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Peripheral and Central Inputs to the Effort Sense during Cycling Exercise*

Enzo Cafarelli

John B. Pierce Foundation Laboratory and
Department of Epidemiology and Public Health, Yale University School of Medicine,
290 Congress Avenue, New Haven, Connecticut 06519, USA

Summary. The relationships between some physical and physiological events, and perceived effort were studied at several equivalent work outputs ($\dot{W}$) at two pedalling rates (30 and 60 rev · min^{-1}). Subjects judged effort throughout a 4 min exercise bout. After 4 min *at any $\dot{W}$ it was always more effortful to pedal at 30 rev · min^{-1}* even though there were no differences in $\dot{V}_E$, $\dot{V}O_2$, or integrated electromyography per minute (IEMG · min^{-1}) between pedalling rates. Effort was related to $\dot{V}O_2$ and IEMG · min^{-1} but it was more effortful to pedal at 30 rev · min^{-1}. Effort was also related to pedal resistance and IEMG of single contractions but was influenced by pedalling rate after 4 min of exercise. *At any resistance it was more effortful to pedal at 60 rev · min^{-1}*, however, when effort was plotted as a function of resistance after 15 s, there was virtually no effect of pedalling rate. The rate effect grows with time from the onset of exercise and appears to be related to the central signal to the effort sense. The interaction of peripheral and central signals suggests a model of the effort sense during exercise.

Key words: Perceived effort – Integrated electromyography – Magnitude estimation – Model of the effort sense.

Introduction

The signals that determine the sensation of effort during muscular exercise have been considered to be of peripheral and central origin [12, 19]. The prevailing hypothesis states that interoceptive drive for effort originates peripherally in the working limbs and is also linked in some fashion to central physiological processes [12]; these processes primarily being the circulatory and respiratory responses to increased metabolic demand for oxygen from working muscle. Although the concept of two inputs may seem intuitively reasonable there is little experimental evidence that

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demonstrates a central signal for effort. Indeed, a few controlled studies have demonstrated that differences in effort between equivalent tasks are not attributable to the so called central processes of heart rate or ventilation [8, 12, 17, 20, 25]. Although these results do not rule out a central input they have been interpreted by some to mean that the primary signal to the effort sense originates peripherally in the working muscles [8, 22].

Isolating its inputs would aid considerably in understanding how the effort sense works. One approach to this problem has been to vary pedalling rates at equivalent work outputs ($\dot{W}$) on a mechanically braked bicycle ergometer. Under these conditions the sensation of effort is greater at slow than at faster rates even though the work output is the same [14, 22, 26]. Since central processes such as heart rate, oxygen uptake, and pulmonary ventilation are relatively constant in each case [2], these results also support the notion of a strong peripheral signal to the sense of effort.

The higher levels of subjective effort that occur under conditions of heavy external resistance and slow pedal rate suggest that the sensations are proportional to the intensity of individual contractions during rhythmic exercise. The present experiment sought to examine this hypothesis. If effort depends primarily on peripheral events, the intense sensations accompanying high intensity-low frequency contractions can be explained by external (pedal) resistance. Moreover, since integrated electromyography (IEMG) is essentially linear when related to force of dynamic contractions over the range studied [4, 13], it should also be possible to relate effort sensations to a peripheral physiological process associated with muscular contraction.

Methods

Ten paid volunteers (7 male, 3 female) between the ages of 22–32 (24.4 ± 3.5) participated in the study. On the average, the subjects were 173 ± 6 cm tall, weighed 69 ± 10 kg and had a $\dot{V}O_2$max of 3.11 ± 0.47 $l \cdot min^{-1}$. Each served on 3–4 occasions for approximately 1.5–2.0 h separated by 24 h periods. The initial sessions were to familiarize the subjects with the equipment and procedures used in the experiment and to determine the maximal oxygen uptake ($\dot{V}O_2$max) during cycling exercise.

