EVIDENCE FOR THE INVALIDITY OF THE WINGATE TEST FOR THE ASSESSMENT OF PEAK POWER, POWER DECREMENT AND MUSCULAR FATIGUE

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Abstract. We hypothesized that the protocol-induced initial cadence of the WAnT is too high to allow high muscle force production and peak power generation. Twenty endurance, strength or power trained subjects (9 male, 11 female) completed two 30 s maximal exertion stationary cycle ergometer tests involving the traditional peak cadence start (TRAD) vs. a stationary start (STAT). Inertia corrected mechanical power, cadence, EMG from the vastus lateralis, and applied force to the pedals were measured continuously throughout both tests. Peak power was higher during TRAD; 11.32 ±1.41 vs. 10.40 ±1.35 Watts/kg ($p < 0.0001$), as was peak cadence; 171.4 ±16.3 vs. 120.9 ±15.1 rev/min ($p < 0.0001$). However, during TRAD EMG root mean squared (rms) increased continuously throughout the test, force applied to the pedals increased from 1 to 3 s (0.73 ±0.27 vs. 0.90 ±0.39 N/kg; $p = 0.02$) and thereafter remained relatively stable. EMG mean frequency also increased from 1 to 3 s, but then decreased throughout the remainder of the test. During TRAD, mechanical power decreased near immediately despite increasing EMG rms, EMGmean frequency and force application to the pedals. The initial 10 s of data from the WAnT is invalid. We recommend that intense cycle ergometer testing should commence with a stationary start.

Key words: ergometry, fatigue, force, cadence, motor unit recruitment

Introduction

The Wingate Anaerobic Test (WAnT) was devised in the early 1970's, and since this time has been widely used and assumed to be a valid and reliable assessment of mechanical peak power and power decrement during maximal effort cycle ergometry (Bar-Or 1987; Dotan and Bar-Or 1983; Dotan 2006; Inbar et al. 1996). The WAnT requires that a subject first increase cycling cadence to a peak rate with no external loading, and the test commences with the application of a gender and body mass adjusted predetermined load with the subject attempting to sustain maximal cadence cycling for 30 s. The two main variables of interest from the WAnT are peak power and the power decrement over the 30 s test, with the latter also expressed as a relative fatigue index (FI; % decrement).

Research of the WAnT has established recommended strategies for gender specific load determination (Dotan and Bar-Or 1983), corrections to mechanical power measurements for the moment of inertia of the ergometer flywheel (Bassett 1989), estimated contributions from muscle metabolic energy systems to the ATP turnover of repeated intense muscular contractions (Beneke et al. 2002; Smith and Hill 1991; Withers et al. 1993), documenting suitability for or improvement from specific types of intense exercise (Maud et al. 1989; Zupan et al. 2009), and for furthering knowledge about the mechanisms of exercise-induced fatigue (Fernandez-del-Olmo et al. 2011). Such broad applications of the test have also led to its widespread use in high school and university education, and for elite athlete assessment in most of the world's Olympic or professional athlete/sports training centers and institutes.

Application of the WAnT to the study of muscular fatigue is especially important to the purpose of this research study. Data from the WAnT reveals an instantaneous peak power at test onset, immediately followed by a rapid decrease in mechanical power output (Figure 1a). The same trend occurs for cadence (Figure 1b); as for any constant load ergometry test cadence will drive the change in power calculations. These rapid changes in mechanical power and cadence have been interpreted as muscle fatigue, defined in this context as a decrease in mechanical power output despite continued maximal effort. To date, no research has been able to explain the cause of this near instantaneous fatigue response during the WAnT.

We questioned the validity of the instantaneous peak power response of the WAnT, and moreover, the interpretation that the subsequent drop in power indicates immediate muscle fatigue. For example, application of a general understanding of muscle energy systems is inconsistent with such an immediate and rapid power decrement. The phosphagen energy system has the capacity to provide rapid ATP replenishment for repeated intense muscle contractions for at least 10 s, and indeed, such rapid ATP buffering for a more sustained maximal power output has been shown during isokinetic cycling (Jones et al. 1985) as well as from invasive research involving artificial electrical stimulation of the human quadriceps muscles (Spriet et al. 1987; Spriet et al. 1987b). Added scrutiny then led to our concern over the initial 5 s unloaded segment of the protocol that occurs immediately prior to test onset. For example, during the unloaded preparation phase of the test, subjects can attain a peak cadence approximating 170 rev/min, computing to average single leg crank angular velocities of 510°/s. Prior research of the mechanics of cycling has consistently revealed that the optimal cadence for the development of peak mechanical power approximates 100 to 120 rev/min (Dorel et al. 2010; Hautier et al. 1996; Katch et al. 1977; MacIntoch et al. 2000; MacIntosh et al. 2003; Reiser et al. 2003; Samozino et al. 2007). Similarly, there is evidence that the velocity at peak power is highly influenced by muscle motor unit proportions (Hautier et al. 1996). In short, empirical evidence reveals that it is impossible for muscular contractile power to be at a peak value at the onset of the WAnT where cadence can exceed 170 rev/min.

