Total power output generated during dynamic knee extensor exercise at different contraction frequencies

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Ferguson, Richard A., Per Aagaard, Derek Ball, Anthony J. Sargeant, and Jens Bangsbo. Total power output generated during dynamic knee extensor exercise at different contraction frequencies. J Appl Physiol 89: 1912–1918, 2000.—A novel approach has been developed for the quantification of total mechanical power output produced by an isolated, well-defined muscle group during dynamic exercise in humans at different contraction frequencies. The calculation of total power output comprises the external power delivered to the ergometer (i.e., the external power output setting of the ergometer) and the “internal” power generated to overcome inertial and gravitational forces related to movement of the lower limb. Total power output was determined at contraction frequencies of 60 and 100 rpm. At 60 rpm, the internal power was 18 ± 1 W (range: 16–19 W) at external power outputs that ranged between 0 and 50 W. This was less (P < 0.05) than the internal power of 33 ± 2 W (27–38 W) at 100 rpm at 0–50 W. Moreover, at 100 rpm, internal power was lower (P < 0.05) at the higher external power outputs. Pulmonary oxygen uptake was observed to be greater (P < 0.05) at 100 than at 60 rpm at comparable total power outputs, suggesting that mechanical efficiency is lower at 100 rpm. Thus a method was developed that allowed accurate determination of the total power output during exercise generated by an isolated muscle group at different contraction frequencies.

Different approaches have been used for the quantification of energy turnover and mechanical efficiency during exercise in humans. During whole body exercise, such as cycling, pulmonary oxygen uptake (VO₂) has been measured and used as an estimation of energy turnover. This has then been related to the mechanical output during exercise, which, in cycling, is usually determined as the external power delivered to the ergometer (e.g., Ref. 7). This approach, however, clearly underestimates the work performed because the power generated to overcome the inertial and gravitational forces opposing the movement of the limb, the so-called “internal” work, should also be taken into consideration (15). For example, values for mechanical efficiency during cycle exercise, defined as the ratio between external mechanical work and net energy expenditure, ranged between 18 and 22% at a range of work rates (9). When both external and internal mechanical work were determined, efficiency varied between 21 and 30% (9).

Measurements of pulmonary VO₂ during whole body exercise do not, however, provide accurate information about the energy turnover and mechanical efficiency of the working muscles, because tissues other than the skeletal muscle contribute to energy turnover. Another problem in determining the energy turnover of the working skeletal muscles is that it is difficult to quantify the mass of the muscles involved in the exercise and to determine to what extent these muscles are active (12). To determine the energy turnover of the active muscles during exercise in humans, an isolated muscle model needs to be used. One such model is single-leg, dynamic knee-extension exercise (2), which allows a functionally significant muscle group (i.e., quadriceps) to be investigated in a well-defined and reproducible movement and in which the active muscle mass can be readily estimated. Energy turnover can be determined from measurements of thigh blood flow, femoral venous and arterial blood gas and metabolites, as well as changes in metabolites in the contracting quadriceps muscle (3, 4).

The external power delivered to the ergometer during knee-extensor exercise can be determined with high precision, and estimates of mechanical efficiency have been made (~24%) (3). However, in these calculations, the internal work was not taken into account. This needs to be carefully evaluated if this exercise model is to be used to determine the true mechanical efficiency of contracting muscle in vivo. This is even

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more obvious if measurements are to be made during exercise at different contraction frequencies. During cycle exercise, for example, it has been shown that internal work is greater as contraction frequency increases (13, 14). There is, however, no clear picture as to how contraction frequency affects mechanical efficiency. Some investigations report a decline in mechanical efficiency (7), whereas others report an increase (5) as contraction frequency increases.

Therefore, one of the aims of this study was to develop a method to quantify the total mechanical power output produced during exercise of an isolated muscle group in humans, i.e., during dynamic, single-leg knee-extensor exercise. It was proposed that external power alone would markedly underestimate the actual total power developed. Another aim of the study was to evaluate whether there was any difference in energy turnover during exercise at the same total power output at two different contraction frequencies (60 and 100 rpm). The study demonstrated that the method can be used to determine the internal and total power developed by an isolated muscle group during exercise at different contraction frequencies. It was also shown, based on pulmonary VO\textsubscript{2}, that the energy cost of exercise is higher at 100 compared with that at 60 rpm at the same total power output, suggesting that mechanical efficiency is lower at the higher contraction frequency.