The protocol consisted of 4 min exercise bouts on a mechanically braked cycle ergometer at levels that required approximately 35, 50, 65 and 80% $\dot{V}O_2$max. Each level was performed at 30 and 60 $rev \cdot min^{-1}$. Four to five bouts in irregular order were performed in a single day. Rest periods varied with the severity of the preceeding bout so that those following light bouts were 4–5 min long and those following more intense activity lasted 10 min or longer. The mean rest period was approximately 7 min long. The subjects were free to extend the rest periods until they felt completely recovered.

Perceived effort was measured by the method of magnitude estimation [31]. Briefly, the method of magnitude estimation, a direct ratio scaling technique, requires the subject to first assign any appropriate number to the first level of sensation and to assign other numbers to subsequent sensations in proportion to their perceived level. For example, a sensation that feels twice as intense as the original is assigned a number twice as large. A sensation that feels half as intense as the original is assigned a number that is half as large, and so on. Magnitude estimations of effort were made at 15 s intervals during the first minute of exercise and every minute thereafter for the remainder of the 4 min bout. All estimations of effort for one day's testing were relative to the first (15 s) estimation of the first bout.

In order to tie together the estimations made on different days, the following procedure was adopted. On the second day one bout from the first series was presented for 1 min and the four estimations (15, 30, 45 and 60 s) from this period were summed. The ratio between this sum and the sum of the corresponding estimations from the first day was used to adjust all of the second days' estimations to the same number scale. Similar steps were taken to normalize the data across subjects. In either case the procedure does not affect mean values but corrects for the variability due to the choice of initial estimates from day to day and from subject to subject. The magnitude estimation data of all subjects are expressed as geometric means $(\bar{X}_g)^1$. This measure of central tendency is preferred because the distribution of magnitude estimates is typically positively skewed but the distribution of its logarithms is essentially normal [32].

The relation between perceived effort and work output or force level is a function of the general form:

$$E = E_0 + aS^{\beta} \qquad (1)$$

where E is effort, E_0 is effort of unloaded pedalling, a is a constant related to the choice of units, and S is the stimulus. Unloaded pedalling refers to operating the ergometer with no resistance applied to the flywheel. The E_0 term accounts for the internal friction of the subject and ergometer [7]. When E—E_0 is plotted versus S in logarithmic coordinates, the function is linear with a slope equal to β [5, 6, 29]. Studies of effort during static and dynamic muscular contraction reveal exponents of about 1.2—1.8 [5, 6, 7, 28, 29]. Thus when these functions are plotted in linear coordinates they are positively accelerated, but in logarithmic coordinates they may be fit with a straight line having a slope greater than 1.0. This means that at progressively higher levels small increases in $\dot{W}$ produce disproportionately larger increases in effort.

Measurements of $\dot{V}O_2$max were obtained by a discontinuous method [24]. A short warm-up and at least 10 min of recovery were followed by 6 min of exercise at roughly 50% $\dot{V}O_2$max. Heart rate and oxygen uptake ($\dot{V}O_2$) were measured during the last 2 min of this period. From these data Åstrand's [1] nomogram was used to predict $\dot{V}O_2$max. Starting with exercise intensities that would produce about 90% $\dot{V}O_2$max predicted, 3 min periods of exercise were interspersed with 10—15 min rest periods. The work output was increased 150 kgm · min^{-1} with each 3 min bout until there was no accompanying increase measured in $\dot{V}O_2$. Throughout the procedure the subject's feet rested on platforms fitted to the ergometer pedals and secured with straps so that the axis of rotation of the ankle was directly over the axis of rotation of the pedals, thus isolating the quadriceps during pedalling. The peak value was assumed to represent the $\dot{V}O_2$max for this mode of exercise.

The surface electromyogram (EMG) was obtained from silver/silver chloride cup electrodes placed on the lateral border of the vastus lateralis at the level of its belly and at its insertion on the patellar tendon. Resistance of the electrodes, leads, and leg was always below 10 kΩ and usually less than 5 kΩ. The raw EMG signal was fed into a wide band AC preamplifier and then fully rectified and integrated with a time constant of 0.02 s. The vastus lateralis EMG was assumed to provide a representative sample of muscle fiber activity of the entire quadriceps group [4, 13].

