Published in final edited form as:

Am J Physiol Heart Circ Physiol. 2007 January ; 292(1): H215–H223. doi:10.1152/ajpheart.00681.2006.

Changes in regional blood volume and blood flow during static handgrip

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Abstract

Increased blood pressure (BP) and heart rate during exercise characterizes the exercise pressor reflex. When evoked by static handgrip, mechanoreceptors and metaboreceptors produce regional changes in blood volume and blood flow, which are incompletely characterized in humans. We studied 16 healthy subjects aged 20–27 yr using segmental impedance plethysmography validated against dye dilution and venous occlusion plethysmography to noninvasively measure changes in regional blood volumes and blood flows. Static handgrip while in supine position was performed for 2 min without postexercise ischemia. Measurements of heart rate and BP variability and coherence analyses were used to examine baroreflex-mediated autonomic effects. During handgrip exercise, systolic BP increased from 120 ± 10 to 148 ± 14 mmHg, whereas heart rate increased from 60 ± 8 to 82 ± 12 beats/min. Heart rate variability decreased, whereas BP variability increased, and transfer function amplitude was reduced from 18 ± 2 to 8 ± 2 ms/mmHg at low frequencies of ~ 0.1 Hz. This was associated with marked reduction of coherence between BP and heart rate (from 0.76 ± 0.10 to 0.26 ± 0.05) indicative of uncoupling of heart rate regulation by the baroreflex. Cardiac output increased by $\sim 18\%$ with a 4.5% increase in central blood volume and an 8.5% increase in total peripheral resistance, suggesting increased cardiac preload and contractility. Splanchnic blood volume decreased reciprocally with smaller decreases in pelvic and leg volumes, increased splanchnic, pelvic and calf peripheral resistance, and evidence for splanchnic venoconstriction. We conclude that the exercise pressor reflex is associated with reduced baroreflex cardiovagal regulation and driven by increased cardiac output related to enhanced preload, cardiac contractility, and splanchnic blood mobilization.

Keywords

exercise pressor reflex; mechanoreflex; metaboreflex

Static voluntary isometric contraction evokes both central command (9), which is independent of afferents arising from exercising muscle, and the exercise pressor reflex, which is strictly dependent on sensory afferents from exercising muscle (29). The exercise

pressor reflex results in vagal withdrawal (14) and sympathetic activation (26), which together increases heart rate (HR) and blood pressure during exercise. The pressor reflex is composed of two parts comprising the muscle mechanoreflex (22) and the muscle metaboreflex, which is evoked as muscle blood flow fails to keep up with metabolic demands (6, 21, 29). Since Alam and Smirk's landmark work (1) the most marked pressor responses are believed to develop during the phase of muscle ischemia and depend on the amount of muscle involved (10), on age (31), on gender (30), and on the magnitude of muscle tension (42) produced by the activation of group III and IV muscle afferents (22, 23).

Effects of the exercise pressor reflex on regional hemodynamics are only partially understood. Thus in experiments on conscious dogs, O'Leary's laboratory (45) studied a graded metaboreflex and demonstrated increased cardiac output in part due to increased sympathetic stimulation of the heart (36) and in part due to increased central blood volume estimated by increased right atrial pressure and implying increased ventricular preload. Increased stroke volume has also been demonstrated in humans during rhythmic handgrip followed by post-fatigue ischemic occlusion (7). In the canine model, total peripheral resistance remains relatively unchanged during mild to moderate exercise (2) even though skeletal muscle resistance is increased (35) and there is marked renovascular constriction (2). Marked increases of renovascular resistance have also been observed in humans and primarily attributed to the mechanoreflex component of the pressor reflex (32, 46).

The observed increase in cardiac output requires increased venous return. It remains unclear precisely how this occurs. The purpose of the present work therefore was to explore the hypothesis that the splanchnic vasculature is the major source of centrally redistributed blood during the exercise pressor reflex and that redistribution occurs as a result of pressor reflex-induced splanchnic arterial vasoconstriction and venoconstriction. These studies are unique in that they employ impedance plethysmography to measure changes in regional blood volumes related to the stimulation resulting from static pressor reflex activation.