Based on prior research evidence and the application of logic, we hypothesized that the unloaded preparation phase of the WAnT allows cadence to increase to values inconsistent with optimal time for sufficient motor unit recruitment to generate peak muscular force and power. Consequently, the peak power of the WAnT is likely to be protocol dependent, and, therefore, independent of true physiological muscular power.

Given the possible invalid start to the WAnT and the subsequent erroneous measure of peak power and other measurements dependent on peak power, it is interesting to examine the prior results of modified versions of the WAnT that employ a stationary start (Jastrzębski 1995; MacIntosh et al. 2003). Using such a stationary start, Jastrzebski (1995) demonstrated a power curve that revealed a steady increase in power from the start to peak cadence which occurred in 6 s, followed by a near linear gradual decrement. More recently, MacIntosh et al. (2003) compared the stationary start to the traditional “flying” start method of the WAnT. However, they differed in their traditional start methodology by using a 15 s preliminary period of increased cadence rather than a 5 s period, and resistance was based on force transducers attached to the friction belt of an ergometer rather than a continuously applied load. No detail was provided for the electronic load application of their ergometer. Such differences were influential to the results, as there was an average time to peak power of 5.63 s for their traditional method, which revealed that resistance was not constant or independent of cadence for their equipment. Stationary start methods of the test resulted in time to peak power of between 3.3 to 3.8 s, with cadence at peak power data not reported.

Based on these prior research findings, the unclear methods and results from MacIntosh et al. (2003), and our hypotheses for explaining the invalidity of the WAnT, the purpose of this study was to compare mechanical power, EMG signals from the vastus lateralis, and pedal force responses during the WAnT and a stationary start version of the test. We further hypothesized that pedal forces and muscle EMG data would both reveal increasing signals during decreasing mechanical power output and cadence. Collectively, such results would reveal that during the initial mechanical power decrement phase of the WAnT there is increasing EMG activity and increasing muscular force development, both of which oppose the definition of muscular fatigue. The peak power and subsequent early power decrement of the WAnT would therefore be artificial as they are caused by an unacceptably high cadence at test onset. Added variables dependent on peak power, such as the fatigue index, would also be invalid.

Materials and methods

Subjects

Subjects were recruited from university varsity athletic teams, university student populations, and interested adults from the local community. Prior to involvement in the study, subjects read, were verbally explained, and then signed an institutional approved informed consent. Subjects also completed a health history questionnaire.

Subject inclusion criteria consisted of current exercise training at least 3-days/week, familiarity with cycling, an age <45 years for males and <50 years for females, no evidence of family history of cardiovascular, respiratory, or metabolic diseases or cancer, and no evidence of current musculoskeletal injury or systemic illness.

A priori power estimation was performed using a freely available computer program (GPower 3.01, Henirich Heine Universitat, Dusseldorf, Germany) and revealed the need for 16 subjects at $p = 0.05$ and $d = 0.8$ for a two group repeated measures ANOVA design. We increased this sample size to 20 to account for subjects that may elect to withdraw from the study, or need to be removed due to equipment failure or errors during data acquisition.

Procedures

After giving written informed consent, the subjects were scheduled to arrive at the laboratory for all test sessions in a given day. For subjects who had not experienced the completion of a WAnT, a preliminary session was arranged to practice the 30 s test to ensure subject compliance to the maximal effort requirement of the test. After the signing of the consent, subjects had their weight (SECA 884, Hamburg, Germany) and height (custom built stadiometer) measured, and their current exercise training involvement was recorded.

Measurement of Surface Electromyography (EMG)

After arriving at the laboratory for the WAnT trials, the subjects were first prepared for bilateral surface electrode EMG of the vastus lateralis, vastus medialis and rectus femoris muscles. Surface electrodes (Vermed ECG Sensor Electrodes, Bellow Falls, VT USA) were placed during full extension of the legs along the central muscle belly in alignment with the muscle architecture with a center-to-center inter-electrode distance of 4.8 cm. The ground electrode was placed at the medial tibial plateau. Skin impedance was reduced by careful skin preparation consisting of shaving with a dry razor, light abrasion with a gauze pad, and cleaning with an alcohol swab. Electrodes and lead wires were held in place by having the subjects wear stretch mesh stockings. During exercise, surface EMG signals were acquired at 1,500 Hz using a commercially available bipolar EMG system (Telemyo 2400 G2, Noraxon , Scottsdale, Az).

Applied Pedal Force Measurement

Force applied to the pedals was measured by a series of 4 piezo-resistive force/pressure sensors sandwiched between an aluminum base plate and top Perspex cover secured to each pedal (FlexiForce, Tekscan South Boston, MA). The sensors respond to applied force with a decrease in output resistance. The resistance signal was converted to output voltage via a custom developed electronic circuit as recommended by the manufacturer. The force sensors in the pedals were calibrated over an 80 kg load range immediately prior to each subject test to pre-load and condition the sensors, derive a voltage:force regression equation, and allow subsequent force data conversion to Newtons.

