Femoral artery inflow in relation to external and total work rate at different knee extensor contraction rates

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Osada, Takuya, and Göran Rådegran. Femoral artery inflow in relation to external and total work rate at different knee extensor contraction rates. J Appl Physiol 92: 1325–1330, 2002; 10.1152/japplphysiol.00848.2001.—Whether limb blood flow is directly regulated to match the work rate, independent of the rate of contraction, remains elusive. This study therefore investigated the relationship between femoral arterial blood flow (FABF; Doppler ultrasound) and “external” (applied load) as well as “total” [external + “internal” (potential and kinetic energy changes of the moving lower leg)] work rate, during steady-state one-legged, dynamic, knee extensor exercise (1L-KEE) in the sitting position at different contraction rates. Ten subjects performed 1L-KEE at 30, 60, and 90 contractions/min (cpm) 1 at constant resistive loads of 0.2 and 0.5 kg inducing incremental external work rates (study I) and 2 at different relative resistive loads inducing constant external work rates of 9 and 18 W (study II). Moreover, 3 six subjects performed 1L-KEE at 60 and 100 cpm at incremental total work rates of 40, 50, 60, and 70 W (study III). In study I, FABF increased (P < 0.001) with increasing contraction frequency and external work rate, for each resistive load. In study II, FABF increased (P < 0.001) with increasing contraction frequency for each constant external work rate. Of major importance in study III, however, was that FABF, although increasing linearly with the total work rate, was not different (P = not significant) between contraction rates, at the total work rates of 40, 50, 60, and 70 W, respectively. Furthermore, FABF correlated linearly and positively with both the external and total work rate for each contraction frequency. In conclusion, the findings support the concept that leg blood flow during 1L-KEE in a normal knee extensor ergometer is matched directly in relation to the total work rate and metabolic activity, irrespective of the contraction frequency. The rate of contraction seems erroneously to influence the results only when it is related to the external work rate without taking into account the internal work component.

exercise hyperemia; human; knee extensor exercise; limb blood flow; Doppler ultrasound

EXERCISE-INDUCED SKELETAL muscle hyperemia is hypothesized to be governed by an interplay between local and central vascular control mechanisms (7, 24). The previously observed linear relationship between leg blood flow and work intensity during one-legged, dynamic, knee extensor exercise (1L-KEE) at 60 contractions/min (cpm), by using thermodilution venous outflow (3) and Doppler ultrasound arterial inflow measurements (20), furthermore implies a close coupling between the metabolic activity and the steady-state blood flow response. The arterial inflow is, however, also known to be closely related to the muscle contraction-relaxation duty cycles; there is a mechanical hindrance to femoral arterial blood flow (FABF) at the high intramuscular pressures occurring during the contraction phase and with an unimpeded FABF at the low intramuscular pressures occurring during the relaxation phase (22). Muscle mechanical factors may accordingly both impede and promote skeletal muscle blood flow during dynamic exercise (22, 25, 27, 28). It has furthermore been suggested that the vasodilatation predominantly increases muscle blood flow when the muscular contraction force is lower than that which may restrict arterial inflow (6, 23). The vasodilatation may thus be impeded by the intramuscular pressure at higher muscle contraction forces (4, 23). The precise influence on limb blood flow by muscle mechanical factors obtained by altering the contraction frequency at different work rates is, however, not clear. This is, furthermore, of great importance because a modification of the rate of contraction could be a means whereby 1) leg blood flow and muscle perfusion could be optimized and 2) the influence of muscle metabolic vs. mechanical factors in the control of exercise hyperemia could be addressed. Thus whether leg blood flow is influenced by altering the contraction frequency is yet to be determined.

An increase of the contraction frequency during plantar flexion exercise has previously been suggested to impede skeletal muscle perfusion, on the basis of plethysmography measurements performed postcontraction (15). An extensive investigation on the relationship between leg blood flow and work rate during knee extensor exercise at different contraction frequencies was recently also performed (12). Hoeltj and co-workers (12), however, used a modified intermittent isometric knee extensor equipment, allowing a limb...

