B).

Reproducibility of the primary measurements required for fatigue assessment for 11 participants was determined from the mean bias and limits of agreement (LoA) using a Bland-Altman plot. Mean bias LoA were as follows: V˙O2 peak0.1 l/min (Fig. 6A); baseline Piso 130 W (Fig. 6B); Piso at R0 35 W (Fig. 6C); and EMG at Piso 14% (Fig. 6D).

DISCUSSION

Our main findings were that the relationship between leg muscle activity and Piso during cycle ergometry was linear and velocity dependent (Fig. 2) and was reproducible across visits. Using standard incremental exercise followed immediately by measurement of maximal voluntary Piso and leg muscle surface EMG, we were able to make sensitive assessments of the
proportion of PF that resulted from a reduction in maximal voluntary muscle activity (AF), and that from the reduction in muscle power production for the achieved muscle activity (MF) (Figs. 4 and 5). Additionally, we showed that the dynamics of power recovery are rapid (30), returning to baseline in \( \sim 2-3 \) min (half-time \( \sim 30-60 \) s; Fig. 5B). The reproducibility of the primary measurements was similar to the test-retest reproducibility of VO\(_{2}\) peak measurement, i.e., \( -10\% \) (1) (Fig. 6), suggesting that this new fatigue assessment is useful for localizing the variables that determine exercise intolerance during whole body exercise.

The EMG-P\(_{iso}\) Relationship During Isokinetic Cycle Ergometry

As a reference for the power production expected from a given muscle activity, we measured the relationship between surface EMG and P\(_{iso}\) in the rested, or baseline, state. We reasoned that, should this relationship be confidently predictable and reproducible, then we would be able to generate a “physiologic normalization” for an individual that could be used as a baseline comparator for measurements made in a fatigued state. The EMG-P\(_{iso}\) relationship was strikingly linear and highly reproducible over the range of \( -25\% -100\% \) maximal effort (Fig. 2), providing a robust and sensitive framework for determining fatigue components in whole body exercise.

The EMG-P\(_{iso}\) was significantly dependent on contraction velocity (steepening between 50 and 100 rpm). This reinforces that any fatigue comparisons need to be made at an isokinetic velocity (8). This velocity dependence of the EMG-P\(_{iso}\) relationship is expected from the parabolic nature of the muscle power-velocity curve. Motor activity is maximal at any point on the power-velocity curve; in this study, each individual’s absolute RMS EMG was not different across maximal efforts at each contraction velocity. Thus, maximal motor activity at 50 rpm gives rise to a lower power than at 100 rpm. The fact that the difference between EMG-P\(_{iso}\) slopes was greater between 50 and 70 rpm than between 70 and 100 rpm reflects the power-velocity curvature, and that at 100 rpm individuals are approaching the velocity at which P\(_{iso}\) is maximal (3).

Power Reserve at the Limit of Tolerance

Previous attempts to identify a reserve in power production at the limit of tolerance have led to the controversial conclusion that exercise is constrained not by inability to produce power, but by perception of effort, and that a large reserve in power production capacity is present (24). Unfortunately, this conclusion was heavily influenced by an ergometry protocol that did not control for contraction velocity (8, 23). Conversely, we showed a large (55%) reduction in peak P\(_{iso}\) from 630 W at baseline to 335 W at the limit of tolerance, which was associated with a relatively small (18 \( \pm 11\% \) of P\(_{iso}\)) reserve in power at the limit of tolerance (95% confidence interval 28, 120 W) in agreement with our laboratory’s previous work using isokinetic ergometry (13). Our data confirm that the limit of tolerance is reached with a severe depression in the locomotor muscle power-generating capacity, and this power is not substantially more than that produced at the end of ramp incremental exercise (cf., Ref. 24). The difference, 74 \( \pm 75\) W, may result from the \( \sim 1\)–2 s of recovery as the participant accelerates the ergometer flywheel from \( \sim 50\) rpm at the limit of tolerance in the ramp to the reference isokinetic velocity of 70 rpm. However, we believe that this difference more likely reflects (a very limited) capacity for short-term power production. In fact, large swings in instantaneous power, measured at the crank arm, were clearly visible during the ramp incremental phase (Fig. 3). In other words, the very smooth linear increase in power at the flywheel is not indicative of the rapid fluctuation in power measured at the crank arm. The flywheel power...
represents the mean power required for the task, while the fluctuations in crank power can exceed the flywheel power by >30%. In some participants, these swings amplify during the final minutes and seconds of ramp exercise as the motor system becomes increasingly challenged by the demands of the task, demonstrating some capacity for fleeting increases in power production as intolerance encroaches.

How local metabolic factors, afferent feedback, and motor system excitability combine to modulate muscle power and impose limits on exercise is unknown. However, our method provides information on the proportion of performance deficit that is attributable to MF: a reduction in power at the achieved maximal voluntary motor activation. The balance of the performance deficit, therefore, is due to a reduction in muscle activation. Reduced activation may result from feedback generated by the MF, or fatigue-related metabolites, or other physiological processes and perceptions contributing to symptom limitation, depending on the exercise condition or state of health (18). In this study of healthy volunteers there was a strong association between \( \text{VO}_{2 \text{peak}} \) and relative PF, suggesting that individuals with greater aerobic capacity are better protected from fatigue and are able to reach a greater proportion of their baseline peak \( P_{\text{iso}} \) before reaching intolerance. While the average contribution of MF to the performance deficit at the limit of tolerance was \( \sim 35\% \), there was variability among individuals in the MF fraction, e.g., participants 5, 7, 9, and 13 (Fig. 5A). None of the variables measured in this study could clearly explain these exceptions. Thus the next steps are to employ the method to 1) associate the magnitude of these fatigue components to physiological, morphological, or sensory differences among individuals; and 2) identify how interventions modify the components of fatigue, especially in patients with chronic cardiopulmonary and muscle diseases that predispose to poor exercise tolerance.