The angular rotation of the crank was quantified by the placement of a “U” shaped (slot type) photointerupter (Omron Electronics, Sydney, Australia) on a metal housing fixed to the frame of the ergometer. The sensor frame was placed so that each tooth of the crank wheel passed between the sensor frame causing a decrease in voltage as the infrared light beam path was interrupted. For angular reference, another “U” shaped sensor was fixed to a different location of the frame so that a metal tab fixed to the inside of the crank wheel passed through the sensor frame once each crank cycle. A 360° rotary encoder (Omron Electronics, Sydney, Australia) was connected to the rod of the pedal, and supported by connection to a frame mounted to the pedal housing. This allowed quantification of pedal rotation. Force and rotational sensor signals were acquired at 2,000 Hz using a custom developed program (LabVIEW, National Instruments, Austin, TX).

Ergometer Testing

Subjects performed two different versions of the WAnT on a mechanically braked cycle ergometer (Monark 824E Ergomedic, Varberg, Sweden) based on a balanced pre-determined Latin Squares design. The two test versions were the traditional WAnT (TRAD) and a stationary start version (STAT). A 30 min passive rest period separated trials, resulting in at least a 45 min test end to second test start time interval. Each testing session began with a standardized 5 minute warm up at 1 kg resistance and cadence of 70 rev/min. The seat and handlebar positions were adjusted to comfort for each subject prior to warm-up and the same subject specific settings were used for each trial. Subjects were instructed to remain in the seated position for the entire test.

For the TRAD trial, subjects increased pedal cadence to maximal revolutions with no resistance over 5 s at which time the resistance was applied equivalent to 7.5% of body mass for females and 9.1% of body mass for males (15). Subjects were verbally encouraged to sustain their highest cadence possible throughout the 30 s test. Data was acquired via a commercial software program (SMI Opto Sensor 2000™, Sports Medicine Industries, St Cloud, MN) that collected data for pedal cadence from an optical sensor attached to the front frame of the flywheel and computed ergometric data at 1 Hz. Data from TRAD were adjusted for flywheel inertia by the commercial software. At the end of the 30 s, subjects continued pedaling at 1 kg resistance for 5 min as a standardized cool-down, followed by self selected recovery comprising walking, seated rest, or supine rest. EMG electrodes were left on between trials.

For the STAT trial, subjects remained passive during the 5 s count down with the right leg at 80º and 60º of hip and knee flexion, respectively, as determined by goniometry. At time 0, subjects began pedaling as quickly as possible to maximal pedal cadence and sustained maximal effort for the remainder of the 30 s test.

Post-acquisition data processing

Ergometric data was saved and exported as text files that were subsequently imported into a commercial spreadsheet program (Excel, Microsoft Corporation, Seattle, WA). For pedal force and EMG data, post-acquisition data processing was performed using custom developed programs (LabVIEW, National Instruments, Austin, TX). Raw pedal force data were used within the software to manually isolate each pedal stroke. The peak voltage and corresponding time of each contraction were obtained after the data segment spanning the peak was fitted with a second order polynomial function. This was repeated for all contraction peaks of each trial. Processed data were saved as text files and imported into a graphics program (Prism, GraphPad Software, San Diego, CA) for nonlinear curve fitting to decrease variability and allow calculation of time specific data values for use in statistical analyses and graphical presentation.

For EMG data, the root mean square (rms) of the EMG signals was used to manually isolate each muscle contraction. For each contraction segment, data were converted to rms and processed via Fast Fourier Transformation (FFT). Mean signal frequencies were calculated from the FFT spectrum data. Data for each contraction were then saved to a text file for importing into a commercial spreadsheet program (Excel, Microsoft Corporation, Seattle, WA) for nonlinear curve fitting (Prism, GraphPad Software, San Diego, CA) as explained for the pedal force data. The EMG rms data was converted to relative units by dividing by the peak contraction EMG rms for a specific trial and expressed as a percentage.

Due to constraints regarding manuscript length, number of figures, and our primary purpose of assessing the validity of the WAnT, the data analyzed and presented have been confined to the EMG data for the right vastus lateralis, force data from the right pedal, without biomechanical data of pedal or crank angles or use of the angles to compute coordinate vectors of pedal force.

Statistical analyses

Data were entered into a commercial spreadsheet program (Excel, Microsoft Corporation, Seattle, WA) for data screening, mathematical conversion for body weight and peak data adjusted expressions. Time data were obtained from processed data nonlinear equation derived values for 1, 3, 7, 10, 15, 22, and 30 s.

Mean differences in contraction pedal force application, relative EMG rms, EMG mean frequency, cadence, and power were analyzed by repeated measures 2-way ANOVA (TRIAL [2] × TIME [7]). Differences between trials for peak power, time to peak power, cadence at peak power, cadence decrement, power decrement and fatigue index were analyzed by paired t-tests. Pearson correlation coefficients were computed to explore potential relationships for select independent variables to the dependent variable time to peak power. All statistical analyses were completed using a commercial statistical package for the personal computer (SPSS 17.0.2, IBM Statistics, IBM Australia, St Leonards, NSW). Statistical significance was accepted at $p < 0.05$, and all grouped data are presented as mean ± SD.