METHODS

Subjects. Twelve healthy men volunteered to participate in the investigation. All subjects were physically active, but none was specifically trained, and all gave written, informed consent. Mean (±SE) age, height, and body mass were 24 ± 1 yr, 180.4 ± 1.8 cm, and 75.4 ± 2.4 kg, respectively. The study was approved by the local Ethics Committee.

Exercise model. Single-leg, dynamic knee-extension exercise (2) was performed in a supine position, with the subjects secured to the seat with straps across the torso, hips, and both thighs (Fig. 1). The subjects were instructed to contract the knee-extensor muscles during extension and to completely relax all muscles during passive flexion. During habituation, consisting of at least three, 30-min sessions, the subjects were fully familiarized to the experimental procedures.

Total power determination. The determination of total muscle power required continuous measurements of the force exerted between the moving lower limb and the pedal arm of the ergometer, which was recorded by a strain gauge incorporated in the rod attachment of the ergometer (Fig. 1). It also required continuous measurement of knee joint angle and the angle between the lower limb and the connecting bar attached to the pedal arm of the ergometer. Therefore, a flexible electrogoniometer (type XM180, Penny and Giles Biometrics, Gwent, UK) was placed across the lateral aspect of the knee joint and attached proximally to the strapping that firmly secured the thigh and distally to just below the lateral epicondyle of the tibia (Fig. 1). A second electrogoniometer was positioned between the lower limb (i.e., the boot attachment) and the connecting bar of the ergometer (Fig. 1). During exercise at both 60 and 100 rpm, the angular displacement of the knee ranged between 90 and 170° at flexion and extension. In situ calibration of the electrogoniometer at the knee was performed immediately before and after each exercise bout by moving the lower limb between a fully extended and a 90° flexed position. During experiments, the strain gauge and electrogoniometer signals were recorded onto the hard drive of a computer (Apple Macintosh Performa 5200) using an A-D board with a 400-Hz analog-to-digital sampling rate (MacLab 8s data acquisition system, Chart v3.3 software, ADInstruments, Sydney, Australia).
the subsequent analysis, electrogoniometer signals were smoothed by means of a digital, fourth-order, zero-lag Butterworth low-pass filter (16), using a cut-off frequency of 3 Hz.

Typical changes in joint angle and strain gauge output from one representative subject during exercise at an external power output of 30 W at either contraction frequency are illustrated in Fig. 2. At 60 rpm, from the beginning of extension, there is a rapid increase in strain gauge force that peaks at approximately midextension. As extension continues, the force declines but then begins to rise slightly at the transition between extension and flexion. As flexion continues, there is a slight negative force. At 100 rpm, from the beginning of extension, there is an increase in strain gauge force that initially peaks at midextension. As flexion begins, there is a sharp and rapid increase in force as the momentum of the flywheel causes the connecting rod to reverse direction, thus overcoming the kinetic energy of the leg generated during the fast extension phase. There is a rapid decline in force during the flexion phase, with a large negative peak before the extension phase begins again.

Instantaneous muscle power was calculated as the product of knee joint angular velocity (rad/s) and total muscle moment (16). In accordance with previous reports (1, 10), total muscle moment \( M_{\text{total}} \) was calculated as

\[
M_{\text{total}} = M_{\text{ergometer}} + M_{\text{gravity}} + M_{\text{inertia}} \tag{1}
\]

where \( M_{\text{ergometer}} \) was the muscle moment generated to overcome the resistance of the ergometer flywheel system, \( M_{\text{gravity}} \) was the muscle moment produced to overcome the force of gravity acting on the lower limb, and \( M_{\text{inertia}} \) was the muscle inertial moment depending on the acceleration and deceleration of the lower limb.

