Invited Paper

Physiological, Biochemical and Mechanical Issues Relating to Resistive Force Selection During High-intensity Cycle Ergometer Exercise

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High-intensity cycle ergometry of 30 seconds duration has been widely employed to assess indices of muscle performance during maximal exercise. Traditionally, the resistive force established for such a test is determined from total body mass (TBM) for a friction-loaded Monark cycle ergometer, i.e. $75\,g \cdot kg^{-1}$. More recent studies have shown that traditional forces may be too light to elicit maximal performances and that optimization protocols can produce higher peak power outputs. Conceptually, selecting the optimal resistive force according to TBM may not be the best approach. Fat-free mass or active muscle tissue may be a more preferable alternative. Because body mass, and not composition, is the most commonly used index to determine cycle ergometer resistive force, over- or underestimations in power calculations may occur. The aim of this paper is to outline friction-loaded cycle ergometer performance using resistive forces derived from TBM and fat-free mass, to quantify the upper body contribution to high-intensity cycle ergometry. A further aim is to outline mechanical issues related to cycle ergometer design and to quantify discrepancies in resistive force application. [*J Exerc Sci Fit* • Vol 7 • No 2 (Suppl) • S51–S60 • 2009]

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Introduction

Tests of high-intensity power and capacity have been extensively used by exercise physiologists to help characterize athletic groups and to investigate the high-intensity potential of healthy and special populations. To date, there is no specific test that can be considered a valid indicator of both power and capacity, because different test protocols measure different components of high-intensity performance (Smith 1987). Measurements of these different characteristics can be achieved either by computing the amount of mechanical work that can be performed in a specified time, or by monitoring the time taken to perform a given amount of high-intensity work (Winter et al. 1991). The evaluation of high-intensity power and capacity may also depend on the interpretation of experimental data. Details of units of measurement and data evaluation need to be examined closely prior to experimental data collection (Vandewalle et al. 1987). Evidence would suggest that the amount of work performed during an intense maximal test depends on glycolytic power, glycolytic capacity and

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aerobic ability. The relative contribution from each system seems to be related to the intensity of the exercise and the duration of the task. High-intensity performance has been assessed in the main by cycling on stationary friction-loaded cycle ergometers and recording the power profiles obtained. Cumming (1973) introduced a friction-braked cycle ergometer test that was further developed at the Wingate Institute in Israel and became known as the Wingate Anaerobic Test (WANT). The prototype was presented by Aylon et al. (1974), and since its introduction, a comprehensive description has been published (Bar-Or 1981).

Resistive Force Selection

In test protocols using cycle ergometry where a single exercise bout is performed, it is important to set a resistive force that matches the capability of the muscle to contract. In this way, true maximal power output can be measured at, or close to, optimal velocity.

A number of authors have addressed the possibility of predicting the optimal resistive force from body mass. This issue however has not been fully resolved (Bar-Or 1987; Aylon et al. 1974). Drop loaded, cradle or friction-loaded ergometers have permitted rapid applications of load and quantification of the subsequent values for power produced. In the original studies of Aylon et al. (1974) using Monark ergometers, the loads were in the order of $75\,g \cdot kg^{-1}$ total body mass (TBM). Dotan and Bar-Or (1983) declared that a higher optimal value, namely $87\,g \cdot kg^{-1}$ TBM, produced greater power outputs. Several other researchers have indicated that these load ratios may still be too small, especially for athletes involved in sprint or power-based activities (Winter et al. 1991; Nakamura et al. 1985). Optimal values for resistive forces used during high-intensity cycle ergometry testing have been based on TBM indices. These indices include both active muscle tissue and fat mass. Resistive forces used, which are currently inclusive of the fat component of body composition, may not be representative of the lean tissue mass or muscle mass utilized during maximal cycle ergometer performance. Power measurements during cycle ergometry also include an unknown upper body contribution that contributes to the power profiles obtained (Baker et al. 2002, see Table 1; Baker et al. 2001c, see Figure 1).

Body size, structure and composition differ markedly among individuals, suggesting that a standard ergometer load may not provide optimal resistances for different populations, and may be individual-specific. This suggests that the assessment of physique should be considered in any evaluation of high-intensity performance.