Area under the integrated EMG (IEMG) curves was determined with a compensating polar planimeter. Three IEMG curves were taken from the last 30 s period of exercise and measured 3 times with the planimeter. The mean of all nine measurements was taken as the average IEMG for that bout. The duration of individual contractions was measured directly from the primary EMG trace. Reliability of IEMG measurements was checked by repeat runs for one bout per subject per day. The correlation between duplicates ($r = 0.93$; $t = 0.186$ not significant) agrees with a similar measure reported by Bigland-Ritchie and Woods [4].

The PO_2 and volume of expired air was determined by an open circuit technique using a polarographic O_2 electrode (Yellow Springs Instruments) and a dry gas meter (American Gas). Oxygen uptake was calculated as $F_{IO_2} - F_{EO_2} \times \dot{V}_E$. Assuming that R.Q. ranged between 0.80 and 1.10, the expected error is about ± 4.0% [11].

The data were analyzed statistically with a two-way analysis of analysis of variance (work output × pedal rate) for repeated measures. Magnitude estimates of effort were converted to logarithms before computation. Scheffe's method for individual comparisons was used where there were significant main effects. The level of significance for all analyses was preset at $p < 0.05$. Regression lines were fitted to mean values by a least squares technique.

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[1] $\bar{X}_g = [(x_1)(x_2)(x_3)\ldots(x_n)]^{1/n}$ = the antilog of the arithmetic mean of the logarithms of n estimations

Results

Predictably, there was no significant main effect of pedalling rate on $\dot{V}O_2$ and $\dot{V}_E$ responses to a given $\dot{W}$.

Perceived effort after 4 min of exercise as a function of $\dot{W}$ at 30 and 60 rev · min^{-1} is shown in Figure 1A. At both pedalling rates the relationship is positively accelerated and can be fitted with a power function (Eq. 1). In logarithmic coordinates (Fig. 1A, inset) both functions have been fitted with straight lines with a slope (exponent) of 1.83 at 30 rev · min^{-1} and 1.45 at 60 rev · min^{-1}. These values agree with exponents found in previous studies for dynamic as well as static exercise [7, 28]. The finding that each $\dot{W}$ was significantly more effortful at 30 rev · min^{-1} agrees

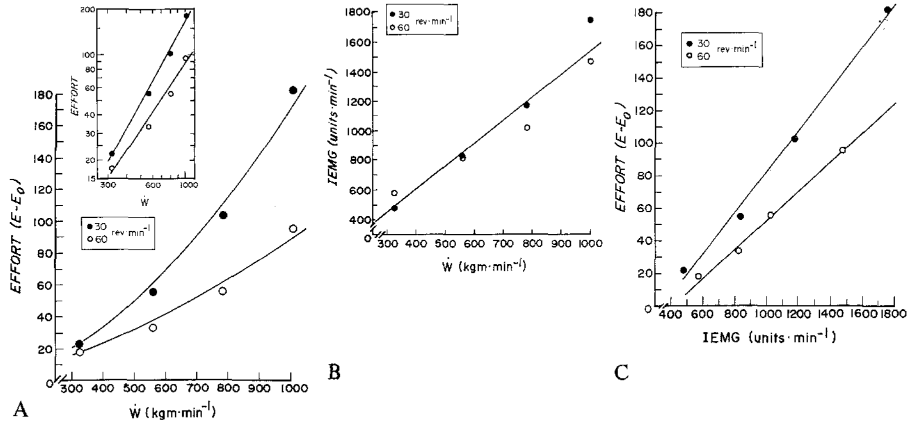

Fig. 1. The relationship between effort, work output and IEMG per unit time. **A** Effort as a function of work output. E_0 is effort during unloaded pedalling. Inset shows same data plotted in logarithmic coordinates. **B** IEMG · min^{-1} is a linear function of $\dot{W}$ independent of pedalling rate, $r = 0.96$. **C** Effort as a function of IEMG · min^{-1}. IEMG units are arbitrary