Results

Subjects were of varied gender, training status and sports specialization (Table 1), ranging from involvement in recreational resistance training to highly trained and competitive athletes. Mechanical data from a representative subject (#20) for the TRAD and STAT trials are presented in Figure 1. Raw acquired data (volts) for subject #20's pedal force application are presented in Figure 2 for the TRAD and STAT trials, respectively. The inset figures within Figure 2b each present the second order polynomial fit to a segment of the pedal force volts data for two different contractions occurring early and late into the test. Figure 3 presents subject #20's peak values and nonlinear curve fitting for peak pedal forces as raw voltage output for the TRAD and STAT trials. The data and curve fitting shown in Figure 3 represent the methods used to obtain data points for the times used in data conversion to force (Newtons) from the calibration regression equations, statistical analyses and for the mean data presentations that follow (see Materials and Methods). A similar data presentation is shown in Figure 4 for EMG rms and EMG mean frequency for each of TRAD and STAT, respectively.

Table 1. Descriptive characteristics of the subjects

#	Age	Gender	Height (cm)	Weight (kg)	Training Status	Type of sport
1	2	3	4	5	6	7
1.	25	F	173.99	61.4	Endurance	Running
2.	21	M	189.23	111	Endurance	Track and field
3.	20	F	169.20	64.2	Endurance	Cross country skiing
4.	25	M	175.26	66.2	Resistance	
5.	21	F	160.02	53.8	Power/Resistance	Pole vault
6.	21	F	172.72	61.2	Power/Resistance	Pole vault
7.	20	F	175.60	71.8	Endurance	Running
8.	23	F	168.91	72.6	Power/Resistance	Track and field
9.	24	M	193.04	109.8	Power/Resistance	Football
10.	25	F	154.94	57.6	Endurance	Triathlon

1	2	3	4	5	6	7
11.	32	M	182.88	70.6	Endurance	Cycling
12.	19	M	177.80	69.4	Endurance	Running
13.	19	F	165.10	53.8	Power/Resistance	Track and field
14.	19	F	162.56	53.8	Power/Resistance	Track and field
15.	25	F	165.10	72.6	Power/Resistance	Hockey
16.	20	M	180.34	73.4	Endurance	Running
17.	24	F	167.64	58.4	Power/Resistance	Track and field
18.	22	M	182.88	69.8	Endurance	Running
19.	24	M	182.88	78.8	Power/Resistance	Strong man
20.	45	M	179.10	86.8	Endurance/Resistance	Swim, cycle, weight lifting

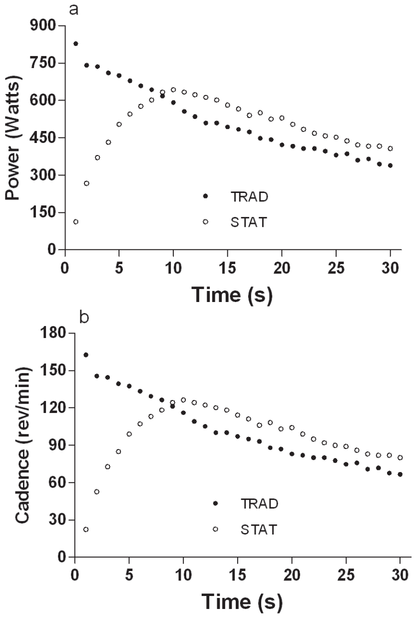

Figure 1. Mechanical power and cadence for the TRAD and STAT trials for Subject #20

