Muscle Power Output into Pedals Exceeds Friction-Load Determined Power of a Cycle Ergometer

Peter Schneider,*1 Norman Schneider2

1,2 University Clinic, University of Würzburg
Oberdürbacherstr.6, 97080 Würzburg, Germany

*schneider_p@ukw.de; ichnorman@web.de

Abstract—It was hypothesized that the muscle power input created by both legs would far exceed the ergometers’ constant-torque-based nominal value. It was supposed that this effect depended on the cadence and positive or negative acceleration of the eddy current brake wheel. If so, it is important to estimate the magnitude of this effect in order to protect patients who are at risk of overload. Four normally-trained subjects performed a series of tests involving ergometer pedaling with either ascending power loads in the range of 25 – 250 Watts, or a variation in cadence at intervals of 100 and 150 Watts. The ergometer’s nominal power was compared to the power calculated with the measured time series recordings of the pedal force sensors and the power calculation thereof. Highly significant discrepancies were found between the nominal power load of the ergometer and the power transferred through the pedal sensors. The pedal power was up to 88% higher than the nominal loads, and there were inter-individual differences. The measured power far exceeded the nominal value by 80-166% at a low cadence of 40, and was only approximately congruent at a cadence of 55 to 65 per minute. Confirming the hypothesis, the results suggested that the excess power requirement was mainly due to acceleration power and depended on individual muscular performance. It is important to be aware of this when patients at risk undergo ergometer testing. Power estimation using a direct pedal force measurement may provide new insight into sport-specific stationary cycling science and energy metabolism.

Keywords—Ergometer; Muscle; Force; Power; Power Spectrum; Bicycle

I. INTRODUCTION

Stress tests using bicycle ergometry are very common and well established [1]. In clinical routines for nuclear medicine, patients who are at risk of having difficulties performing this test are frequently observed. Subjectively, it appears that some patients work exhausting hard, particularly during the moment of load increase. This observation raised suspicions that under certain circumstances, such as during crank cycles, more power is required than the calibrated nominal load suggests. Ergometer testing could expose at-risk patients to increased danger because physicians are accustomed to relying on the ergometer’s calibrated power settings.

A number of studies reported that cycling biomechanics are affected by many anthropometric parameters, remaining on a theoretical level [2-8]. It has been shown that pedal torque strongly depends on pedaling cadences, and that there is an important reduction in mean torque at higher cadences [8]. In addition, at an electronically controlled and eddy current braked ergometer power settings, the forces vary over the crank angle in such a way that lower revolutions per minute (rpm) require higher accelerating pedal forces to produce the required torque. Such measurements were collected at the crank but not at the point where the force was directed into each pedal axis. While pedaling cadence and crank cycles vary, the electric motors used to calibrate ergometers produce a constant torque and velocity at a given power setting.

This lead to the hypothesis that the forces created and the power transferred into the pedals by a pedaling human requires additional acceleration work per time (= power) to compensate for the deceleration at the top and bottom dead points of a crank cycle. The compensating power should depend on the eddy current brake load or inconvenient cadence. Measuring the differences between the calibrated ergometer brake power and the power generated by the cyclist’s muscles provided a method that could calculate the power transferred directly into each pedal. Clinical implementation of such an approach would be beneficial because the actual produced power could be estimated more realistically.

II. METHODS

A. Device

Force sensors for measuring the tension and compression forces (K-2145 type, Lorenz Meßtechnik GmbH, Alfdorf, Germany) were installed on the pedal cranks of an electronically controlled semi-reclined ergometer (Ergoselect 1000 P/K, ergoline GmbH, Bitz, Germany). The sensors (Fig. 1) recorded the force vector perpendicularly, acting onto the pedal axis at a sampling rate of 100 Hz. The measuring range of factory-calibrated cells was 0.5 to 1000 Newtons, and the accuracy was 0.25%. The brake-load-dependant pedal forces were recorded as a function F(t) of time representing the cadence, and were then transformed into the frequency domain [9] to obtain the instantaneous power \( \Pi \) from the power spectral density of the quasi-periodic time series according to the formula \( \Pi = \frac{1}{2\pi} \sum_{\omega} X^2 \), with \( X \) representing the rotational ergometer pedal displacement and \( F = F \sin \omega t \) representing the force on the sensor system. The

DOI: 10.5963/BER0401001
time averaged power was obtained by using the complex numbers; hence, it was $\langle \Pi \rangle = \frac{1}{T} \int \Pi(t) \, dt$. Because the power spectral curve had the units of volts$^2$ per hertz (the bare sensor voltage signals), the area under the power spectral curve had the unit of volts$^2$. Therefore, electrical power was proportional to volts$^2$ [10]. Because of the high linearity of the piezoelectric force sensors ($< \pm 1\%$), the electrical power derived from the system was also linearly proportional to the mechanical power as calculated according to Parseval’s theorem [10-13]. Fig. 2 shows the block diagram of the system. The software implementation was straightforward using well-known algorithms [9]. The ergometer application was granted a German patent [14].