Fat-free Mass *vs.* TBM

It would seem appropriate to exclude fat mass from any resistive force protocol that attempts to establish a relationship between power production and the capacity of active muscle. Van Mil et al. (1996) have reported performance in high-intensity experimental procedures as being highly related to the subjects' lean body mass, or the mass of the muscles that perform the test. The direct method of determining the resistive force for individual subjects during high-intensity cycle ergometry is to provide the subjects with a test protocol that requires them to perform the test repeatedly, each time against a different breaking force until a maximal value for power is obtained (Dotan & Bar-Or 1983; Evans & Quinney 1981). An alternative semi-direct approach has been to assign a braking force that is based on individual subjects' TBM and a performance ratio (normally $75\,g \cdot kg^{-1}$ TBM; Aylon et al. 1974). The assumption has been that for most healthy individuals, the relationship between TBM and muscle mass is similar. This is clearly not the case, and the relationship may be compromised further in populations that include the athletic, the undernourished and the obese. This would result in power estimation error

Table 1. Blood lactate levels for both protocols analyzed from capillary blood samples taken at three time periods*

Pre-exercise		Immediately post-exercise		4 min post-exercise	
WG ($mmol \cdot L^{-1}$)	WOHG ($mmol \cdot L^{-1}$)	WG† ($mmol \cdot L^{-1}$)	WOHG ($mmol \cdot L^{-1}$)	WG† ($mmol \cdot L^{-1}$)	WOHG ($mmol \cdot L^{-1}$)
0.98 ± 0.99	0.99 ± 0.79	5.68 ± 1.37	5.58 ± 1.74	9.14 ± 1.41	7.62 ± 1.94

*Data are presented as mean ± standard deviation; †significant differences ($p < 0.05$) were recorded pre-exercise to post-exercise and 4 minutes post-exercise for both the with handgrip (WG) and without handgrip (WOHG) protocols. No differences ($p > 0.05$) were recorded at the immediate post-exercise stage between groups, but differences were observed between immediately and 4 minutes post-exercise for the WG protocol only ($p < 0.05$).

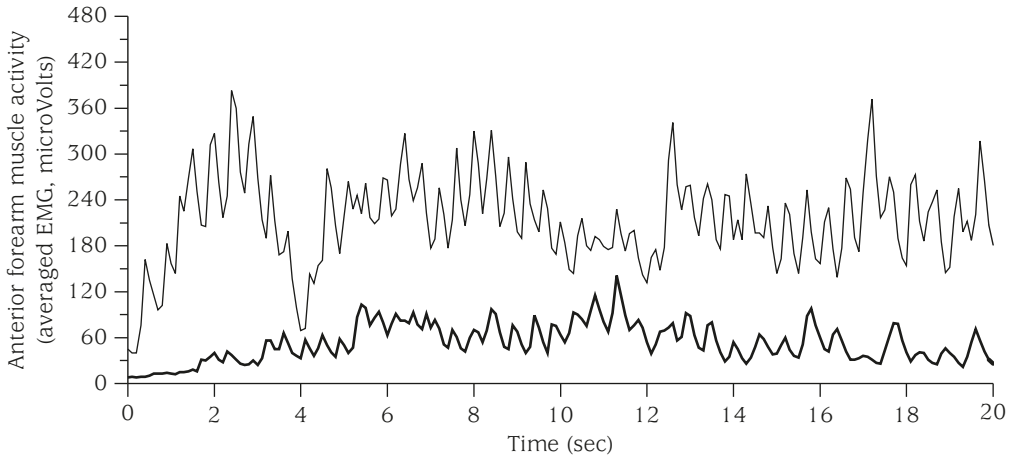

Fig. 1 Schematic diagram of left anterior forearm muscle activity (electromyography) of the two test protocols. The with-grip protocol consisted of the subject placing their hands on the handlebars of the cycle ergometer in a traditional gripping fashion (thin solid line). The without-grip protocol (thick solid line) consisted of the subject placing the posterior aspect of each wrist on the handlebars so that the open palms faced superiorly. Contact with the handlebar was maintained at the most distal points of the radial and ulnar styloid processes. The figure clearly demonstrates an increase in muscle activity when the with-grip protocol is used.

during high-intensity exercise performance tasks. The differences observed may reflect the inconsistent muscle mass to TBM ratio in individuals. The protocol for friction-loaded high-intensity cycle ergometry exercise has undergone many modifications and refinements since its introduction in 1974. The use of a higher force in order to maximize power output is a major challenge and is highly recommended (Bar-Or 1987).

To date, during force/velocity relationship assessment, the loads used have been based on TBM values and have ranged from $75\,g \cdot kg^{-1}$ to $130\,g \cdot kg^{-1}$ (Inbar et al. 1996). The resistive forces have also been based on specific guidelines for different populations and sexes (British Association of Sport and Exercise Sciences 1988) or have been derived individually using various optimization procedures. Several investigators are of the opinion that the fat-free mass (FFM) method of resistive force selection appears to be more representative of active muscle tissue activity (Baker et al. 2000; Inbar et al. 1996; Van Mil et al. 1996). Figures 2 and 3 clearly demonstrate that more of the variance in performance is accounted for when resistive force selection reflects FFM as opposed to TBM.