\( M_{\text{ergometer}} \) was calculated as

\[
M_{\text{ergometer}} = F_{sg} \cdot L \cdot \sin(\phi_{\text{pull}}) \tag{2}
\]

where \( F_{sg} \) was the force applied between the ergometer system and the lower limb, as measured by means of the strain gauge positioned in the connecting bar; \( L \) was the length of the lower limb segment (lateral femoral condyle-lateral malleolus) plus an additional 6.2 cm distance from the lateral malleolus to the bar-lower limb connection; and \( \phi_{\text{pull}} \) was the angle between the connecting bar and the lower limb, as measured by the electrogoniometer (Fig. 1).

\( M_{\text{gravity}} \) was calculated as

\[
M_{\text{gravity}} = m \cdot g \cdot r \cdot \sin(\phi_{g}) \tag{3}
\]

where \( m \) was the mass of the lower limb (shank-foot), estimated as 0.061 \cdot body mass (16) + the boot’s mass (0.622 kg); \( g \) was acceleration due to gravity (9.81 m/s\(^2\)); \( r \) was the distance (radius) from the knee joint axis to the center of mass of the lower limb, estimated as 0.606 \cdot L (16); and \( \phi_{g} \) was
the angle of the lower limb relative to vertical, as measured by the electrogoniometer (Fig. 1).

\[ M_{\text{inertia}} = I\alpha \]  \hfill (4)

where \( I \) was the moment of inertia of the lower limb (shank-foot) relative to the knee joint axis, estimated by \( I = mr_G^2 \) (16), with the radius of gyration \( (r_G) \) estimated as 0.735 \( z \) (15); and \( \alpha \) was the angular acceleration of the lower limb (in rad/s²), as obtained by double differentiation of the signal measured by the electrogoniometer (Fig. 1).

Figure 3 illustrates the contribution from each moment to the determination of \( M_{\text{total}} \) and the calculation of instantaneous muscle power output during a typical single flexion-extension cycle for an external power output of 30 W at 60 and 100 rpm. The magnitude of \( M_{\text{ergometer}} \) during the leg extension phase was slightly lower at 100 compared with at 60 rpm. Because the determinants of \( M_{\text{gravity}} \) do not change between the conditions examined (see Eq. 4) the magnitude of \( M_{\text{gravity}} \) was the same at 60 and 100 rpm. As expected from the differences in acceleration, \( M_{\text{inertia}} \) was greater at 100 compared with 60 rpm. Thus the greater \( M_{\text{total}} \) at 100 rpm was primarily a result of the higher \( M_{\text{inertia}} \). The instantaneous angular velocity was obviously higher at 100 rpm. Thus the product of \( M_{\text{total}} \) and angular velocity, resulting in the instantaneous muscle power output, had a greater magnitude during the extension phase at 100 compared with at 60 rpm.
Total power output was determined for each extension cycle by integration of the instantaneous power output curve and by subsequent division by the time period of integration. From each exercise trial, five sections of data at the end of each minute of exercise were analyzed, and each section was 10 s in duration. The total power output values from each individual extension cycle were averaged to obtain a mean total power output for each section. The five sections analyzed for each trial were further averaged to obtain the mean total power output for the trial. Internal power output was calculated as the difference between the measured total power output and the external power output delivered to the ergometer.

Experimental protocol. In the first part of the study, subjects performed, in randomized order, exercise at external power outputs (i.e., ergometer power settings) of 0, 10, 30, 40, and 50 W at contraction frequencies of 60 and 100 rpm. Each exercise bout was performed for 5 min, with at least 15 min of passive rest separating the bouts. Total power output was calculated for each exercise bout as described in Total power determination.

In the second part of the study, an attempt was made to match the total power output during exercise at the two contraction frequencies. To do this, the data for each individual subject from the first part of the study were used to establish a linear relationship between total and external power outputs for each contraction frequency. Initially, the external power for exercise at 60 rpm was set at 30 W, and the total power output for this was predicted using the regression equation for 60 rpm. To match this total power output, the external power output at 100 rpm was predicted from the regression equation between total and external power at 100 rpm. With these two external power outputs, exercise was performed for 5 min, again in a randomized order, with a 30-min rest period between each bout. Total power output was again calculated for each exercise bout, as described in Total power determination.