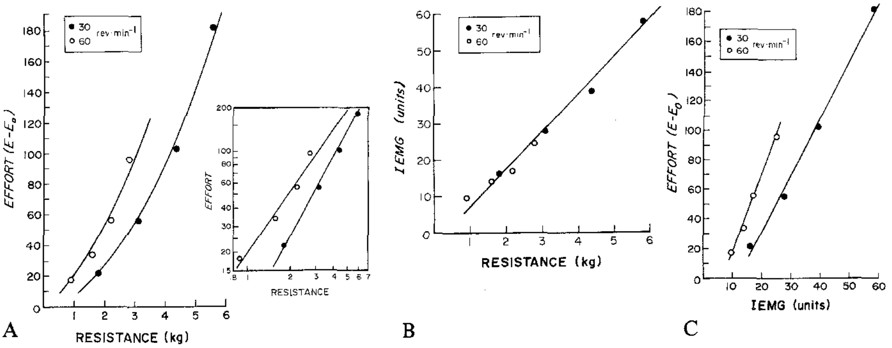

Fig. 2. The relationship between effort, pedal resistance and IEMG for individual contractions. **A** Effort as a function of resistance. Inset is same data plotted in logarithmic coordinates. **B** IEMG is a linear function of resistance independent of pedalling rate. **C** Effort as a function of IEMG

with existing data [14, 22, 26] but the increasing differences in effort between the two tasks as the level of $\dot{W}$ grows has not been described.

To approximate muscle fiber activity per unit time, the product of the IEMG for individual contractions and the pedalling rate is expressed as IEMG · $\min^{-1}$. Figure 1B shows the relationship between IEMG · $\min^{-1}$ and $\dot{W}$. None of the differences between points at a given $\dot{W}$ was significant, indicating no effect of pedalling rate on muscle fiber activity per unit time.

Figure 1C shows effort as a function of IEMG · $\min^{-1}$ for both pedalling rates. It is similar to Figure 1A in that for any level of IEMG · $\min^{-1}$ it is more effortful to pedal at 30 rev · $\min^{-1}$. Assuming that there is a link between muscle fiber activity and the peripheral signal for effort sensations, it may not be the amount of fiber activity per unit time but rather the amount of fiber activity per contraction that drives the sense of effort.

In order to explore this assumption further the effort data are replotted using pedal resistance and IEMG of individual contractions as the independent variables (Fig. 2). The relation between the physical stimulus, resistance, and the subjective response, effort, is once again described by power functions. The functions have the same exponents as the effort-$\dot{W}$ relation, but their relative positions have reversed (Fig. 2A and inset). For any pedal resistance it is more effortful to pedal at 60 rev · $\min^{-1}$ than at 30 rev · $\min^{-1}$. The fact that the data in Figure 2A fit two separate functions indicates that the difference in effort between pedalling rates cannot be completely accounted for by the reciprocal changes in resistance.

The linear relationship obtained between IEMG and force is shown in Figure 2B and effort as a function of IEMG at two pedalling rates is shown in Figure 2C. The rate of contraction (pedalling rate) has no effect on the IEMG of individual contractions but it does have a significant effect on the relation between effort and IEMG. For any level of EMG activity it is more effortful to pedal at a greater frequency.