Optimization for FFM appears to provide more accurate and meaningful direct comparisons within and between sport-specific and non-athletic groups. When applying the FFM method of resistive force selection in conjunction with a force velocity protocol, the results obtained seem to provide not only a realistic method for determining optimal resistances, but also accurate and reliable power profiles. Tharp et al.

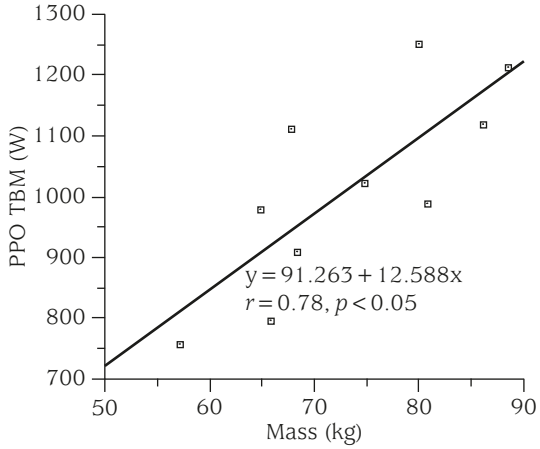

Fig. 2 Relationship between total body mass (TBM) resistive force selection and peak power output (PPO) values recorded during high-intensity cycle ergometry.

(1984) suggested that the values generated during high-intensity cycle ergometry exercise are highly correlated to body mass. They also suggest that although a heavier person should produce a higher cycle ergometer score, the values obtained when expressed relative to FFM produced a better index of high-intensity performance when comparisons between subjects were made. However, heavier resistive forces based on TBM computations may produce greater errors in power calculations that are related to frictional forces transmitted to the ergometer flywheel and may compromise relationships with other measures of high-intensity effort (Baker & Davis 2002; see Figure 4).

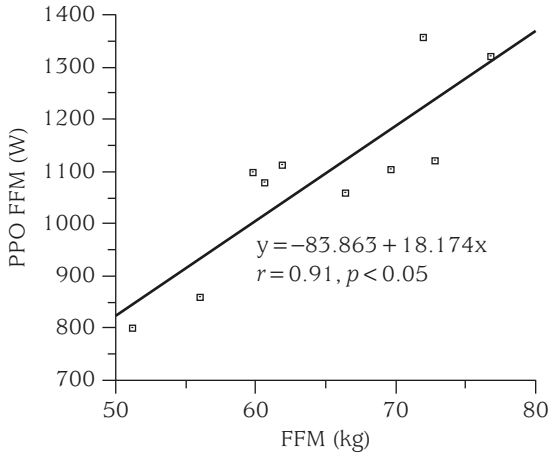

Fig. 3 Relationship between fat-free mass (FFM) resistive force selection and peak power output (PPO) values recorded during high-intensity cycle ergometry.

	30 m	40 m	10 m	VJ	HJ	PPO
30 m		0.83*	0.80*	0.91*	0.87*	0.51
40 m			0.90*	0.81*	0.80*	0.51
10 m				0.80*	0.80*	0.51
VJ					0.86*	0.51
HJ						0.51
PPO						

Fig. 4 Correlation matrix for field measures of high-intensity exercise and peak power outputs (PPOs) obtained during high-intensity cycle ergometry when resistive forces were optimized. From the correlations obtained, it can be seen that there are no significant relationships between the field measures and the high-intensity cycle ergometer values. The field tests, however, are all interrelated. *$p < 0.01$. 30 m = 30-m sprint; 40 m = 40-m sprint; 10 m = 10-m sprint; VJ = vertical jump; HJ = horizontal jump; PPO = peak power output.

McInnis and Balady (1999) have stated that because TBM consists of fat and FFM, individuals who weigh the same may have very different body compositions. Differences observed may also reflect specificity of training status between subjects. The FFM protocol appears to identify more subtle changes in resistive force profiles, which may have resulted from smaller relative load increments during an optimization procedure.

The smaller load increases appear to accommodate the sensitive changes in power outputs during a force velocity test that the TBM protocol disregards.

The higher peak power outputs (PPOs) observed for FFM indicate that this method of resistive force selection does not overestimate the capacity of the active muscle mass, and therefore maximizes both resistive force and pedal revolutions. When using the TBM method of resistive force selection, the increases in braking force are greater for any given loading stage; as a result, the increased pedal velocity contribution to power production may be overlooked. The relative strengths of the correlations recorded between power outputs and resistive forces generated for the two protocols (greater for FFM), and the significant differences between loading procedures for TBM and FFM (Baker et al. 2000; Figures 2 and 3) suggest that the FFM optimization procedure is related more closely to the active tissue utilized during short-term high-intensity exercise.