MATERIALS AND METHODS

Subjects and Experimental Outline

To test this hypothesis we studied 16 healthy volunteer subjects aged 20–27 yr (median = 24.5 yr, 7 male, 9 female). Average weight ($\pm$ SD) was 70 ± 14 kg, average height was 169 ± 10 cm, average body mass index was 24 ± 4 kg/m². Patients were normotensive. All measurements were made supine. Resting supine systolic blood pressure was 118 ± 10 mmHg, resting supine diastolic blood pressure was 63 ± 8 mmHg, and resting heart rate was 59 ± 9 beats/min.

All subjects were free from systemic illnesses. Subjects were not taking medications and were nonsmokers. All subjects had no evidence of cardiovascular or systemic illness. There were no competitive athletes or bedridden subjects. Informed consent was obtained. All protocols were approved by the Committee for the Protection of Human Subjects (IRB) of New York Medical College.

Laboratory Evaluation

We monitored blood pressure and heart rate continuously and estimated changes in thoracic, splanchnic, pelvic, and calf segmental blood volumes and blood flows by impedance plethysmography while the subjects were supine and throughout static handgrip as explained below. All subjects had blood volume and resting cardiac output measured by indocyanine green (ICG) dye dilution methods, and results were compared between impedance measurements and venous occlusion plethysmography, green dye cardiac output, and dye estimates of splanchnic blood flow (see *Dye dilution measurement of blood volume and cardiac output*).

Protocol

Tests began at least 2 h after a light breakfast. An intravenous catheter was placed in the right antecubital fossa. After a 30-min acclimatization period, we used supine impedance plethysmography (IPG) to continuously measure resistance (R_0) and beat-to-beat change in resistance (ΔR) of thoracic, splanchnic, pelvic, and leg segments (terms defined below) while we simultaneously estimated supine cardiac output, blood volume, and effective portal vein blood flow by the ICG dye dilution technique. Next, calf blood flow was measured by strain-gauge plethysmography (SPG). We verified respective correlations of impedance measurements of thoracic blood flow, calf blood flow, and splanchnic blood flow with dye dilution cardiac output, SPG calf blood flow, and the exponential decay coefficient of the concentration of ICG, which approximates portal blood flow divided by blood volume within a constant representing hepatic dye extraction (see *Dye dilution measurement of blood volume and cardiac output*). We also compared impedance methods with reference methods for measuring blood flow using Bland-Altman Plots (3). These results appear in Fig. 1. On this basis we used impedance measurements to continuously estimate changes in segmental blood flows throughout the handgrip evaluation.

Early in the experiment, each subject performed two brief maximal voluntary contractions (MVCs) with their left hand using a handgrip dynamometer (Lafayette Instruments, Lafayette, IN). Subsequently, subjects performed static handgrip while impedance, heart rate, and blood pressure monitoring continued. Handgrip was preceded by a baseline phase lasting 5 min, during which impedance, heart rate, and blood pressure data were collected. Subjects then performed 120 s of sustained isometric handgrip at 35% MVC. This was typically exhaustive exercise. Posthandgrip circulatory arrest was not performed because we were interested in the exercise pressor reflex rather than measurements made after exercise. A feedback system allowed subjects to maintain force near constant. Blood pressure, electrocardiogram, and IPG flow and volume measurements were made continuously but are reported at baseline, 1 min, and 2 min of sustained handgrip and during a recovery period.

Details of Method

Heart rate and blood pressure monitoring—A single electrocardiogram lead was recorded for rhythm. Upper extremity blood pressure was continuously monitored with a finger arterial plethysmograph (Finometer, FMS, Amsterdam, The Netherlands) placed on the right middle or index finger. Electrocardiogram and Finometer pressure data were interfaced to a personal computer through an analog-to-digital converter (DI-720 DataQ Ind,

Milwaukee, WI). Heart rate was derived from arterial pressure data. Finometer data were calibrated to a brachial artery oscillographic pressure. All data were multiplexed with strain gauge and impedance data and were thereby synchronized. Continuous blood pressure data were used to identify pulses and to compute heart rate and blood pressure variability indexes and to perform and compare coherence analyses among the subject groups.