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movement of only ~25% of that of the normal dynamic knee extensor ergometer (2, 3, 12). According to Hoeltling et al., the muscle tension developed on their ergometer was up to four times greater than on the normal knee extensor ergometer. Moreover, in contrast to previous findings on the normal knee extensor ergometer (3, 20), the mean blood flow response was found not to be linearly related to the power output but rather was attenuated at higher intensities (12). FABF furthermore decreased with increasing contraction frequency (12). Moreover, in the intermittent isometric knee extensor exercise model of Hoeltling et al., the relaxation duration decreased with increasing contraction frequency. Their findings thus reflect the influence of altering the duration of the relaxation phase, rather than the effect of altering the contraction rate per se. The experiments were also performed with the subjects in the supine position, which may influence the magnitude of the blood flow response by altering venous return, stroke volume, and cardiac output (11, 16, 18, 19). Thus it is not readily apparent how the results of Hoeltling et al. relate to the majority of prior experiments performed on the normal dynamic knee extensor ergometer.

The purpose of the present study was, therefore, to test the validity of the traditional hypothesis that steady-state leg blood flow during 1L-KEE is regulated directly to match the exercise intensity independent of the contraction frequency. This was accomplished by having the subjects perform 1L-KEE at different contraction frequencies and relating the blood flow response to both the “total” and “external” work rates. Both types of work rates were studied because 1) the external work rate is the intensity traditionally referred to when the normal dynamic knee extensor ergometer is used and 2) as the total work rate corresponds to the actual work performed, accounting for the internal work component.

METHODS

Study Population

Ten healthy, male volunteers, with a mean ± SE (range) age of 31.6 ± 1.5 yr (26–36 yr), height of 182.2 ± 1.8 cm (177–190 cm), and body weight of 75.7 ± 3.2 kg (61–90 kg), participated in the experiments. The subjects’ engagement in exercise varied from daily activities to regular endurance training. They had no prior history of cardiovascular disease, hypertension, or anemia. The subjects who volunteered were informed about the experimental procedures, its potential risks, and discomfort, and they were told that they could withdraw at any time without any consequences. They participated after providing signed informed consent. The experiments were carried out with approval from the Ethical Committee’s of Copenhagen and Fredriksberg (KF-01-013/96).

Experimental Design

Before the experiments, all subjects were familiarized with the 1L-KEE model (2, 3) by training at 30, 60, and 90 cpm until they were comfortable and could fully relax the hamstring muscles so that the work was performed solely by the knee extensor muscle group. The upper body and both thighs were fixed by belts attached to the seat. The blood flow measurements were performed at steady state after 3–5 min of exercise (20). Three different experimental protocols (studies I–III) were performed on different days. It was verified that sufficient amount of time was allowed between the exercise bouts to allow blood flow to return to rest control level. The control of contraction rate was accomplished by kicking in relation to the sound of a visible metronome. The contraction rate was furthermore visualized in real time on a monitor. In the present study, by using the dynamic knee extensor exercise model, the muscle contraction and relaxation duration composed a “relatively” constant percentage of the full contraction and relaxation cycle duration (e.g., knee extension cycle revolution), independent of the contraction frequency. To differentiate the influence on FABF of work rate and contraction frequency, the study design below was applied (studies I–III). For the reason stated in the last paragraph of the introduction, both the external and total work rates were studied.

Study I: FABF in relation to increasing contraction frequency and external work rate at constant resistive loads. 1L-KEE was performed by 10 subjects in the sitting position at 30, 60, and 90 cpm at constant resistive loads of 0.2 and 0.5 kg. This corresponded for each load, at the three contraction frequencies, to external work rates of 6, 12, and 18 W as well as of 15, 30 and 45 W, respectively.