Morphological and Metabolic Factors

For all force velocity relationships in humans, morphological factors contribute to force and power measurements, and may bias or improve power profiles (Bosco & Komi 1979). Morphological factors that relate to differences in size and structure of lever arms include length and pennation angle of muscle fibers. Force velocity relationships are also interrelated to factors that modify longer duration performances such as the efficiency of oxygen utilization, muscular blood flow and perceived exertion (Pugh 1974).

Power, the composite product of two factors (force and speed) can incorporate an infinite number of values. Therefore, a range of results is possible with varying contributions from both factors, especially when the criterion is optimization of absolute maximal power (Inbar et al. 1996). Baker et al. (2001a) have substantiated this suggestion. A greater power was achieved during a TBM and FFM protocol by increasing both the applied forces and increasing the number of pedal revolutions. With the increasing load, recruitment of more motor units with more muscle fibers per motor unit is most important until the load becomes too heavy (Åstrand & Rodahl 1986). Maximal muscular tension can be produced when the muscle is lengthened, and it declines during the concentric phase of muscle contraction. Within the range of force velocity interrelationships, those associated with maximized short-term power would be expected to most closely approximate the maximum single contraction as defined by the force velocity curve of Hill (1938).

Deviations from this relationship are mostly due to fatigue and the necessary muscular coordination associated with repetitive high-frequency motion. The intersubject differences observed between the TBM and FFM protocols may be related to individual inability to generate high levels of velocity. There may be many reasons for this, including the proportion of fast twitch fibers (type II) in the exercising muscle, and differences in physiological and biochemical factors that relate to both genetics and training status.

Type II fibers are known to have faster contraction times and rates of tension development than slow twitch (type I) fibers, and are more dependent on glycolysis to maintain ATP rather than the slower process of oxidative phosphorylation (McCartney et al. 1983). Thorstensson et al. (1975) have confirmed a greater proportion of type II fibers in athletes engaged in activities requiring short-lived or sprint-type power development. In the classical experiments describing the effects of contraction time on the work and efficiency of the elbow flexors (Hill 1922) and quadriceps group during cycling (Dickinson 1929), it was demonstrated that brief maximal and submaximal contractions were associated with an increased waste of potential energy.

In a system performing mechanical work where heat is liberated and free energy wasted, relatively more free energy must be supplied to maintain performance (Wilkie 1960). Baker et al. (2001a; see Table 2) suggested that during both a TBM and FFM protocol, the PPO values obtained were recorded with energy supplied almost exclusively from the degradation of phosphocreatine and glycolysis.

Table 2. Increases in peak power output and pedal revolutions with decreases in time to reach peak power output with corresponding decreases in resistive force when resistive forces reflect fat-free mass as opposed to total body mass*

Variable	TBM	FFM	*p*
R/Force (kg)	7.6 ± 1.4	6.7 ± 1.1	< 0.05
PR (rpm)	129.4 ± 8.2	136.3 ± 8.0	< 0.05
PPO (W)	1015 ± 165	1099 ± 172	< 0.05
MPO (W)	751 ± 109	769 ± 130.2	NS
FI (%)	27.8 ± 6.1	28.8 ± 8.4	NS
WD (J)	14,985 ± 2190	15,301 ± 2454	NS
T to PPO (s)	3.8 ± 1.4	2.9 ± 1.0	< 0.05
RPE	18.4 ± 1.6	19.8 ± 0.4	< 0.05
HR_{pre} (bpm)	78.4 ± 13.1	74.3 ± 16.5	NS
HR_{post} (bpm)	173.5 ± 9.1	172.3 ± 13	NS

*Data are presented as mean ± standard deviation. TBM = total body mass; FFM = fat-free mass; PR = pedal revolutions; PPO = peak power output; MPO = mean power output; FI = fatigue index; WD = work done; T to PPO = time taken to reach PPO; RPE = rating of perceived exertion; HR_{pre} = heart rate pre-exercise; HR_{post} = heart rate post-exercise.

Wilkie (1968) demonstrated that in muscle, the breakdown of phosphocreatine and glycogen over a cycle of relaxation and contraction is directly proportional to the sum of the heat and work produced. Moreover, during contraction, heat production is at a maximum under conditions in which the work is maximal (Fenn & Marsh 1935). Baker et al. (2001a) indicated that during the initial stages of performance, the work production was greatest when the subjects were optimized for FFM. This suggests a greater or more efficient utilization of muscle phosphagens when FFM is compared to TBM. In most cases, the time to PPO increased when the subjects were optimized for FFM, indicating a possible alteration in energy system contribution, with glycolysis being used to a lesser extent in the early stages of the FFM protocol. This may also indicate an increased degradation of phosphocreatine and glycogen, and greater changes in metabolic substrates. These factors could have exerted inhibiting effects on the biochemical processes associated with muscle contraction, and may contribute to fatigue. Increased H^+ in muscle may decrease force generation by impairing Ca^{2+} release from the sarcoplasmic reticulum (Nakamura & Schwartz 1972), or by disturbing cross-bridge formation.