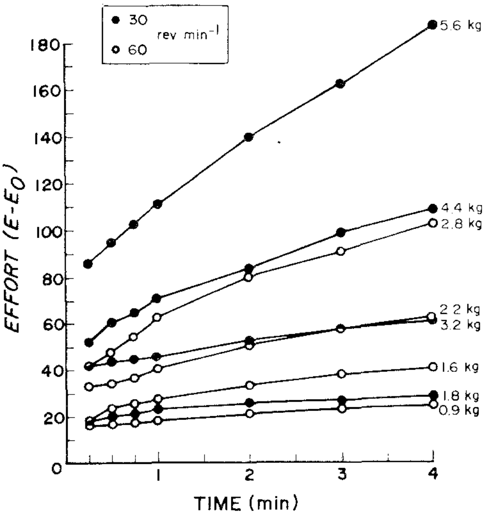

Fig. 3. Effort as a function of time. Pedal resistance, in kg, is the parameter

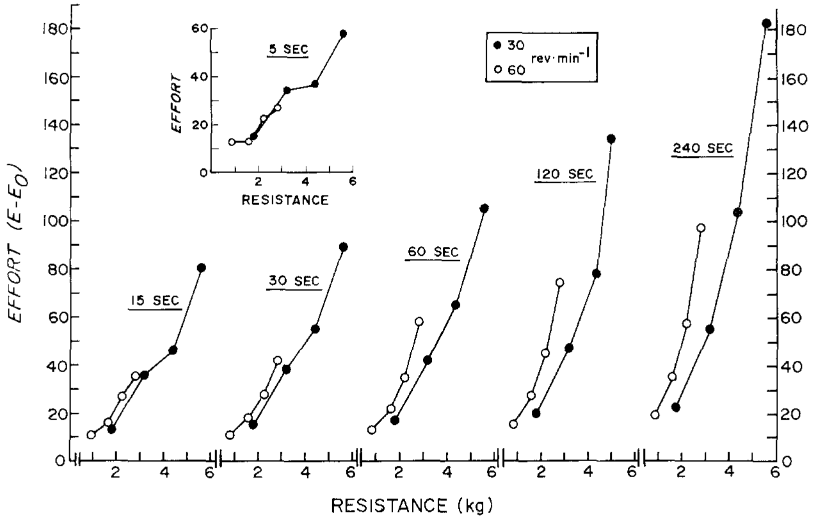

Fig. 4. Effort as a function of pedal resistance at selected intervals during 4 min bouts. Data at 5 s are extrapolations from a logarithmic plot of Figure 3

All of the data in Figure 2 demonstrate the strong relationship between the sense of effort and external peripheral (resistance) and internal peripheral (IEMG) events. However, the inability of these data to account for the difference in effort at two rates of contraction illustrates that the peripheral signal is not the only input to the overall sensation of effort.

Figure 3 shows that the growth of effort with time depends on pedal resistance and the work output. After 4 min of exercise the relation between effort and resistance is influenced by the pedalling rate. Under the assumption that at some point near the onset of exercise effort is entirely dependent on resistance, vertical cuts were made along the horizontal axis of Figure 3. Effort was then replotted as a function of pedal resistance (Fig. 4).

Going back in time from the terminal values at 4 min (240 s), the effort-resistance functions for the two pedalling rates show a marked tendency to grow closer and closer together until at 15 s they are essentially merged. In order to examine this tendency still further, the data in Figure 3 were fitted with power functions, plotted in log-log coordinates and extrapolated back to 5 s. The resulting effort-resistance functions shown in Figure 4 demonstrate the absence of a rate effect at the beginning of exercise.

Discussion

The results of the present study agree with Ekblom and Goldbarg [12] and Noble et al. [19] that the sensation of effort is a function of at least two inputs. One input is closely related to the muscular tension developed to overcome an external resistance (Fig. 2A). Previous reports that it is more effortful to maintain a given $\dot{W}$ when pedal resistance is high [14, 22, 26] tend to support this view.

Effort is also related to the IEMG of individual contractions. Although IEMG is an efferent signal it is a linear function of muscle tension [3, 4, 13] and is a reasonable physiological estimate of the peripheral activity. Despite the relationships between effort, resistance, and IEMG it is apparent that they cannot account for the large differences in effort between pedalling rates. If the peripheral signal were the only input to the effort sense the data in Figure 2A and C would have fit a single curve. Moreover, the dynamic relationship between effort and resistance over time depicted in Figure 4 suggests that the two signals are somewhat independent of one another.