Heart rate and blood pressure variability, coherence analysis—To investigate the effects of handgrip on the cardiovagal baroreflex regulation of heart rate, we measured indexes of heart rate and blood pressure variability. There is evidence for important effects of both the mechanoreflex and the metaboreflex components of the exercise pressor reflex on baroreflex function (18, 24, 54). The transfer function between blood pressure and heart rate at middle frequency (~0.1 Hz) relates to sympathetic modulation of blood pressure transduced to heart rate changes primarily by way of vagal efferents of the baroreflex (28). We examined coherence and transfer function phase and amplitude (baroreflex gain) during handgrip: baseline heart rate and blood pressure data were captured during the 5-min resting period. Beats were acquired for the first minute and second minute of handgrip. Data were analyzed for each minute separately and for both minutes combined as a single beat sequence. Data were also collected for a 1-min period during recovery from handgrip centered on the time of minimum blood pressure for comparison. We used custom software to collect digital sequences containing RR interval and systolic, diastolic, and mean blood pressures for each heartbeat, as previously described (47). The coherence function was also calculated at low frequencies. Usually a coherence of at least 0.5 is used to indicate a significant baroreceptor-mediated relationship between changes in blood pressure and changes in heart rate (38). Coherence less than 0.5 suggests an “uncoupling” of the baroreflex modulation of heart rate by blood pressure (28).

Dye dilution measurement of blood volume and cardiac output—We used ICG dye dilution technique to measure blood volume and cardiac output (4) and to estimate splanchnic blood flow in terms of portal uptake of the dye (44). We used a spectrophotometric photosensor (DDG2000, Nihon-Kohden) previously validated in clinical studies (16, 19). The dye decay curve fits a monoexponential $V_0 \exp - [K_t]$, where K represents clearance by the liver divided by blood volume and $\text{clearance} = (1 - \text{hematocrit}) \times Q \times E$ (44).

We defined effective portal blood flow $= Q \times E = (K \times BV)/(1 - \text{hematocrit})$, where BV is blood volume, Q is portal blood flow, and E is the hepatic dye extraction ratio.

We measured the hematocrit from antecubital venous blood, and extrapolated the dye decay curve to the time of dye injection ($t = 0$) yielding estimated blood volume. A semilogarithmic fit to the exponential decay yields the parameter K , which was used to estimate portal blood flow and thus splanchnic blood flow within a constant. Echo-Doppler measurements of portal venous blood flow and ICG clearance methods compare favorably (5).

Calf blood flow by SPG—We used venous occlusion strain-gauge plethysmography in all subjects to measure calf blood flow. Supine measurements were made at the beginning of

experiments, and measurements were compared with impedance estimates of blood flow. We have previously employed these techniques (48, 50).

Impedance plethysmography to measure changes in segmental blood

volumes and blood flows—Impedance plethysmography (IPG) has been used to detect internal volume shifts during orthostatic stress (8, 11, 33, 53). IPG is routinely used to measure changes in cardiac output in the form of impedance cardiography (8, 11). We have used this technique extensively before with repeatable and physiologically reasonable outcomes (49, 51). Our device employs a Tetrapolar High Resolution Impedance Monitor (THRIM) four-channel digital impedance plethysmograph (UFI) applied to four anatomic segments defined in practice by electrode placement. These are designated the thoracic segment (supraclavicular area to xyphoid process), the splanchnic segment (xyphoid process to iliac crest), the pelvic segment incorporating lower pelvis to the knee (iliac crest to knee), and the leg or calf segment (upper calf just below the knee to the ankle) (49, 51). Ag/AgCl electrocardiogram electrodes were attached at these segmental boundaries and also to the right foot and right hand, where they served as current injectors. The IPG uses a 50-kHz, 0.1 mA RMS constant current signal between the foot and hand electrodes. Electrical resistance values were measured using the segmental pairs as sampling electrodes. The midline distance between the sampling electrodes (L) was measured with a tape measure. We also measured the circumferences of calf, thigh, hips, waist, and chest to obtain approximate volume contents of each anatomic segment. We estimated the change in blood volume in each segment during handgrip from the formula:

$$\Delta \text{segmental blood volume (ml)} = \rho \times (L^2 / R_0 R_1) \times \Delta R \quad (11, 13).$$

where ρ is electrical conductivity of blood estimated as $53.2 \times \exp(\text{hematocrit} \times 0.022)$ given by Geddes and Sadler (12). R_0 is the resistance of a specific segment before handgrip, R_1 is the resistance during handgrip, and ΔR is the change in resistance ($R_1 - R_0$) in a specific segment during handgrip. ρ was regarded as constant during the maneuver.

IPG was also used to measure segmental blood flows (33). Transient blood flows have been similarly quantitated during orthostasis (49) and during the Valsalva maneuver (51). Pulsatile changes in electrical resistance for each segment were employed to compute the time derivative R/t , which we used to obtain the blood flow responses of each body segment during handgrip.