High levels of blood acidity and lactate accumulation are also observed following maximal exercise (Harris et al. 1977). At high rates of contraction, there is less time for the dispersion of metabolites from muscle, and the intramuscular accumulation of waste products may proceed at an accelerated rate (Grimby & Saltin 1977). Results using animal studies have demonstrated that individual fast twitch motor units, and whole muscles with a high percentage of type II fibers, are capable of higher levels of tetanic tension and are more susceptible to fatigue than type I fibers (Vandewalle et al. 1987). Studies on intact human muscles have reported that individuals with muscles containing a high proportion of type II fibers are capable of faster contraction velocities, and therefore greater force output (Thorstensson et al. 1975), but are more prone to fatigue during repeated dynamic contraction. Nilsson et al. (1977) demonstrated a strong correlation ($p < 0.05$) between an increase in the ratio of electromyographic activity to power associated with fatigue, with a high percentage of type II fibers, suggesting that diminished force was due to a selective drop out of this type of fiber. Di Prampero (1981) has suggested that a reduction in contractile speed rather than the depletion of high-energy phosphates may be a

major cause of fatigue during activities requiring maximal power output. The circular motion of the pedals further complicates maximizing power output during short-duration cycle ergometry. The circular motion affects the nature of force application, which is influenced by the degree of skill and coordination required for a given motion sequence frequency (Soden & Adeyefa 1979). It has also been demonstrated that the internal work associated with the acceleration and deceleration of the leg mass increases with the square of the increased pedaling rate (Kaneko & Yamazaki 1978). Therefore, the energy loss at 80 rpm already amounts to 5% of the external power output and would exceed 20% at 120 rpm. The increase in power output observed when the subjects were optimized for FFM may be the result of increased voluntary command of the supraspinal centers.

This greater contribution may increase fiber recruitment, by the optimization of individual motor unit firing frequency, and by the synchronization of the firing patterns between the motor units themselves (MacDougall et al. 1991). This increase depends on the muscles' ability to translate high-frequency impulse excitation through the various excitation processes with minimal time delay. In addition, the muscle needs to associate and dissociate the actin and myosin as they repeatedly rotate through successive cross-bridge cycles. It is possible that an increase in neural stimulation will enhance recruitment frequency of the muscle spindles, which would result in a corresponding increase in muscular contraction. The results recorded for the FFM protocol indicate that existing optimization protocols should be reviewed if increased power output is desirable. Increased PPO values resulting from higher pedaling rates during optimization procedures for FFM may maximize muscle contraction dynamics. These findings are in contrast with those of previous authors (Patton et al. 1985; Katch 1974). However, other researchers (Baker et al. 2004, 2003, 2001a, 2000; Dore et al. 2001; Inbar et al. 1996; Van Mil et al. 1996; Blimkie et al. 1988) have found that during high-intensity cycle ergometry, the power profiles generated are related to the subjects' FFM or to the mass of the muscles that perform the test.

Special Populations

Although originally used with able-bodied healthy subjects, high-intensity cycle ergometry can be used in conjunction with specific populations to assess subjects with chronic disease or physical disability. The rationale for such an application has been that the factors limiting physical performance may be muscular or neurological in nature rather than cardiorespiratory (Inbar et al. 1996). Therefore, testing their peripheral function may have diagnostic and prognostic value. However, important questions remain about the feasibility and reliability of high-intensity cycle ergometry when people with a physical disability perform it. Problems of standardization arise for such subjects because of the marked variation in ability, fitness levels and active muscle mass that may be independent of resistive force selection (Inbar et al. 1996).

For example, many people with cerebal palsy (athetosis or spasticity) cannot keep their feet on the pedals during the performance of high-intensity cycle ergometry even when stirrups are used. However, these problems have been overcome and meaningful results obtained when the subjects had their feet taped to the pedals (Parker et al. 1992). Further difficulties were encountered in patients with extreme muscle weakness. These subjects on occasion find it impossible to complete a full pedal revolution. A mechanical solution to the problem was found by decreasing pedal crank length, thus facilitating the rotation of the flywheel at a smaller pedal circumference. Although these problems are to a certain extent mechanistic/technical, selection of resistive forces that relate to active muscle tissue in these populations may be desirable. The greater mechanical resistance to motion inherent using resistive forces derived from TBM as opposed to FFM may further compromise and confound the mechanistic problems outlined.