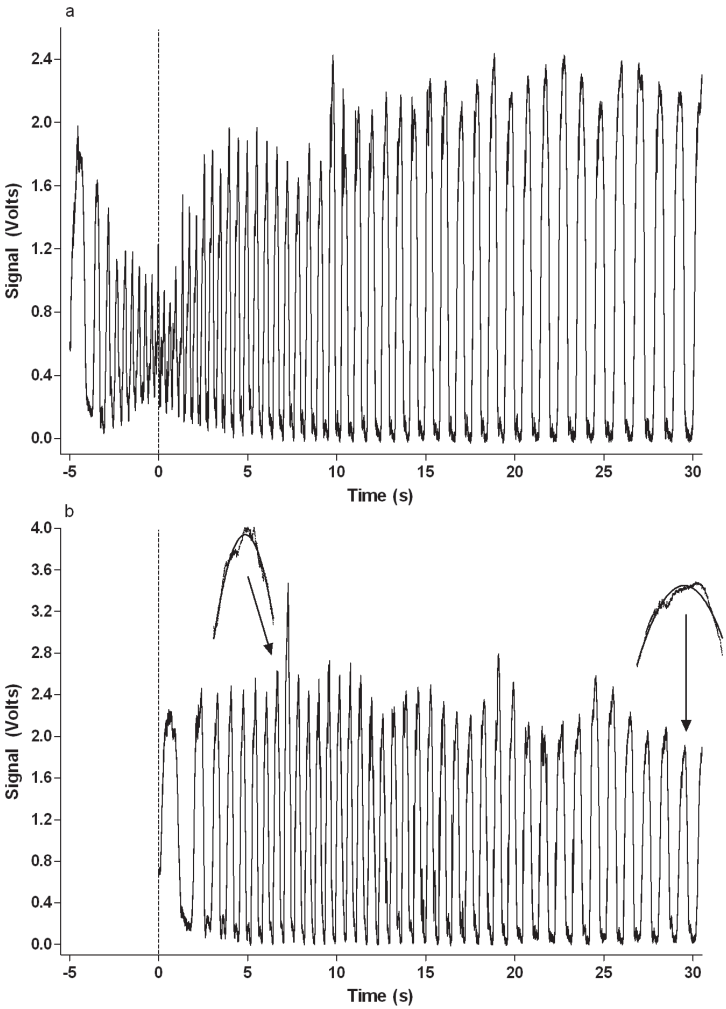

Figure 2. Acquired raw data (volts) from Subject #20 for pedal force application for the a) TRAD and b) STAT trials, respectively. Inset figures of Figure 2b show the 2nd order polynomial functions fit to peak curve segments of contractions 8 and 38 to compute peak force and time of peak force

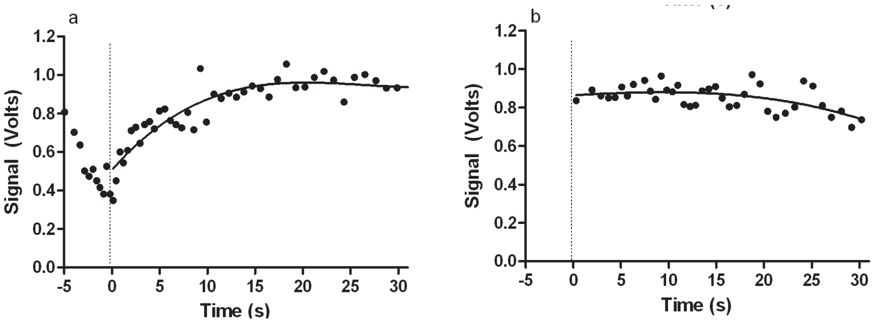

Figure 3. Raw peak pedal force data (Volts) and nonlinear curve fitting for Subject 20 for the a) TRAD and b) STAT trials

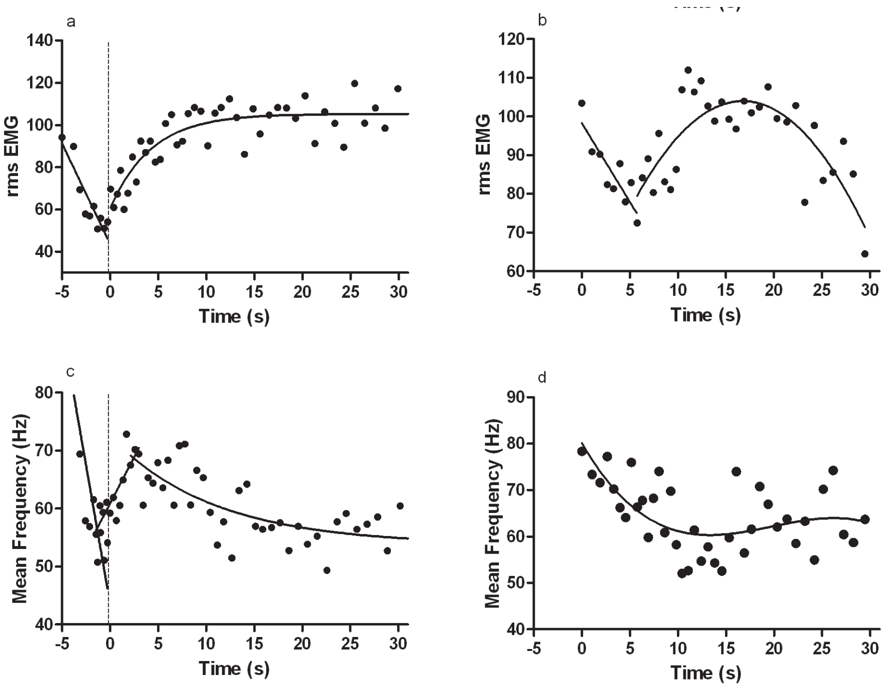

Figure 4. Contraction data and nonlinear curve fitting for Subject 20 for a) EMG rms TRAD, b) EMG rms STAT, c) EMG mean frequency TRAD, d) EMG mean frequency STAT

Grouped data for the mechanical power and cadence for TRAD and STAT are presented in Figure 5. As expected for power, there was a significant TRIAL x TIME interaction ($p < 0.001$), where peak power for TRAD was near instantaneous whereas it did not occur until 8.3 s for STAT. Actual time to peak power data for TRAD vs. STAT were 2.20 ±1.73 vs. 8.30 ±2.56 s, respectively ($p < 0.001$). Peak power (PP) was significantly higher in TRAD than STAT at 812 ±247 vs. 734 ±193 Watts ($p = 0.002$), and 11.32 ±1.41 vs. 10.40 ±1.35 Watts/kg

($p = 0.001$), respectively. Peak cadence was significantly higher in TRAD than STAT at 171 ±16 vs. 121 ±15 rev/min, respectively ($p < 0.001$). The fatigue index was significantly greater for TRAD than STAT at 50.8 ±9.4 vs. 37.6 ±11.0%, respectively ($p < 0.001$).