Table 1. Internal power output generated during knee-extensor exercise at 60 and 100 rpm at five external power outputs

<table>
<thead>
<tr>
<th>External power, W</th>
<th>60 rpm</th>
<th>100 rpm</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>16 ± 1</td>
<td>38 ± 3*</td>
</tr>
<tr>
<td>10</td>
<td>18 ± 2</td>
<td>38 ± 3*</td>
</tr>
<tr>
<td>30</td>
<td>19 ± 3</td>
<td>32 ± 3*†</td>
</tr>
<tr>
<td>40</td>
<td>18 ± 2</td>
<td>27 ± 4†</td>
</tr>
<tr>
<td>50</td>
<td>17 ± 2</td>
<td>29 ± 4†</td>
</tr>
</tbody>
</table>

Values are means ± SE (n = 12). Internal power output was calculated as the difference between the measured total power output and the external power output (i.e., the external power output setting of the ergometer). *P < 0.05, 100 compared with 60 rpm; †P < 0.05, compared with 0 and 10 W.

RESULTS

Relationship between external and total power output. Total power output increased as external power increased from 0 to 50 W and was significantly (P < 0.05) greater at 100 compared with 60 rpm at all external power outputs (Fig. 4). At 60 rpm, the mean internal power output (calculated as the difference between total and external power) was 18 ± 1 W (range: 16–19 W) and did not change as external power output increased (Table 1). At 100 rpm, the mean internal power output was 33 ± 2 W (27–38 W), with internal power being lower (P < 0.05) at an external power of 30, 40, and 50 W compared with 0 and 10 W (Table 1). No correlation was observed between lower limb dimensions (i.e., limb length, estimated mass, and
moment of inertia) and the level of internal power output at any of the external workloads.

**Matched total power at 60 and 100 rpm.** In the second part of the investigation, we sought to match the total power produced during exercise at 60 and 100 rpm. For an external power output of 30 W at 60 rpm, the regression analysis resulted in a target total power output of 45\(\pm\)6 W (38–54 W) (Table 2). To match this total power output during exercise at 100 rpm, an external power output of 14\(\pm\)6 W (7–26 W) was predicted to be required (Table 2). When exercise was performed at each respective external power output for either 60 or 100 rpm, there was no difference in the measured total power output between the two contraction frequencies (Table 2). In addition, there was no difference between the target total power output and the measured total power output during these exercise bouts (Table 2).

Table 2. **External and total power outputs during knee-extensor exercise at 60 and 100 rpm to produce the same total power output**

<table>
<thead>
<tr>
<th>Subject</th>
<th>External Power</th>
<th>Measured Total Power</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>60 rpm</td>
<td>100 rpm</td>
</tr>
</tbody>
</table>
| 1       | 30     | 16     | 44     | 42           | 43     
| 2       | 30     | 21     | 54     | 50           | 56     
| 3       | 30     | 7      | 50     | 52           | 50     
| 4       | 30     | 10     | 39     | 49           | 44     
| 5       | 30     | 9      | 48     | 47           | 50     
| 6       | 30     | 9      | 42     | 48           | 36     
| 7       | 30     | 8      | 43     | 46           | 48     
| 8       | 30     | 26     | 38     | 40           | 47     
| 9       | 30     | 17     | 47     | 50           | 46     
| Means \(\pm\) SE | 30     | 14\(\pm\)2 | 45\(\pm\)2 | 47\(\pm\)1 | 47\(\pm\)2 |

All power output values are in Watts. Pred values were predicted from the regression equation between total and external power output at 100 rpm.

**Pulmonary \(\dot{V}O_2\).** At 100 rpm, pulmonary \(\dot{V}O_2\) was ~0.3 l/min higher \((P < 0.05)\) at all external power outputs than at 60 rpm (Fig. 5A). Even when \(\dot{V}O_2\) was expressed in relation to total power output, the oxygen cost at 100 rpm was greater than at 60 rpm at all power outputs (Fig. 5B). In the second part of the study, when exercise was performed at the same total power output, pulmonary \(\dot{V}O_2\) was ~0.1 l/min higher \((P < 0.05)\) at 100 compared with at 60 rpm.