Additional support for the hypothesis that there are separate signals derives from an experiment where subjects made separate magnitude estimates of both peripheral and central effort sensations during cycling with one and two limbs [7]. At comparable work outputs, peripheral effort sensations grew faster as a function of time for one-limb compared to two limb exercise. This implies that the peripheral signal is stronger when one limb does the same work as two. Central sensations, however, appeared to grow at about the same rate in each case suggesting that the central input was constant at the same work output.

Figure 5 is a conceptual model of the factors which contribute to the sensation of effort during cycle ergometer exercise. Overcoming an external resistance requires muscle tension that produces a sensation of effort. During rhythmic exercise, both the rate and force of contraction function additively to determine the metabolic demand for oxygen of muscle. With time the central component appears to also feed the sense of effort. Since the growth of effort with IEMG activity is dependent on the work output, any level of IEMG activity is more effortful when oxygen consumption is greater. To a first approximation then, effort appears to be a complex multiplicative function of central and peripheral signals.

Figure 5 also illustrates that the peripheral and central signals are not coincident in time. For example, a single phasic contraction produces a sensation of effort even though the central response is essentially nonexistent [29]. Studies of the response time of oxygen uptake, heart rate, and ventilation from the onset of exercise show that their exponential rise has a half time of about 30 s even at high work outputs [16]. The data of the present study show that at about 30 s the effort-resistance functions for the two pedalling rates are clearly separate. This would suggest that before the first 30 s of exercise, the sensation of effort is entirely peripheral, and depends on the force of contraction. Thereafter the difference in effort at a given force of contraction appears to depend on the central signal.

Central and peripheral inputs also depend on both the absolute level of resistance and work output. At approximately 30% $\dot{V}O_2$max (320 kgm · min^{-1}) there is

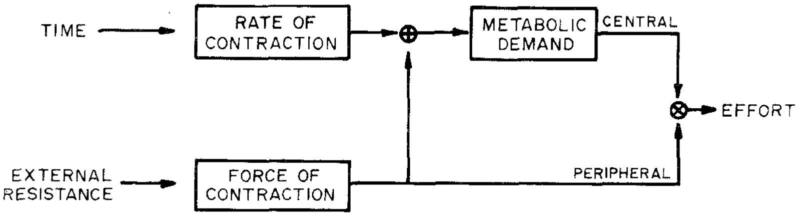

Fig. 5. Block diagram of the components that may drive the effort sense

little difference in effort between the two contraction rates despite a 2 : 1 difference in pedal resistance. However, when $\dot{W}$ requires about 85% $\dot{V}O_2$max the slower rate of contraction is nearly 90% more effortful.

When muscles are engaged in sustained static contractions the buildup of fatigue is more rapid than during repeated static contractions of the same magnitude [10, 30]. Using a constant effort technique, Pandolf and Cain [21] showed that the exponential decline in tension to maintain the same sense of effort is more rapid during isometric handgrip than during isotonic cycling. In experiments where the blood supply to working muscles was occluded at low contraction tensions, the accumulation of fatigue (or inability to sustain a constant force) occurs in a fashion similar to when the contractile force is much greater [9, 23, 27, 30]. All of these data suggest that the more a contraction or series of contractions tends to impede blood flow, the greater the level of effort. One could speculate that the peripheral signal to the effort sense is influenced by the reduction of available substrate and accumulation of metabolites that would result from an inadequate blood flow to working muscle.

The specific nature of the central signal to the effort sense is more difficult to assess. Although there is some correlational evidence to suggest that minute ventilation could be a primary central signal [15, 18], the present study and others [8, 20] tend to rule out this hypothesis. Similarly, the sense effort has been shown to be insensitive to experimental manipulations of heart rate during exercise [12, 20, 25]. Nevertheless, the present study illustrates that the central signal to overall effort is related to the work output and is, therefore, probably reflected in the oxygen uptake. In accordance with the proposed model in Figure 5, the primary signal for effort relates to the force of contraction which, along with the rate of contraction, determines the metabolic demand. Thus, the central input appears to function as an amplifier that potentiates the peripheral signal in proportion to the metabolic demand for oxygen.

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