![Fig. 1 Sensor equipped ergometer pedal](image1)

![Fig. 2 Block diagram of experimental setting](image2)

### B. Subjects

Four healthy, trained subjects, ages 23 and 62 (male) and ages 21 and 23 (female), performed the tests on the ergometer. The tests consisted of cycling on the semi-reclined ergometer. The institutional Ethics Committee had no concerns (decision #05-146), the study was in conformity with the 8th declaration of Helsinki, and written consent was obtained.

### C. Data Acquisition

Force sensor data were continuously collected during each test cycle, which lasted up to 650 seconds and began with a 3-minute warm-up at 50 Watts and at any rpm. Thereafter, the ergometer was set up to break loads at 25 Watt increments every 60 seconds, beginning with 25 Watts to a maximum of 250 Watts over 60 seconds. Then the eddy current brake wheel was unloaded. The test runs were repeated on different days. The subjects performed the initial tests at 60 ±0.02 rpm. To consolidate or reject the hypothesis, subsequent tests were performed at four different cadences (40, 50, 60, and 70 rpm) at a preset load of 100 and 150 Watts. An acceleration phase of 5 seconds preceded each step, which lasted a total of 45 seconds.
D. Statistics

The precision of the power calculation from data sequences at the specified load settings was 7% as estimated from the analysis of consecutive 5-second intervals, and 2% from intervals increased by 5 seconds to the full length of 40 seconds. The variation excluded the 5-second acceleration interval to reach the next rpm level. The total power input into the ergometer was the sum of each pedal. STATISTICA V10.0 (StatSoft, Inc.) was used to generate the graphs and calculate the correlations and precision.

III. RESULTS

At preset ergometer loads, the pedal-force estimated power strongly depended on the cadence. Any drop in rpm during a pedaling cycle at dead point crossing obviously needed to be regained by additional acceleration forces. At a cadence of 60 rpm, the power showed an almost linear increase above 125 Watts compared to the nominal break power (Fig. 3). While the sensor-derived power fell slightly short of the nominal power at a setting below 125 Watts, there was some, although scattered, agreement around the nominal value of 125 Watts. At a 250-Watts nominal load, the sensor measurements were 88% above that value. The subjects showed very congruent intra-individual results in the test runs, but there were inter-individual differences of up to 30% (Figs. 3 and 4a). The lower the cadences, the higher the measured pedal power. At 100 Watts, it ranged from 66% at 70 rpm to 177% at 40 rpm (Table 1) compared to the nominal load. At 150 Watts, the observation was similar. Finally, the power during each 5 seconds of acceleration towards the required rpm was considerably higher than the power during the subsequent 40 seconds of constant rpm (Fig. 4b).

| Average power over 40 seconds | 100 Watts nominal | 150 Watts nominal |
|-------------------------------|------------------|------------------|
| Subject 1 | Subject 2 | Subject 1 | Subject 2 |
| rpm | right | left | sum | right | left | sum | right | left | sum | right | left | sum |
| 40 | 92 | 85 | 177 | 100 | 72 | 172 | 212 | 176 | 388 | 200 | 205 | 405 |
| 50 | 59 | 56 | 115 | 70 | 57 | 127 | 137 | 97 | 234 | 138 | 125 | 263 |
| 60 | 46 | 30 | 76 | 53 | 45 | 98 | 92 | 72 | 164 | 94 | 92 | 186 |
| 70 | 37 | 29 | 66 | 42 | 33 | 75 | 70 | 54 | 124 | 72 | 72 | 144 |