Blood flow was estimated for an entire anatomic segment from the formula:

$$\text{Flow} = [\text{HR} \times \rho \times L^2 \times T \times \partial R / \partial t_{\max}] / R_0^2 \quad (11)$$

where HR is heart rate, T is the ejection period, R is the pulsatile resistance, and R_0 is the baseline resistance at a given angle of tilt. Respiratory artifact was removed from the signal using a custom, Fourier-based frequency selection technique. IPG flows are expressed in milliliters per minute for anatomic segment and can be normalized by dividing by estimated segmental volume.

Statistics

All tabular and text results are reported as means $\pm$ SD. Graphics are presented as means $\pm$ SE. Changes in heart rate and blood pressure variability and in impedance estimates of regional blood flow and regional blood volume, heart rate, and mean and systolic blood pressures were compared by analysis of variance for repeated measures at baseline before handgrip, 1 min after the start of handgrip, 2 min after the start of handgrip, and during recovery using the minimum of blood pressure during recovery as the time of comparison. Results were calculated using SPSS (Statistical Package for the Social Sciences) software version 11.0.

RESULTS

All subjects completed the protocol successfully. Blood volume for each subject was normal, and the mean was 5.2 ± 1.5 liter (73 ± 12 ml/kg) and the mean hematocrit was $41 \pm 3\%$. Data were similar for males and females and pooled gender data are presented.

Heart Rate and Blood Pressure During Handgrip

Heart rate and blood pressure are shown in Fig. 2. Data from a representative subject shows the approximately linear increase in blood pressure as a function of time during handgrip. This linearity was observed in all subjects. Heart rate increased rapidly during the early course of static handgrip and more slowly thereafter. Percent changes in heart rate and blood pressure are based on paired data assessments and are also shown in Fig. 2. Absolute blood pressures (systolic and mean) and heart rates at stages of handgrip are shown in Table 1. On average systolic and mean arterial blood pressure increased significantly from baseline at 1 min of handgrip ($P < 0.0001$) and increased further at 2 min of handgrip ($P < 0.0001$) returning at recovery to a pressure that was slightly lower than baseline ($P < 0.001$). Heart rate increased significantly from baseline at 1 min ($P < 0.0005$) and further at 2 min ($P < 0.005$), returning at recovery to a heart rate that was similar to baseline.

Heart Rate and Blood Pressure Variability, Coherence Analysis

Decreased HRV during handgrip varied BPV—Data are shown in Table 1. Total heart rate variability (HRV) was significantly decreased compared with baseline ($P < 0.001$) at 1 and 2 min of handgrip and returned to baseline values during recovery. Low-frequency HRV power decreased ($P < 0.01$) during the first minute of handgrip but then was similar to baseline during the second minute of handgrip. High-frequency HRV power was decreased during both 1 and 2 min of handgrip but was similar to baseline during recovery. The ratio of low-frequency to high-frequency power (LF/HF) was increased during the second minute of handgrip compared with baseline ($P < 0.001$). Blood pressure variability (BPV) was decreased in the first minute of handgrip ($P < 0.01$) and was increased during the second minute ($P < 0.05$) compared with baseline.

Lack of significant low-frequency HRV-BPV coherence during handgrip—

Although transfer function gain (transfer magnitude) was markedly reduced ($P < 0.001$) throughout handgrip and recovery, this was associated with a reduction of coherence ($P < 0.001$) well below the 0.5-cutoff for significant coherence. Thus changes in transfer

magnitude (i.e., baroreflex gain) may not be strictly interpretable under these circumstances. The data suggest an uncoupling of baroreflex mediation of heart rate and blood pressure during handgrip.

Impedance Plethysmographic Changes in Segmental Blood Volumes and Blood Flows

Reciprocal thoracic (central)–splanchnic blood volume changes during handgrip—Representative changes in segmental impedance and in the percent change in blood volume during handgrip are shown in Fig. 3. There are opposite changes within the thoracic and splanchnic segments. Significant increases in impedance also occur within pelvic and calf segments but are of much smaller magnitude than splanchnic increments.

Central blood volume increases and peripheral volume decreases during handgrip—Percent changes in all segmental blood volumes are shown in Fig. 4. There is an increase in thoracic volume noted during the first minute of handgrip, which reaches significance ($P < 0.025$) during the second minute. Splanchnic volume changes in reciprocal fashion with smaller changes noted in pelvic and calf volumes, which both decreased significantly ($P < 0.05$) during the second minute of handgrip. Comparable changes in the percent volume of the calf diameter were measured by SPG.