For most healthy non-athletes, the assumption has been that the relationship between muscle mass and TBM is similar. However, in certain segments of the population, i.e. those subjects who are obese, undernourished, have muscle atrophy, muscle hypertrophy or neuromuscular disease, this relationship deviates from the norm. In these groups, the FFM is smaller or greater than expected in relation to TBM. For these populations, the assignment of a resistive force based on TBM may not only yield an overestimation/underestimation of maximal anaerobic performance, but may further compromise the health status of the patients themselves (Van Mil et al. 1996). The FFM protocol may also be an attractive alternative for the assessment of high-intensity potential in the elderly population. This subject group may possess different lean tissue mass to fat mass ratios for reasons that may be medical or non-medical. The differences

observed may be related to issues that are to a certain extent independent of health status, such as social standing, depravation and emaciation. Optimal high-intensity cycle ergometer resistive forces for this patient population are not known because guidelines for resistances used with healthy persons are not applicable to patients with a disability in which the TBM to FFM ratio is abnormal. However, further research is needed to pinpoint the optimal resistive force for subgroups such as children and the overweight, and patient populations that include the underweight and the disabled.

The biochemical and neural events associated with high-intensity assessment are also warranted to facilitate a better understanding of the health issues relating to high-intensity exercise ability. The development of the FFM protocol appears to be most attractive in both the clinical and athletic evaluation of high-intensity exercise performance in various subject populations. While this practical solution still requires validation, designing a study to find optimal resistive forces based on FFM for the disabled will be difficult because of the wide spectrum of diseases and levels of residual ability found within and between this specific subject group. However, Van Mil et al. (1996) has reported that an anthropometric estimate of lean tissue volume is a valid predictor of the optimal resistive force during a high-intensity cycle ergometer test in both children and adolescents with neuromuscular disease. The findings of the study are encouraging and should contribute to a greater understanding of high-intensity ability in similar populations and subgroups.

Mechanical Issues

We investigated the mechanical deformity of the cycle ergometer to investigate resistive force transition during the test. A friction-loaded cycle ergometer (Monark 864; Monark AB, Vansbro, Sweden) was used to identify any inaccuracies in the calibration procedure. Saddle heights were adjusted individually to accommodate partial flexion of the knee between 170° and 175° (with 180° denoting a straight leg position) in the middle dead center during the down stroke. Feet were firmly supported by toe clips and straps, and the subject was instructed to remain seated during the test. Individual subjects performed a standardized 5-minute warm-up following procedures outlined by Jaskolska et al. (1999), and 5-minute rest periods were allowed between loads. Prior to data collection, sensor installation was checked to ensure data capture was viable. The calibration method followed the guidelines for friction-loaded ergometers outlined by Coleman (1996). Briefly, the protocol consisted of a series of five calibration tests, using various resistive forces ranging from 1 kg to 2.5 kg, and pedal velocities up to 135 rpm. The tests were performed to obtain moment of inertia and frictional torque regression values that were compatible over several conditions. A correlation coefficient of 0.96 was required prior to data collection. An additional loading range was added that increased resistive force by 0.5 kg until a final resistive force of 10.5 kg was reached. This was included to represent more closely the type of resistive forces that may be encountered during testing for subjects of different body mass. The same pedal velocity was observed over the additional range. This procedure was repeated for all resistive forces 10 times, with 2 rest days separating each calibration trial. Correlations were obtained for each group of four stages and plots were obtained for flywheel decelerations and resistive forces at each individual stage. Values for flywheel deceleration and correlation coefficients were obtained using a computer (Coleman 1996). Data transfer was made possible using a suitably mounted sensor unit and power supply attached to the fork of the ergometer. The sampling frequency of the sensor was 18.2 Hz.

Measurement of Resistive Force

Prior to the calibration procedure, static force was obtained using the range of forces that were used to resistance during the dynamic calibration test. This was established to indicate any differences in resistive force application to the ergometer flywheel that may have occurred between both a static test and during the dynamics of high-intensity performance. Force application measures were quantified using a strain gauge attached to the ergometer cradle braking cord. The strain gauge was interfaced to a computer and tension changes were recorded in volts. This procedure was repeated in both static state conditions and during the dynamic calibration test itself.

The computer was set at zero prior to the application of each load; differences in zero load and applied load were recorded. Graphical illustrations of tension changes were downloaded and saved via a graphical computer package (see Figures 5 and 6).

Conclusions and Future Directions

The total power and relative contribution of the energy systems involved during experimental high-intensity cycle ergometer exercise need re-evaluating. Findings also suggest that the present loading methods used for cycle ergometry that are inclusive of TBM significantly underestimate attainable maximal power outputs. The results of biochemical analysis show that greater PPOs are obtainable with no subsequent differences in neurophysiological or metabolic stress as determined by plasma adrenaline, noradrenaline and blood lactic acid concentrations when resistive forces reflected FFM and not TBM during loading procedures (Baker et al. 2003; see Table 3). In addition, significantly greater muscle damage was observed during the TBM protocol with an accompanying decrease in PPO (Baker et al. 2001b; see Table 4).