Study II: FABF in relation to contraction frequency at fixed external work rates and different relative resistive loads. 1L-KEE was performed by 10 subjects in the sitting position at 30, 60, and 90 cpm at the specific external work rates of 9 and 18 W. This corresponded at each work rate to applied loads for the different contraction frequencies of 0.3, 0.15, and 0.1 kg and of 0.6, 0.3, and 0.2 kg, respectively.

Study III: FABF in relation to total work rate at increasing contraction frequency. 1L-KEE was performed by six subjects in the sitting position at 60 and 100 cpm at total work rates of 40, 50, 60, and 70 W, respectively. This was accomplished by applying loads of 0.379, 0.543, 0.706, 0.869, and 1.033 kg, respectively, at 60 cpm and of 0.019, 0.149, 0.279, 0.409, and 0.539 kg, respectively, at 100 cpm.

Calculation of external and total work rate. The total work rate was calculated by using the formula previously established and validated by Ferguson et al. (9), defined as follows: total work rate (W) = 16.8 + 1.02 × external work rate for 60 cpm; and total work rate (W) = 38.5 + 0.77 × external work rate for 100 cpm. The total work rate corresponds as follows: the external (applied load) + the internal (the potential and kinetic energy changes of the moving lower leg) work rate. The external work rate was calculated according to the dynamic knee extensor ergometer model (2, 3) and defined as follows: external work rate (W) = [contraction rate (cpm)/60 s] × [distance of a knee extensor revolution (6 m)] × [load weight (kg) × 9.81 (m/s²)].

FABF Measurements

The procedure of FABF measurements has previously been validated and shown to produce accurate absolute values at rest and during exercise (20). The equipment used was a Doppler ultrasound (model CFM 800, Vingmed Sound, Horten, Norway) equipped with an annular phased array transducer (Vingmed Sound, Horten, Norway) probe (11.5 mm diameter), operating at an imaging frequency of 7.5 MHz and variable Doppler frequencies of 4.0–6.0 MHz (high-
pulsed repetition frequency mode 4–36 kHz). The site for vessel diameter determination and blood velocity measurements in the common femoral artery, which was always the same, was distal to the inguinal ligament but above the bifurcation into the superficial and profunda femoral branch. The position was chosen to minimize turbulence from the bifurcation and influence of blood flow from the inguinal region; also, the arterial diameter was unaffected by the contraction and relaxation per se at this site proximal to the muscle location. The blood velocity measurements were performed with the probe in as low insonation angle as physically possible and always below 60° (10). The insonation angle was continuously measured and corrected for in the subsequent blood flow calculation. The systolic and diastolic diameters in the femoral artery were measured in relation to the electrocardiogram on the Doppler ultrasound monitor. The mean vessel diameter was calculated in relation to the duration of the blood pressure curve according to the following formula: diameter = [(systolic diameter × s) + (diastolic diameter × s)]/2. The diameter measurements were obtained under perpendicular insonation. Special care was taken to ensure that the probe position was stable, that the insonation angle did not vary, and that the sample volume was positioned in the center of the vessel and adjusted to cover the width of the diameter. Contributions to the signal by turbulence occurring at the vascular wall were reduced with a low-frequency rejection filter. FABF was calculated by multiplying the cross-sectional area of the femoral artery [area = π × (diameter)²] with the angle-corrected, time and space-averaged, and amplitude (signal intensity)-weighted mean blood velocity (Vmean, m/s) (20); e.g., FABF = Vmean × area × 6 × 10⁴ (l/min), where the constant 6 × 10⁴ is the conversion factor from meters per second to liters per minute. Furthermore, before the experiments the intraobserver variability was determined by 25 repeated measurements in one subject. At rest, the coefficients of variation determined for the diameter and blood flow were 2.7 ± 0.3 and 3.4 ± 0.5%, respectively.