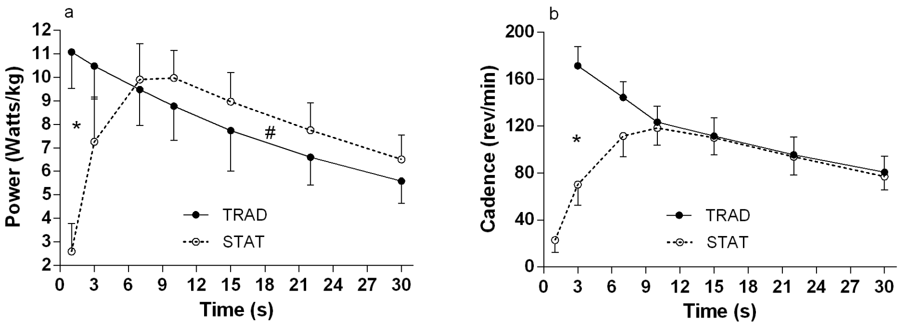

Figure 5. Results from the ANOVA analyses for a) mechanical power and b) cadence for the two trials. * = significant interaction and mean differences between TRAD vs. STAT at 1 and 3 s for Power, and 1, 3 and 7 s for Cadence. # = significant differences for Power between TRAD and STAT for 10, 15, 22 and 30 s

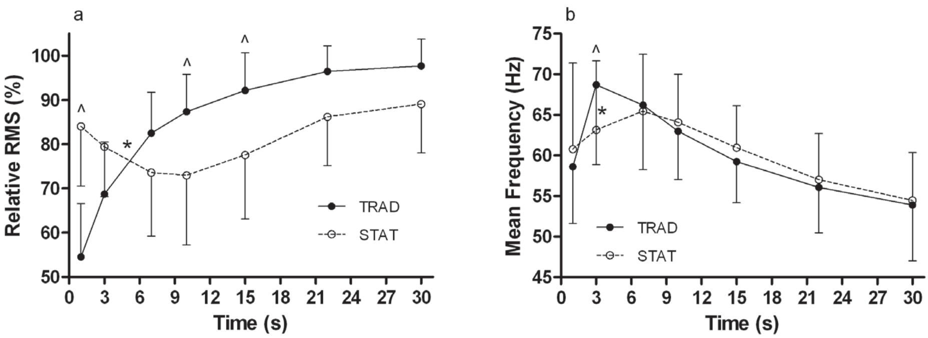

Figure 6. Results from the ANOVA analyses for vastus lateralis a) relative rms EMG activity and b) mean frequency for the two trials. * = significant TRIAL x TIME interaction. ^ = significant differences between TRAD and STAT for that time point

The trial differences in peak power and time to peak power occurred with significant differences between trials for both vastus lateralis relative EMG rms and EMG mean frequency (Figure 6). For EMG rms there was a significant TRIAL x TIME interaction ($p < 0.001$), where relative EMG rms for TRAD was low, and increased exponentially to a near plateau by 30 s. Conversely, the data for STAT commenced at a higher value ($p < 0.05$) and changed variably through the test. Data were very consistent between subjects for TRAD, but differences existed in the timing of the peak value during STAT, and consequently, mean data for relative EMG rms for STAT did not approach 100% at any time of the test. After 10 s, the trend for increasing relative EMG rms was similar between trials. For EMG mean frequency there was also a significant TRIAL x TIME interaction ($p = 0.023$), which was also confined to the initial 10 s of the test and caused by comparatively lower signals for the TRAD after test onset.

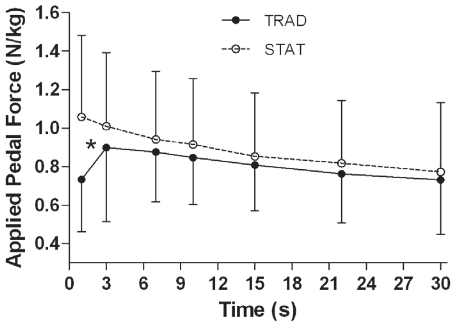

Figure 7. Results from the ANOVA analyses for applied pedal force corrected to Newtons (N) and expressed relative to body mass (N/kg). * = significant interaction and mean differences between TRAD vs. STAT at 1 and 3 s

As expected from the EMG data, relative force applied to the pedals (Figure 7a, b) also attained a significant TRIAL x TIME interaction ($p < 0.001$) which was confined to the initial 10 s of the test. The initial relative pedal force at 1 s was significantly lower ($p < 0.001$) in TRAD vs. STAT, and after 7 s was no longer different between trials. Results from the 2-way repeated measures ANOVAs and t-tests for all remaining variables are presented in Table 2.