The TBM protocol also produced significantly greater oxidative stress with a reduction in PPO compared to the FFM method of resistive force selection (Baker et al. 2004; see Table 5).

The experimental findings indicate that procedures producing realistic power values, which are less damaging and relate to the active muscle tissue utilized during this type of exercise, may need to be explored in preference to methods that include both lean and fat mass. The results also demonstrate significant upper body contributions in the assessment of lower leg power profiles (see Figure 6). In addition, we may need to consider redesigning Monark ergometers for use in high-intensity exercise tests.

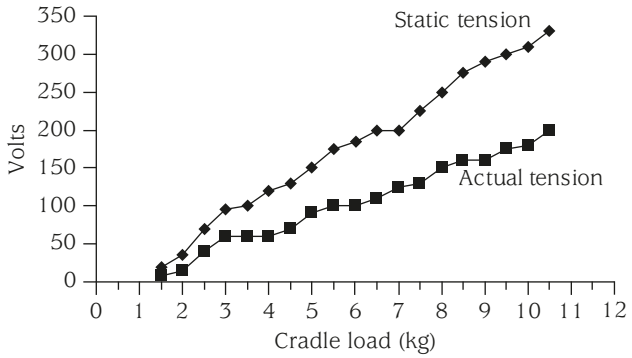

Fig. 5 From the measured forward deflection of the handlebars, there is a resulting decrease in tension of the ergometer rope attached to the ergometer cradle measured by strain gauges. The observed decrease in tension manifests itself as a decrease in resistive force application. The actual forces encountered during the test are almost one third lighter than the forces encountered during a static calibration.

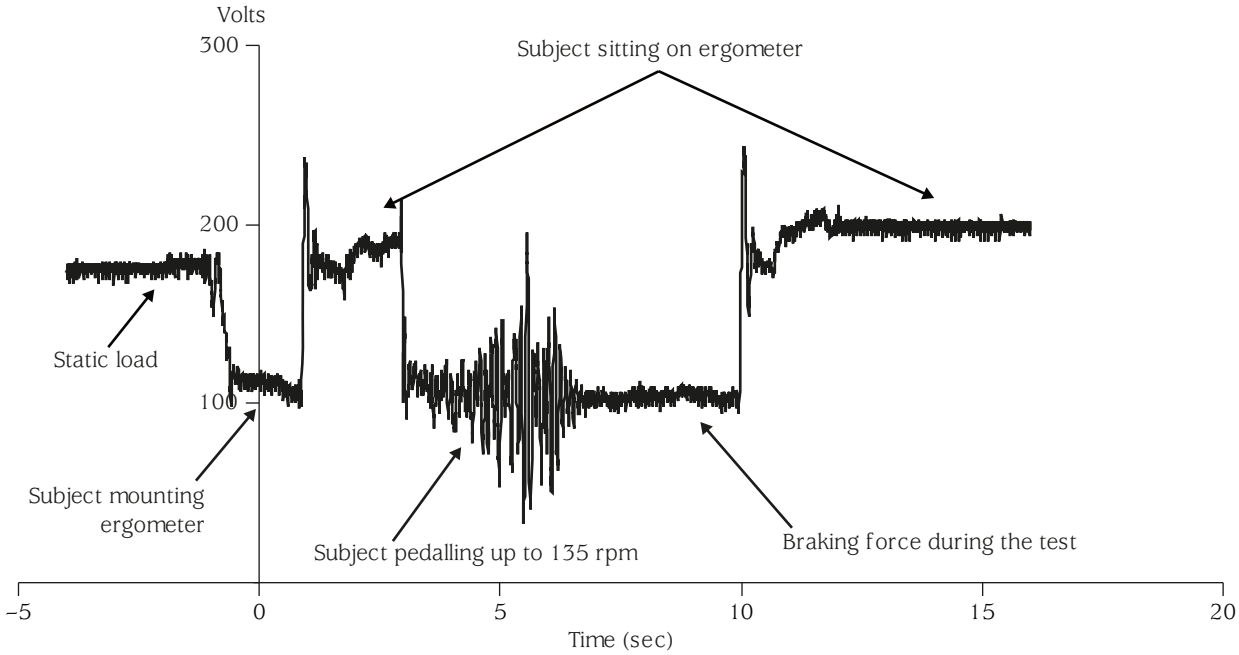

Fig. 6 From the measured forward deflection of the handlebars, there is a resulting decrease in tension of the ergometer rope attached to the ergometer cradle measured by strain gauges. The observed decrease in tension manifests itself as a decrease in resistive force application. This results in discrepancies in load application during the test and spurious power calculation.