Statistical Analysis

Parametric statistics was used for data analysis (i.e., multiple analysis of variance for repeated measures and Tukey honestly significant difference post hoc tests when comparing more than 2 groups over time, and paired t-test when comparing 2 groups only). A P value <0.05 was considered as statistically significant. P = NS indicates not statistically significant. The values are mean ± SE.

RESULTS

The mean femoral arterial diameter and FABF at rest were 9.6 ± 0.1 mm and 337 ± 33 ml/min, respectively. In study I, FABF increased (P < 0.001) as a function of increasing contraction frequency (30, 60, and 90 cpm) and external work rate for each of the constant resistive loads of 0.2 and 0.5 kg. FABF at the external work rate of 15 W (30 cpm) was not different (P = not significant (NS) compared with 12 W (60 cpm)). FABF at the external work rate of 30 W (60 cpm) was not different (P = NS) compared with 18 W (90 cpm). Significantly different (P < 0.05) at 90 than at 60 and 30 cpm, as well as at 60 than at 30 cpm (Fig. 2). FABF at the external work rate of 9 W (90 cpm) was also higher (P < 0.001) than at 18 W (30 and 60 cpm), related to the higher rate of contraction.

A positive linear correlation (P = 0.08, r = 0.92 at 30 cpm; P < 0.05, r = 0.98 at 60 cpm; and P < 0.05, r = 0.97 at 90 cpm) was observed between FABF and the specific calculated external work rate during 1L-KEE at each contraction frequency (Fig. 3). The regression lines at the three different contraction frequencies, reflect the mean value of FABF for a given value of external work rate: i.e., FABF (l/min) = 1.127 + 0.059 × external work rate at 30 cpm; FABF (l/min) = 1.115 + 0.083 × external work rate at 60 cpm; and FABF (l/min) = 2.621 + 0.058 × external work rate at 90 cpm.

Of major importance, however, was the observation that FABF, although increasing linearly with the total work rate, was not different (P = NS) between contraction rates at the corresponding total work rates of 40, 50, 60, and 70 W (Fig. 4). Similar (P = NS) positive linear correlations (P < 0.01, r = 0.998 at 60 cpm; and P < 0.001, r = 0.999 at 100 cpm) were observed between FABF and the total work rates during 1L-KEE at the incremental contraction frequencies, where FABF (l/min) = 0.353 + 0.059 × total work rate (W) at
to the level of activity and metabolic demand. The present study furthermore also supports the idea that leg blood flow during incremental, steady-state, 1L-KEE is linearly related to the external work rate, independent of contraction frequency (2, 3, 20). This advances previous knowledge such that the linear relationship between leg blood flow and external work rate is true not only for 60 cpm but also for 30 and 90 cpm.

In addition, the findings that FABF increased as a function of contraction frequency for the constant resistive loads of 0.2 and 0.5 kg is in agreement with the finding that the external work rate increased in parallel (Fig. 1). Of further interest, however, is that FABF at the external work rate of 15 W (30 cpm) was the same as at 12 W (60 cpm) and that FABF at the external work rate of 30 W (60 cpm) was the same as at 18 W (90 cpm). Furthermore, FABF at the external work rate of 18 W (90 cpm) was proportionally much higher than at 15 W (30 cpm). These results could be interpreted such that the contraction frequency may be an important means of optimizing muscle perfusion during dynamic exercise and thus uncouple the close relationship between the exercise intensity-related metabolic activity and the leg blood flow response. This concept was further supported by the observation that FABF was higher at 90 than at 60 and 30 cpm, as well as at 60 than at 30 cpm, at the different relative resistive loads rendering specific external work rates of 9 and 18 W (Fig. 2). It could thus be speculated that the relative importance of the contraction frequency and muscle mechanical factors for promoting and impeding
muscle blood flow, and for muscle vasodilatation to enhance blood flow, could vary at different contraction frequencies. Of major importance, however, is that the external work rate is not a proper measure of the full metabolic stress and activity, or the true work performed, because it does not include the internal work component related to the potential and kinetic energy changes inherent to the movement of the lower leg in the knee extensor ergometer. Our results, based on the model of Ferguson et al. (9), relating the leg blood flow response to the total work rate, instead demonstrate that blood flow indeed primarily is set by the total work rate and not the contraction frequency, verifying the very close relationship between the metabolic activity and the leg blood flow response to exercise. Thus the increase in blood flow related to the contraction frequency in the external work rate experiments seems rather to be a result of an increase in the internal work component than induced by the rate of contraction per se. Future work should specifically focus at directly measuring the internal work component to determine its dependence on the rate of contraction.