Table 2. Results for data and analyses not presented in figures

Variables	Mean±SD	t : df	p
TRAD PP, Watts	812.0 ±247.1		
STAT PP, Watts	734.2 ±193.1	3.60 : 19	0.0020
TRAD PPkg, Watts/kg	1.32 ±1.41		
STAT PPkg, Watts/kg	10.40 ±1.35	4.09 : 19	0.0010
TRAD TtoPP, s	2.20 ±1.74		
STAT TtoPP, s	8.30 ±2.56	–8.50 : 19	0.0001
TRAD PCad, rev/min	171.4 ±16.3		
STAT PCad, rev/min	120.9 ±15.1	13.79 : 19	0.0001
TRAD FI, %	50.8 ±9.4		
STAT FI, %	37.6 ±11.0	6.29 : 19	0.0001

PP = peak power; PPkg = PP/kg body weight; TtoPP = time to PP; PCad = peak cadence; FI = fatigue index.

Table 3. Correlation and significance (italics) matrix for select independent variables of the STAT trial

Variables	TtoPP	PP	PP/kg	PCad	FI
TtoPP	1.0	*0.056*	*0.026*	*0.036*	*0.022*
PP	–0.367	1.000	*0.014*	*0.012*	*0.09*
PP/kg	–0.441	0.488	1.000	*0.000*	*0.035*
PCad	–0.441	0.503	0.996	1.000	*0.047*
FI	–0.454	0.312	0.413	0.385	1.000

PP = peak power; PPkg = PP/kg body weight; TtoPP = time to PP; PCad = peak cadence; FI = fatigue index.

Simple linear regression revealed that time to peak power was significantly correlated to relative peak power, peak cadence and the fatigue index (Table 3).

Discussion

In summary, we showed that the decrease in mechanical power during the first 10 s of the TRAD coincides with an increasing EMG rms and increasing force application to the pedals. These results reveal the invalidity of the initial 10 s of power calculations from the TRAD, as well as the poor construct validity for interpreting the initial power decrement as evidence of muscular fatigue.

Dotan (2006) recently compiled a list of criticisms of the TRAD. However, the issue of the invalid procedures at the start of the WAnT was not raised. Similarly, no concerns regarding the start of the test have been raised in prior research using a stationary start modification (Jastrzębski 1995; MacIntosh et al. 2003). Thus, we are the first to present data to (a) show the invalidity of TRAD, and (b) measure variables that may provide a plausible explanation for the invalidity. The rapid decrease in power at the start of the TRAD coincides with a rapid decrease in cadence, increasing EMG rms and increasing application of force to the pedals. The high initial cadence at the start to the TRAD constrains muscle force development due to the inverse relationship between muscle contractile velocity and force. To generate muscle contractile force, cadence needs to be reduced to allow sufficient time for increased motor unit recruitment and force production.

As we show, and has previously been shown but overlooked in all prior research of the WAnT, there is a divergence in the slope of the cadence decrement that occurs at approximately 10 s, though in some subjects this is difficult to detect due to the nonlinear profile of cadence over time. This divergence is more easily detected in the power data (Figure 1a). Interestingly, after approximately 10 s, cadence has decreased to values known to be more closely aligned to peak muscular power development (100-120 rev/min) (Dimitrova and Dimitrov 2013; Hautier et al. 1996; Katch et al. 1977; MacIntosh et al. 2000; MacIntosh et al 2003; Reiser et al. 2003; Samozino et al. 2007). As such, one could argue that all prior peak power measures from the TRAD are artificially high and caused by the excessively high cadence at the test onset. Indeed, Santos et al. (2010) has shown that peak power increases with more rapid data acquisition from 0.2 to 5 Hz. Our results, based on rapid signal acquisition, support this close alignment of peak cadence with peak power immediately prior to onset of TRAD. Since load is constant during the WAnT, a higher cadence means higher mechanical power. This fact alone reveals the dependence of peak power and power decrement during TRAD to the methodology of the WAnT. It is physiologically impossible for humans to generate peak muscular power at such high cadences.

Using a modified stationary start to the WAnT, Jastrzębski (1995) demonstrated a power curve that does not include the initial steep decline in power seen in the standard test but rather a steady increase in power from the start of the test to peak cadence that occurred in 6 s, which was similar to our time to peak power of 8.3 s (Figure 5a). The use of a stationary start resulted in decreased peak power (W/kg) and work output (kJ) suggesting results obtained from a standard test may over estimate peak power and fatigue index and thus the potential for intense exercise performance. MacIntosh et al (2003) compared multiple methods of load determination between a traditional vs. stationary start to the WAnT. It is difficult to compare our results to those of MacIntosh, as they used an atypical and custom modified Monark ergometer (presumably not the model made for the WAnT), did not provide sufficient methodological details of load application to the ergometer, and obtained results indicative of a delayed load application consistent with the use of either a manual load application or a non-instantaneous electronically braked ergometer system. For example, subjects required 5.63 ±1.83 s (mean ±SE) to attain peak power in the traditional version of the WAnT, rather than peak power being almost instantaneous from test onset as shown by our data (Figures 1 and 5) and that of Santos et al (2010). Similarly, MacIntosh (2003) reported similar peak power values between the traditional vs. stationary start versions of the test. This is a peculiar finding given the constant load and the significantly higher peak cadence that should occur during the traditional version of the test compared to stationary start.