Segmental blood flow changes during handgrip: stable splanchnic blood flow suggests venoconstriction—Changes in segmental blood flow are shown in Fig. 5. Cardiac output is increased by nearly 20% ($P < 0.01$) throughout handgrip, whereas splanchnic blood flow tends to be unchanged during handgrip, and pelvic and calf blood flows are increased at 2 min ($P < 0.025$). Stable splanchnic blood flow with decreasing splanchnic volume may indicate active splanchnic venoconstriction (see Discussion).

Segmental arterial resistance increases during handgrip; modest increase in total peripheral resistance—We calculated segmental vascular resistance using the formula $R = \text{mean arterial pressure} / \text{segmental blood flow}$. Percent changes in vascular resistance are shown in Fig. 6. Total peripheral resistance is modestly increased ($P < 0.025$) during the second minute of handgrip but returns to baseline during recovery. Similar increases in splanchnic, pelvic and calf resistances were noted.

DISCUSSION

Main Findings

Regional blood volume redistribution and hemodynamics during handgrip

Our results demonstrate an integrated approach by which changes in hemodynamically important regional circulations contribute to the redistribution of blood volume and blood flow during static handgrip. The most significant new finding is that the increase in central blood volume and cardiac output evoked by the exercise pressor reflex is produced in large part by emptying of the splanchnic vascular bed through splanchnic venoconstriction and arterial vasoconstriction. We observed a 4.5% increase in central blood volume associated with a 2.5% decrease in splanchnic volume. Since the splanchnic vascular bed receives ~25% of the cardiac output and contains ~30% of the blood volume, it is reasonable that the

splanchnic vasculature is able to rapidly transfer its blood to the central circulation (41). Relatively smaller but directionally similar decreases in segmental blood volume occur within the pelvic and calf segments. The finding of splanchnic emptying at relatively constant splanchnic blood flow implies the active contribution of venoconstriction. In addition, there is an increase in total peripheral resistance and therefore cardiac afterload. The magnitude of the increase in cardiac output, the increase in central blood volume, and the increase in end-systolic pressure suggest that there is also an increase in cardiac contractility.

Baroreflex regulation of heart rate during handgrip (cardiovagal regulation)—

Cardiovagal baroreflex regulation is calculated in the present work. This technique mainly estimates parasympathetic control of heart rate. If usual coherence criteria are employed, our data also imply that parasympathetic baroreflex regulation of heart rate may become relatively less important during handgrip. Note that the data do not inform on baroreflex effects on peripheral resistance or on cardiac contractility (28).

Comparisons with the Literature

Heart rate and blood pressure during handgrip—Our observations are consistent with the literature showing changes in both blood pressure and heart rate during and after release of hand grip. The time course and magnitude of increases in heart rate and blood pressure are similar to data observed elsewhere in healthy human subjects (1, 6, 10, 21).

Variability analyses—Our measurements of heart rate variability are consistent with past data of Kluess et al. (25) and with the extended analyses of Iellamo et al. (18) in showing a relative increase in low-frequency to high-frequency spectral components and a reduction in baroreflex gain. We would, however, interpret such data differently in light of the progressive decrease in coherence between heart rate and blood pressure, which falls well below the 0.5 threshold for significance of coherence (Table 1) and is synchronous with the reduction in baroreflex gain. This can be interpreted as resulting from an uncoupling of heart rate from blood pressure modulation of the baroreflex during the isometric static exercise pressor reflex.

Segmental hemodynamics—These data, showing segmental changes in blood volume during hand grip measured by impedance plethysmography, are new in humans and are consistent with other findings in both human and animal literature. They are also unique because these measurements are noninvasive. Thus, for example, O’Leary’s laboratory (45) have demonstrated the increase in central blood volume and ventricular performance (37) with consequent increased cardiac output (2), which occurs during the metaboreflex. This is in agreement with our measurements and with the observations on heart rate and stroke volume of Crisafulli et al. (7) during rhythmic handgrip contraction followed by local ischemic occlusion.