Table 3. Adrenaline, noradrenaline and blood lactate concentrations for both the total body mass (TBM) and fat-free mass (FFM) protocols recorded over three blood sampling stages*†‡

Variable	Condition	Pre-exercise	Post-exercise	24 hr post-exercise
Adrenaline ($nmol \cdot L^{-1}$)	TBM	0.3 ± 0.1	2.5 ± 1.4↑	0.3 ± 0.1↓
	FFM	0.3 ± 0.1	3.0 ± 2.0↑	0.3 ± 0.1↓
Noradrenaline ($nmol \cdot L^{-1}$)	TBM	1.3 ± 0.3	17 ± 6.8↑	1.6 ± 0.5↓
	FFM	1.3 ± 0.3	20 ± 9.6↑	1.3 ± 0.5↓
Blood lactate ($mmol \cdot L^{-1}$)	TBM	0.5 ± 0.7	9.0 ± 1.2↑	0.6 ± 0.6↓
	FFM	0.5 ± 0.7	9.3 ± 1.4↑	0.7 ± 0.8↓

*Data are presented as mean ± standard deviation; †increases ($p < 0.05$↑) were recorded for adrenaline, noradrenaline and blood lactate concentrations for both the total body mass (TBM) and fat-free mass (FFM) protocols from pre- to immediately post-exercise; ‡decreases in concentration ($p < 0.05$↓) were observed from immediately post- to 24 hours post-exercise. No differences were observed between the TBM and FFM protocols for any of the blood sampling stages.

Table 4. Creatine kinase (CK), myoglobin (Mb) and cardiac troponin (cTnI) concentrations for total body mass (TBM) and fat-free mass (FFM) protocols measured at rest, immediately post- and 24 hours post-exercise*

Variable	Condition	Pre-exercise	Post-exercise	24 hr post-exercise
CK ($\mu \cdot L^{-1}$)	TBM	202 ± 162	236 ± 213↑†	175 ± 110↓
	FFM	157 ± 81	172 ± 93↑†	136 ± 67↓
Mb ($ng \cdot mL^{-1}$)	TBM	53 ± 22.1	54.5 ± 25.4↑†	49.7 ± 12.4↓
	FFM	46 ± 13.9	46.3 ± 13†	42 ± 7.5
cTnI ($ng \cdot mL^{-1}$)	TBM	0.06 ± 0.04	0.05 ± 0.04	0.03 ± 0.02
	FFM	0.02 ± 0.03	0.04 ± 0.03	0.05 ± 0.06

*Data are presented as mean ± standard deviation; †significant ($p < 0.05$) between TBM and FFM for condition indicated. Increases ($p < 0.05$↑) in concentration from rest to immediately post-exercise were observed for CK during both the TBM and FFM protocols. The greater concentrations were recorded for TBM and were different when compared to FFM ($p < 0.05$†). Concentrations decreased 24 hours later ($p < 0.05$↓). Differences in concentrations ($p < 0.05$↑†) were observed between groups immediately post-exercise for Mb, with the highest values recorded for TBM. Concentrations decreased 24 hours later ($p < 0.05$↓). There were no differences observed for cTnI under any condition or blood sampling stage.

Table 5. Lipid peroxidation (LH) and malondialdehyde (MDA) concentrations in response to high-intensity cycle ergometry for total body mass (TBM) and fat-free mass (FFM) protocols over three experimental conditions*

Variable	Condition	Pre-exercise	Post-exercise	24 hr post-exercise
LH ($\mu mol \cdot L^{-1}$)	TBM	1.09 ± 0.31	1.5 ± 0.45↑†	0.97 ± 0.36↓
	FFM	1.1 ± 0.56	1.2 ± 0.37†	1.2 ± 0.37
MDA ($\mu mol \cdot L^{-1}$)	TBM	0.6 ± 0.1	0.83 ± 0.18↑†	0.66 ± 0.17↓
	FFM	0.6 ± 0.20	0.76 ± 0.19†	0.55 ± 0.13

*Data are presented as mean ± standard deviation; †significant changes ($p < 0.05$) between TBM and FFM for condition indicated. LH and MDA concentrations increased from rest to immediately post-exercise ($p < 0.05$↑) in the TBM protocol only. Differences ($p < 0.05$†) were also noted between the TBM and FFM protocols immediately post-exercise for both LH and MDA. There were no differences observed between the two groups' pre- and 24 hours post-exercise blood sampling stages. Concentrations of LH and MDA returned to pre-exercise values 24 hours later ($p < 0.05$↓).

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