**DISCUSSION**

The single-leg knee-extensor exercise model provides a unique opportunity to study the physiological mechanisms relating to human muscle metabolism and peripheral cardiovascular control in vivo (2–4). In the present study, a method was developed to quantify the total mechanical power output produced during dynamic exercise at different contraction frequencies using the knee-extensor model. It was demonstrated that internal power contributes significantly to the total power output. At a contraction frequency of 60 rpm, the internal power is the same, regardless of the external workload applied; whereas, at 100 rpm, internal power output is highest at low external power outputs. It was also observed that pulmonary \(\dot{V}O_2\) is higher at a contraction frequency of 100 than at 60 rpm at comparable power outputs, suggesting that mechanical efficiency is lower at the high contraction frequency.

Measurements of total power output at two different contraction frequencies revealed significantly greater levels of internal power at the higher contraction frequency, which has been previously observed during walking and running (11) as well as cycling (6, 8, 9, 13, 14). For example, internal power outputs of 20 and 62 W (16–26 and 50–72 W, respectively) were calculated for the entire leg during cycle exercise at pedal frequencies of 60 and 90 rpm, respectively, at external power outputs of 0, 60, 120, and 180 W (13). Furthermore, in the present study, the internal power at 60 rpm was independent of the external power output values are in Watts. Pred values were predicted from the regression equation between total and external power output at 100 rpm.

**Fig. 5.** Oxygen cost of exercise in relation to external power output (A) and total power output (B) during knee-extensor exercise at 60 and 100 rpm (means \(\pm\) SE, \(n = 12\)). *Difference between 60 and 100 rpm \((P < 0.05)\).
power output. In contrast, at 100 rpm, the internal power was lower at higher external power outputs (Table 1). Thus, at both contraction frequencies, the relative proportion of internal power to total power was lowered as total power increased, and there was an apparent convergence in total power output between 60 and 100 rpm as external power increased (Fig. 4). During cycle exercise at pedal frequencies of 30, 60, and 90 rpm, it was reported that internal power output remained constant as external power increased (13). However, there was considerable variation in internal power such that, during exercise at 90 rpm, the internal power in one subject was 84, 55, 66, and 68 W at external power outputs of 0, 60, 120, and 180 W, respectively (13).

In the second part of the study, the total power produced during exercise at the two contraction frequencies was matched. This was done by predicting the external power necessary at 100 rpm to produce the same total power as that at 60 rpm. The data in Table 2 show that this could be achieved with reasonable accuracy. At the same time, however, considerable intersubject variation was observed in the magnitude of internal power output at 100 rpm. This resulted in a large variation in the external power required to match the total power produced at the slow contraction frequency; the reasons for this are unclear. Internal power is affected by the anatomical dimensions of the lower limb as well as the time history of angular acceleration and velocity. However, there was no relationship between lower limb length or estimated limb mass and the level of internal power produced. It is likely that other factors, such as intersubject differences in limb acceleration and velocity, must have influenced the internal power output. Nevertheless, the total power output was the same at 60 and 100 rpm and allowed a direct comparison of the energy turnover during exercise at these two contraction frequencies. Pulmonary VO₂ was greater at 100 than at 60 rpm, suggesting that muscle mechanical efficiency was lower at 100 rpm. It should be recognized that pulmonary VO₂ does not necessarily represent the energy turnover by the contracting muscles. The method developed in the present study can, however, be used to relate the work performed at different contraction frequencies to muscle VO₂, which is accurately determined as the product of thigh blood flow and femoral arterial and venous blood O₂ difference. Thus true muscle mechanical efficiency can be determined in vivo.

In summary, a novel method has been developed in which the total mechanical power output can be quantified during dynamic exercise of a single, well-defined muscle group in humans in vivo. It is concluded that the internal power is independent of external power at a low contraction frequency, whereas, at a high contraction frequency, internal power decreases as the external workload increases. The model presented allows total power output to be matched, which will enable an accurate evaluation of the physiological responses during exercise at different contraction frequencies. The present data indicate that the muscle energy cost is higher at a frequency of 100 rpm compared with at a frequency of 60 rpm.

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