Increases in total peripheral resistance have also been shown in humans (17, 27) but not in normal dogs (2) during submaximal exercise. Thus in humans the increase in blood pressure during the pressor reflex has often been ascribed to vasoconstriction alone. This species

difference in vasoconstriction may represent a more situational than qualitative difference: thus pronounced vasoconstriction does occur by metaboreflex if cardiac output is held constant by ventricular pacing + β -block (45), during maximal exercise wherein cardiac output cannot increase further (2), during heart failure (15), or in normal dogs after arterial baroreceptor denervation (24).

Whereas a modest increase in total peripheral resistance was observed during our studies, it was combined with a proportionately greater increase in cardiac output related at least in part to an increase in central volume. Whereas an accurate calculation of ventricular elastance was not feasible using current data, the results suggest an increase in cardiac contractility as observed in dogs by Sala-Mercado and coworkers (43).

In addition, we found that arterial resistance increased in splanchnic, pelvic, and calf segments to a similar degree, although blood flow did not change in the splanchnic segment and actually increased in the pelvic and calf segments due to the increase in blood pressure. Similar observations concerning splanchnic resistance have been made by Waaler and associates (52). Since neither blood flow changes nor blood volume shifts were reported, it was not possible to draw conclusions concerning regional blood volume changes from their data. However, the lack of change in splanchnic blood flow with decreasing splanchnic segmental blood volume in our data may support the hypothesis of venoconstriction. Active venoconstriction is confirmed if there is a decrease in venous volume while venous pressure is kept constant. This may also be true if there is perfusion at constant flow provided there is no reduction in central venous pressure (40). Canine data suggest that there is actually an increase in right atrial pressure during ischemic exercise (45), which should result in an increase in central venous pressure. Taken together the data and literature support the proposal of active splanchnic venoconstriction during static handgrip in humans.

Limitations

Impedance plethysmography is an indirect measurement of blood flow, and the accuracy and validity of its use for such measurements may be questioned. We tried to resolve this issue by comparing impedance measurements to standard reference methods such as ICG dye dilution (for cardiac output and for portal venous blood flow) and venous occlusion plethysmography (for calf blood flow). Comparisons show reasonable albeit imperfect correlations between reference standard and impedance measurements. There are both fixed and proportional biases but no nonuniformities of error. Thus, within limits, and for purposes of continuous monitoring, we believe invasive testing can be replaced by noninvasive testing. However, whereas the ICG method can yield accurate cardiac outputs, neither this reference standard method (as implemented in the current study) nor impedance methods are capable of giving true splanchnic flow results. Thus with ICG dye we calculated dye clearance rather than portal or hepatic blood flow, whereas impedance methods are best suited for detecting relative or percent changes in blood volume. Invasive measures to obtain hepatic extraction ratios in each subject are beyond the scope of current studies and would need repeated reassessments throughout static handgrip.

ICG estimates of portal blood flow were hampered by the lack of computation of the hepatic dye extraction ratio. As explained in Materials and Methods, there is an implicit assumption

of equal extraction in all patients with an extraction ratio of 1.0. Similar extraction ratios seem to be a reasonable assumption among healthy subjects.

Heart rate and blood pressure variability indexes are not reference standards for autonomic or baroreflex measurements. However, measured indexes have been consistent with invasive forms of measurement such as microneurography. (39)

Our aim was to assess the hemodynamic responses induced by exercise pressure reflex during exercise. Other confounding factors, such as central command, humoral changes, physical factors, and other blood-borne biochemical factors may be present during exercise. This leads many investigators to perform measures immediately after peak exercise during peripheral circulatory occlusion ischemia, which was not done here. However, central command, whereas responsible for much of the increase in heart rate and respirations during exercise, does not appear to increase sympathetic outflow unless the intensity of the exercise is near maximal (20), and humoral contributions appear to require a more sustained form of exercise to achieve importance (34). Nevertheless, central command, in particular may contribute to measured changes in heart rate.

ACKNOWLEDGMENTS

We thank members of the Department of Pediatrics, especially its Chairman, Dr. Leonard Newman, and the Division of Pediatric Cardiology, especially its Director, Michael H. Gewitz, for unflagging support. We also acknowledge our intellectual debt to our mentors, Dr. Thomas H. Hintze, Dr. Gabor Kaley, Dr. David Robertson, and Dr. Phillip Low for constant inspiration and stimulation.

GRANTS

This study was supported by National Heart, Lung, and Blood Institute Grants 1R01HL-66007 and 1R01HL-074873.

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Fig. 1.