Acceleration power to the designated ergometer load over 5 seconds

| 100 Watts nominal | 150 Watts nominal |
|------------------|------------------|
| rpm | right | left | sum | right | left | sum | right | left | sum | right | left | sum |
| 40 | 122 | 84 | 206 | 82 | 112 | 194 | 264 | 267 | 531 | 215 | 253 | 468 |
| 50 | 150 | 98 | 248 | 142 | 97 | 239 | 189 | 201 | 390 | 222 | 297 | 519 |
| 60 | 66 | 60 | 126 | 92 | 100 | 192 | 165 | 110 | 275 | 180 | 192 | 372 |
| 70 | 55 | 49 | 104 | 93 | 61 | 154 | 101 | 108 | 209 | 114 | 137 | 251 |

Fig. 3 Total power delivered by both legs as transferred through the pedals during an ergometer test cycle using steps from 25 to 250 Watts at increments of 25 and 50 Watts
Fig. 4a Total power delivered by both legs as transferred through the pedals during an ergometer test cycle using rpm steps from 40 to 70 rpm at increments of 10 rpm, as calculated from the recorded interval of 40 seconds during the constant rpm phase at nominal brake loads of 100 and 150 Watts

Fig. 4b Acceleration power delivered by both legs during the test cycle acceleration (ac) phase of 5 seconds using rpm steps from 40 to 70 rpm at increments of 10 rpm. The excess power compared to Fig. 4a is due to cyclic braking effects during the load shift from one leg to the other on the dead point of each cycle at nominal brake loads of 100 and 150 Watts

The intra-individual correlations of the exercise runs were $r = 0.9991$ for the load variation tests (Fig. 3). The deviations from the nominal ergometer load can be described by the polynomial fit: $\Pi_{SE} = 8 + 0.291 \cdot \Pi_{EL} + 0.001 \cdot \Pi_{EL}^2 + 4.412 \cdot \Pi_{EL}^3 - 9.967 \cdot \Pi_{EL}^4$ ($\Pi_{SE} =$ sensor-derived and $\Pi_{EL} =$ ergometer-load derived power). The inter-individual correlation was $r = 0.9979$ (100 Watts) and 0.9988 (150 Watts) for the rpm variation tests (Fig. 4a), all $p < 0.0001$. The cadences during the relevant test cycles were kept constant at ±2 digits during all rpm settings.

IV. DISCUSSION

Neither the accuracy of the sensor method nor that of the ergometer is in question. The calibration of any medically used ergometer must be inspected biannually to comply with European directives 93/42/EEC and 2007/47/EC. Engineers and physicists of the patent offices have evaluated the validity of the sensor method.

The test results confirmed the hypothesis that discrepancies between the bicycle ergometer’s preset power and directly sensor-measured power input were mainly due to the constant-torque calibration method using an electric motor versus direct muscular force and velocity measurement. This difference may be attributed to pedaling kinematics and efficiency. A continuous load shift from one leg to the other passes a dead point during zero force, causing deceleration of the rotating parts [3]. In contrast to a constant-torque electric motor, pedaling requires additional acceleration torque, contributing acceleration power to the constant-torque defined nominal power, as observed in Fig. 4b. A low ergometer power setting or a high moment of inertia at high cadence may compensate for the braking effect [5, 7]. As theoretically investigated, the inertial energy balances against the brake effect until the muscles become effective following the dead points of each cycle. An equation for calculating the maximum power was suggested, which takes the flywheel inertia into account to avoid significant underestimation of power [6]. These experiments provided evidence of this theoretical underestimation or, in other words, the efficiency.

Previous studies investigated the crank torque profile [8] or the power components of a complex dynamic pedaling model, revealing the calculated net power using limb components, gravity, and velocity [7, 15]. Studies have demonstrated that the
calculated instantaneous peak power of muscle groups do not equal the average power over crank cycles, and many geometrical variables influence the simulated power estimation [5].

There are various explanations for the optimal cadence ranging from 60 to 105 rpm [5, 12]. At a higher cadence (higher inertial moment), the brake load may be lower, helping to decrease deceleration during a pedaling cycle. Other studies have assessed effects within or after the end of the chain of the torque transmission, involving several power-consuming items, such as multiple bearings and chain friction, lowering the efficiency of the power input at the pedal force sensor side [16, 17]. One study used two different methods to compare the power before and after the pedal input side, which was, in principal, in agreement with the findings of this study [16]. A more accurate muscle-related power estimation system may be interesting in some experimental settings, considering that forefront measurement at the pedals still does not reflect the internally generated muscle power of the limbs [18]. Although this pedal-sensor system measured uni-axial forces, the inclusion of lateral forces would have a negligible contribution, strengthening the conclusions.