Recovery Dynamics and Alternatives for Measuring Fatigue

Central to our approach was finding an alternative assessment for fatigue that could be applied essentially instantaneously during or following whole body exercise. Alternative methods, such as PNS or TMS, necessitate isometric contraction and are limited to following immediately from similar single-joint exercise, such as knee extension (7, 14), and therefore rarely elicit maximal cardiorespiratory limits. Alternatively, the time delay of transferring from a treadmill or cycle ergometer to make the fatigue measurements limits the ability to assess the mechanisms of fatigue instantaneously at the point of limitation (2). Furthermore, the effect of partially potentiated twitch may confound the otherwise strong \( r^2 > 0.8 \) relationship with MVC-measured force (27), while fully potentiated twitch appears to be more robust and reproducible (22, 27).

Our data show that recovery of voluntary power production following cycling intolerance at \( \text{VO}_{2 \text{peak}} \) is restored rapidly back to near baseline levels within 2–3 min (30, 36). This is in agreement with previous isokinetic measurements (30) and the dynamics of supraspinal and neuromuscular fatigue recovery after single-leg knee-extensor exercise and magnetic/electrical simulation (14, 16) and conforms closely to the recovery dynamics of intramuscular fatigue-related metabolites accumulated during exercise (28, 29). In our data, peak \( P_{\text{iso}} \) at 2 min of recovery (R2) was restored back to baseline \( P_{\text{iso}} \) \( (P = \text{nonsignificant}) \), reinforcing the necessity for fatigue measurements to be made at, or following rapidly, the time of interest (14, 16).

Limitations

Our aim was to develop a method that could evaluate the AF and MF contributions to exercise intolerance during whole body exercise that elicits maximal cardiorespiratory strain. Although methods allowing external neuromuscular stimulation during whole body voluntary exercise are available (31–33, 35), to quantify muscle power and simplify the application of fatigue assessment we chose to rely on voluntary activation. Thus our measurements at the limit of tolerance, and in recovery, rely on the participant to give a maximal voluntary effort and are not independently verified by the imposition of controlled or standardized external stimuli. Naturally, the same argument could be made for most exercise tests. However, in our case, there is an opportunity to interpret a reduction in surface EMG and cycling power to some “AF” when it may result from a submaximal effort. This is of particular concern when no skeletal MF can be identified using the method and represents an opportunity for a “false positive” to identify AF. We do not yet have a solution to this limitation. However, we are confident in the reproducibility of both baseline and R0 \( P_{\text{iso}} \) measurements [LoA ± 10 W (\( \sim 1\% \) of baseline \( P_{\text{iso}} \)) and 20 W (\( \sim 6\% \) of R0 \( P_{\text{iso}} \), respectively]; the likelihood is low that a repeated submaximal effort could be delivered independently at \( \text{VO}_{2 \text{peak}} \) with such precision. It is likely that familiarization to maximal isokinetic efforts afforded by the baseline assessment is important in minimizing these LoA, and these brief efforts can be reliably repeated as needed. Nonetheless, this method remains a functional assessment dependent on a maximal effort by a well-motivated participant.

The EMG-\( P_{\text{iso}} \) relationship must be assessed at each placement of the EMG electrodes. Even with indelible markings on the skin, changes in skin temperature, hydration, skin preparation, and conductance, rule out comparisons between days without normalization in the EMG axis. We found that there was generally little difference in absolute EMG values within an individual between visits; however, this was not the case for every test. Thus a “physiologic normalization” needs to be performed on each visit by each participant, adding time and effort to the exercise test. Nevertheless, using this approach, we found good agreement between EMG at R0 after normalization (LoA ± 14%). It is worth noting that the fatiguing exercise bouts in this study were relatively short (\( \sim 10 \text{ min} \)). We do not know whether longer bouts of exercise that result in large core and muscle temperature change might result in a distortion of the EMG-\( P_{\text{iso}} \) relationship.

Finally, the effects of amplitude cancellation and changes in signal conduction velocity with fatigue might partly explain the variability in EMG (Fig. 6D). Here we use EMG amplitude to reflect muscle activity. However, amplitude cancellation increases act to reduce the apparent EMG amplitude for a given muscle activation, the magnitude of which is influenced by fatigue. Conduction velocity may also change with fatigue, increasing or decreasing EMG amplitude for a given muscle activation. In addition, these effects interact with each other variably, depending on %MVC. These effects confound interpretation of EMG amplitude either in terms of muscle activity...
or as an index of neural drive (11, 20). During cycling, EMG activity in the knee extensors during a single maximal pedal stroke is ∼25% of knee extension MVC (17). Even in normalized EMG at relatively low power, some degree of amplitude cancellation is likely present. Nevertheless, at ∼25% MVC, EMG amplitude is relatively unaffected by conduction velocity compared with high %MVC maneuvers (11), making predictions about muscle activity from EMG somewhat less problematic. While it is tempting to conclude that EMG amplitude is an indicator of the central motor drive, and this has often been the interpretation used in the past, it is currently not certain how the motor input is related to the EMG signal.

Conclusion

Establishing an individual’s isokinetic EMG-power relationship for fatigue assessment is a reliable addition to a standard cycle ergometer exercise test. This relationship is linear, velocity dependent, and reproducible. The method is sufficiently sensitive to quantify the muscle AF and MF components of PF during whole body exercise. Deviation from the baseline isokinetic EMG-power relationship during or following exercise can provide insight on the nature of exercise intolerance in health and disease.

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DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the author(s).

AUTHOR CONTRIBUTIONS


REFERENCES