Top left: estimated cardiac index (CI) calculated using indocyanine green (ICG) dye dilution technique on the abscissa compared with CI calculated using impedance methods; top right: corresponding Bland-Altman plot. Middle left: estimated splanchnic blood using ICG dye dilution technique (ordinate) compared with splanchnic blood flow using impedance plethysmography (abscissa); middle right: corresponding Bland-Altman plot. Bottom left: calf blood flow measured by venous occlusion plethysmography (ordinate) and estimated by IPG (abscissa); bottom right: corresponding Bland-Altman plot. Correlation coefficients are

shown. There are small fixed errors in thoracic blood flow, as well as fixed and proportional biases in splanchnic and calf blood flows, but no nonuniformities of error.

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Fig. 2.

Left: representative heart rate (HR, *top*) and blood pressure (BP, *bottom*) from an actual subject during static handgrip. *Right:* changes in HR and mean arterial blood pressure (MAP) averaged over all subjects. Measurements are shown for 1 and 2 min after the onset of handgrip as well as during the recovery phase. * $P < 0.05$ compared with baseline.

Fig. 3. Splanchnic (dot-dash line) and thoracic (solid line) graphs are shown for measured impedance (*top*) and corresponding calculated percent change in segmental blood volume (*bottom*). Changes are reciprocal with increased impedance in the splanchnic segment and decreased impedance in the thoracic segment during handgrip. End, cessation of the handgrip.

Fig. 4.

Percent changes from baseline in thoracic, splanchnic, pelvic, and calf blood volumes during handgrip averaged over all subjects at 1 and 2 min after starting handgrip and during recovery. The largest percent changes (increases) occur in the thoracic (central) blood segment with smaller reciprocal decreases in splanchnic segment. Changes in the pelvic and calf segments are smaller than measured in the thoracic and splanchnic regions. * $P < 0.05$ compared with baseline.

Fig. 5.

Percent changes in segmental blood flow. From *top* to *bottom*: changes in thoracic, splanchnic, pelvic, and leg (calf). Blood flow increases for the central thoracic, pelvic, and calf segments but is relatively unchanged for the splanchnic segment. * $P < 0.05$ compared with baseline.

Fig. 6.

Percent changes in segmental arterial resistance. From *top to bottom*: changes in thoracic, splanchnic, pelvic, and leg (calf). Total peripheral resistance (TPR, thoracic resistance) was increased by the second minute of handgrip and was increased in splanchnic, pelvic and calf segments during the entire handgrip period. All resistances returned to baseline during recovery. * $P < 0.05$ compared with baseline.

Table 1

Heart rate and blood pressure data

	Baseline	1 min	2 min	1&2 min	Recovery
HR, beats/min	59 ± 8	76 ± 12*	82 ± 12*		58 ± 6
Systolic BP, mmHg	118 ± 10	134 ± 12*	148 ± 14*		116 ± 8*
MAP, mmHg	90 ± 7	102 ± 11*	114 ± 14*		87 ± 7*
Total HRV Power (TP, ms ² /Hz)	3,213 ± 836	1,683 ± 714*	2,207 ± 1156*	2,156 ± 895	3,245 ± 906
Low-frequency HRV Power, ms ² /Hz	1,071 ± 310	562 ± 279*	1,061 ± 710	769 ± 326	1,197 ± 367
High-frequency HRV Power, ms ² /Hz	1,476 ± 498	827 ± 417	847 ± 389	843 ± 377	1,546 ± 423
LF/HF ratio	0.90 ± 0.19	0.80 ± 0.17	1.76 ± 0.63*	1.02 ± 0.31	1.12 ± 0.20
Total BPV Power, mmHg ² /Hz	8.4 ± 2.0	6.8 ± 0.8*	9.8 ± 1.5*	9.5 ± 1.6	10.0 ± 2.3
Transfer coherence	0.76 ± 0.10	0.41 ± 0.06*	0.26 ± 0.05*	0.34 ± 0.05	0.42 ± 0.06
Transfer magnitude, gain·ms ⁻¹ ·mmHg ⁻¹	18 ± 2	11 ± 2*	8 ± 2*	10 ± 2*	11 ± 2*
Transfer phase (degrees)	24 ± 12	80 ± 46*	90 ± 40*	88 ± 42*	80 ± 17*

Values are means ± SD.

HR, heart rate; BP, blood pressure; MAP, mean arterial resistance; HRV, HR variability; BPV, BP variability.

* $P < 0.05$ compared with baseline.