V. CONCLUSIONS

In summary, it was shown that the power input from subjects was much higher at a low rpm or at high nominal load settings for a semi-reclining ergometer than its standard calibration suggested. This result confirmed the hypothesis that periodic acceleration work over time is the main source that requires power in excess of the nominal constant-torque power. This observation may not only be relevant for sport-specific clarifications of the physiological demands encountered by subjects and their energy metabolism during exercise, but may also be relevant in several clinical fields for which ergometry is part of the physical examination of at-risk patients such as coronary heart disease patients. A more accurate power performance estimation would be beneficial and deserves further investigation.

REFERENCES

[1] L.J. Nebelsick-Gullet, T.J. Housh, G.O. Johnson, and S.M. Bauge, “A comparison between methods of measuring work capacity,” Ergonomics, vol. 31, pp. 1413-1419, 1988.
[2] L.M. Rogers, D.A. Brown, and K.G. Gruben, “Foot Force direction control during leg pushes against fixed and moving pedals in persons post-stroke,” Gait and Posture, vol. 19, pp. 58-68, 2004.
[3] C.C. Raasch, F.E. Zajak, B.M. Ma, and W.S. Levine, “Muscle coordination of maximum-speed pedaling,” J Biomech, vol. 30, pp. 595-602, 1997.
[4] H. Gonzales and M.L. Hull, “Multivariable optimization of cycling biomechanics,” J Biomech, vol. 22, pp. 1151-1161, 1989.
[5] Y. Yushihuku and W. Herzog, “Optimal design parameters of the bicycle-ride system for maximal muscle power output,” J Biomech, vol. 23, pp. 1069-1079, 1990.
[6] J.B. Morin and A. Belli, “A simple method for measurement of maximal downstroke power on friction-loaded cycle ergometer,” J Biomech, vol. 43, pp. 141-145, 2004.
[7] V.J. Fregly and F-E. Zajak, “A state-space analysis of mechanical energy generation, absorption, and transfer during pedaling,” J Biomech, vol. 29, pp. 81-90, 1995.
[8] W. Bertucci, F. Grappe, A. Girard, A. Betik, and J.D. Rouillon, “Effects on the crank torque profile when changing pedalling cadence in level ground and uphill road cycling,” J Biomech, vol. 38, pp. 1003-1010, 2005.
[9] A.V. Oppenheim and R.W. Schafer, “Digital Signal Processing,” Englewood Cliffs, NJ, Prentice Hall, pp. 420-429, 1975.
[10] M. Norton and D. Karczub, “Fundamentals of Noise and Vibration Analysis for Engineers,” Cambridge: Cambridge University Press, pp. 37-56, 2003.
[11] Gerard Maral, “VSAT networks,” 2nd ed., Chichester: John Wiley & Sons, pp. 19-20, 2003.
[12] P. Schneider, H. Hänscheid, M. Schwab, and F. Jakob, “Assessment of neuromuscular function with a new ground reaction force platform using power spectrum analysis technique,” in: IFMBE Proc. WC 2009, O. Dössel, W.C. Schlegel, Eds., Germany: Springer-Verlag, 2009.
[13] P. Schneider, M. Schwab, and H. Hänscheid, “Identification of some factors associated with risk of fall using a force platform and power spectrum analysis technique,” J Biomech, vol. 44, pp. 2008-2012, 2011.
[14] P. Schneider, “Ergometer, running shoe, bicycle pedal,” Patent DE102006032081 B4, Sept. 26, 2013.
[15] S.M. Gregor, K.L. Perell, S. Ruhakantanovit, E. Miyamoto, R. Muffoletto, and R.J. Gregor, “Lower extremity general muscle moment patterns in healthy individuals during recumbent cycling,” Clin Biomech, vol. 17, pp. 123-129, 2002.
[16] C.R. Abiss, M.J. Quod, G. Levin, D.T. Martin, and P.B. Laursen, “Accuracy of the Velotron Ergometer and SRM Power Meter,” Int J Sports Med., vol. 30, pp. 107-112, 2009.
[17] D.A. Coleman, J.D. Wiles, R.C.R. Davison, M.F. Smith, and I.L. Swaine, “Power output measurement during treadmill cycling,” Int J Sports Med., vol. 28, pp. 525-530, 2007.
[18] S.A. Haapala, P.D. Faghi, and D.J. Adams, “Leg joint power output during progressive resistance FES-LCE cycling in SCI subjects: developing an index of fatigue,” J Neuroeng Rehabil, vol. 5, p. 14, 2008.