Our findings are thus different from those of 1) Kagaya (15), who, with plethysmography during plantar flexion exercise, found blood flow to be lower at higher muscular contraction frequencies, such as at 80 and 100 cpm compared with at 60 cpm. Our results “seemed” also to differ from those of 2) Hoelting et al. (12), who, with Doppler ultrasound during exercise in a modified intermittent knee extensor model, found lower mean blood flow values at 80 than at 40 cpm. These discrepancies may, however, as described below, be due to the different types of exercise models and range of movements applied, as well as to differences in the type and intensity of work performed. Moreover, the plethysmography measurements applied by Kagaya (15) only allow measurements in the recovery phase postcontraction, which is an erroneous estimate of the blood flow response during exercise (21). Furthermore, as stated by Hoelting et al. (12), the higher muscle tension developed in their model due to the limited range of movement may also influence the blood flow response and therefore interfere with a direct comparison to our results. Another possible explanation for the different findings is that, in the study of Hoelting et al., the duration of the contraction and the return of the limb to the start position remained unchanged at each of the different contraction frequencies. However, the duration of the relaxation phase decreased as the contraction frequency increased. The tension development and the rate of tension developed were furthermore unchanged at the different contraction frequencies, with only the period of relaxation being affected. The variation in the relaxation time may therefore specifically have influenced the blood flow between the contractions. Thus Hoelting et al. predominantly studied the effect of an alteration in the relaxation duration, rather than the effect of the contraction frequency per se. Furthermore, the exercise model of Hoelting et al. resembles the intermittent isometric (“static”) exercise model of Walloe and Wesche (28), rather than the 1L-KEE, where the duration of the relaxation and contraction phase is relatively constant irrespective of the contraction frequency. This further emphasizes that the present study provides new information on the influence of the rate of contraction and is therefore by nature complementary to the findings of Hoelting et al. (12), who primarily addressed the influence of the duration of the relaxation phase. Thus the finding of the studies do not conflict with each other. The nonlinearity of the mean blood flow response at higher power outputs in the study of Hoelting et al. are, however, puzzling but may be inherent to the higher muscle tension developed in their intermittent knee extensor exercise model. A direct comparison to the findings of Hoelting et al. may furthermore, however, be difficult to be performed be-

**Fig. 4.** FABF in relation to “total” work rate at 60 and 100 cpm (study III). A: no difference ($P = NS$) in FABF was observed between 60 and 100 cpm at each total work rate. B: FABF was linearly positively related at both 60 and 100 cpm to the total work rate (TWR) as calculated according to Ferguson et al. (9).
cause their experiments were conducted with the subjects in the supine position, which may influence blood flow both at rest (1, 11, 16) and during exercise (8, 13, 14, 17, 19). Alterations in the venous filling and return, peripheral resistance, and the hydrostatic pressure may accommodate such postural changes and potentially influence vascular tone in the resistance vessels and thereby muscle blood flow of the lower limb muscles. Future studies are encouraged where direct measurements of the external and internal work rate are performed in the supine and in the sitting position at a larger range of contraction rates to estimate the contribution of the internal work rate factor and the body position on the leg blood flow response and their relation to the contraction frequency.

In conclusion, the present study supports that leg blood flow during steady-state 1L-KEE is directly related to the intensity of the activity, such that limb blood flow is set to match the total work rate and metabolic demand, irrespective of the rate of contraction.

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