Our EMG results are particularly compelling in revealing the lack of construct validity of the TRAD. We performed a unique post-acquisition contraction envelop capture of EMG signals to allow a contraction-by-contraction method of EMG processing. This provided high temporal resolution of EMG signals, which in turn allowed us to model the EMG rms response during the 30 s tests and compare EMG rms signals at specific times. While we acknowledge the controversy inherent in interpreting surface EMG data during intense exercise (Dimitrova and Dimitrov 2013; Farina 2006; Hug and Dorel 2009; Hug et al. 2008; Hunter et al. 2003; Inbar et al. 1996; Marsh and Martin 1995), the dramatic differences between TRAD and STAT for EMG rms and EMG mean frequency during the initial seconds of the tests (Figures 4, 6) reveal likely differences in motor unit recruitment patterns. This is especially compelling given that the electrodes and their placements were identical between trials, that all testing was completed on the same day. EMG rms signals were normalized to the peak contraction of each trial (Burden 2010, Hunter et al. 2003), and trial order was balanced between subjects. Clearly, TRAD did not coincide with large EMG rms signals during the high cadence at and immediately after test onset (Figure 6a). Interestingly, this period of TRAD also coincided with a low EMG mean frequency, which increased as cadence decreased to near optimal rates after 3 s (Figure 6b). The excessively high cadence at the start of TRAD compromised motor unit recruitment; this finding complements the applied pedal force data.

The lack of commercial pedal force systems necessitated that we develop our own, and we were motivated to accomplish this based on the work of Dorel et al. (1983) and Reiser et al. (2003). Both research groups used multiple strain gauges positioned along the central pedal support bar. Due to the complexity and expense of this approach, we researched more economical and electronically simple options for applied force measurement. We chose the option of multiple piezo-resistive force sensors. Pilot work with these sensors combined with an inverting summing circuit for multiple sensors produced quality signals that were temporally sensitive to resolve force application throughout 360° at cycling cadences up to 180 rev/min, and produced signals proportional to applied force over more than 100 kg of load range. The quality of these signals is shown in Figure 2. To our knowledge, we are the first to develop such pedal force sensors and apply pedal force measurement during short term intense cycle ergometry.

Our results reveal the importance of pedal force measurement to establish and test the validity of mechanical power measurement during cycling, and to further research and interpret the concept of muscular fatigue. For example, the combined pedal force and EMG data clearly show that there is no muscular fatigue during the initial ~10 s period of the TRAD.

With regard to fatigue, data from Figures 2, 6 and 7 reveal the potential error of using force application as a criterion to define muscular fatigue. For example, the mean data for pedal force application (Figure 7) show an immediate gradual decline throughout STAT when both mechanical power development and cadence increase (Figure 5). The results during the initial 10 s of the test occur concomitantly with a decreasing EMG rms, and increasing EMG mean frequency. We interpret this data to reveal how muscle motor unit recruitment is challenged during the initial 10 s of STAT. During STAT, after approximately 8–10 s, cadence has increased to a more optimal value for peak power development, and muscle contractile force remains amendable to peak power development. During TRAD, a similar time period is needed for cadence to decrease to values conducive to optimal muscular power. After approximately 10 s, in both TRAD and STAT, there is a minimal decrease in applied force, yet larger and sustained decreases in EMG mean frequency, cadence and mechanical power, with increases in EMG rms. These responses revealed that the most impressive data was actually the simplest measure; cadence. Cadence is a simple indirect reflection of contractile velocity, and reveals that despite being able to maintain contractile force, fatigue is more aligned to a decreasing capacity for the rapidity of force development (contractility). We do not present any data on contractility in this manuscript, but we have the data to compute muscle contractility from the raw data of the initial change in pedal force application for each crank cycle as shown in Figure 2. Nevertheless, decreasing cadence results from decreasing contractility, which again is why cadence is an important measure with considerable physiological meaning.

The extremely high correlation between peak cadence and relative peak power (Table 2) reveals the importance of cadence and neuromuscular attributes (mass and fiber types) to mechanical power output. In addition, the significant correlations between time to peak power and all variables dependent on peak power in Table 3 demonstrate the physiological importance of the time to peak power variable. In fact, we propose that time to peak power and peak cadence are equally important variables to peak power during intense cycle ergometry exercise testing, and future application of these variables to actual intense exercise cycling performance is an important topic of future research.

Conclusion

We present data that verifies the invalidity of interpreting the mechanical power measurements derived from the TRAD during the first 10 s to be reflective of muscular power development. The initial 10 s of power computations from the TRAD are induced by the excessively high cadence at test onset against a constant load. The initial 10 s of the test necessitate that cadence decrease to allow time for more optimal motor unit recruitment to generate the force required for peak muscular power development. Such peak power coincides with a cadence of 100 to 120 rev/min, depending on the subject. Thus, the mechanical power at 10s from the WAnT has far more physiological meaning than does peak power.

The TRAD should not be used to quantify peak power or power decrement. Performing short term intense cycle ergometry testing using a stationary start has far greater physiological meaning. The added variables of time to peak power and cadence at peak power that can be measured from a stationary start method may provide further insight into subject specific physiological capacities suited to intense exercise performance, and should be the focus of future research.

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