Evaluation of Contractile State by Maximal Ventricular Power Divided by the Square of End-Diastolic Volume

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Background. Maximal ventricular power (PWRmax) reflects contractile state and has the potential to be noninvasively determined. However, its sensitivities to preload, afterload resistance, and inotropic state are incompletely defined. The present study determines these dependencies and proposes a novel power-based contractile index that is little altered by load.

Methods and Results. Seven open-chest, autonomically blocked dogs were instrumented with a proximal aortic flow probe, central aortic and ventricular micromanometers, and a conductance catheter for ventricular chamber volume. Preload was transiently reduced by left atrial hemorrhage, and afterload was increased by intraaortic balloon inflation. Inotropic state was pharmacologically altered by lidocaine, dobutamine, propranolol, or verapamil. PWRmax was highly preload sensitive, altering 1.7±0.1-fold a given percent change in end-diastolic volume (EDV). This preload dependence was reduced by dividing PWRmax by EDV². This latter index also displayed little change in response to as much as 60% increases in afterload resistance. PWRmax/EDV² varied directly with inotropic state, correlating to both the slope (Em) of the end-systolic pressure-volume relation (PWRmax · 1,000/EDV²=0.31 · Em−0.04, r=0.82, p<0.001) and the slope (A) of the dP/dtmax−EDV relation (PWRmax · 1,000/EDV²=0.025 · A+0.02, r=0.86, p<0.001). PWRmax values determined from the product of ventricular pressure and flow versus central aortic pressure and flow were nearly identical over a broad loading range, indicating that PWRmax may be noninvasively assessed (i.e., without requiring left ventricular chamber pressure).

Conclusions. PWRmax divided by EDV² provides a measure of contractile function that is little influenced by loading conditions and has potential for noninvasive clinical use. (Circulation 1991;84:1698–1708)

Ventricular contractile state is most effectively assessed by indexes derived from pressure-volume relations. Examples such as the end-systolic pressure–volume relation (ESPVR), stroke work or dP/dtmax (maximal rate of pressure rise)–end-diastolic volume (EDV) relation, or the ejection fraction–afterload stress relation are each obtained by measuring a pump function variable over a loading range to generate an index that incorporates load and is therefore more specific to contractile state change. However, these relations require measurement of ventricular chamber pressure in combination with dimensions (or volume); thus, their use has been primarily limited to invasive clinical and experimental studies. Although a noninvasive assessment of contractility has clear practical and clinical benefits, it remains elusive.

A group of contractility indexes with potential for noninvasive use are those based on ventricular power. Power is the instantaneous product of pressure and flow and thus is the rate of ventricular work. Maximal ventricular power (PWRmax) is an ejection phase index occurring early after the onset of aortic flow when central arterial and ventricular pressures are similar. Thus, in the absence of aortic valve disease, central arterial pressure could substitute for ventricular pressure in the determination of power. The power indexes (in particular, maximal power and rate of power rise) were studied 15–20 years ago by several groups. However, when these studies were conducted, methods for power measurement were
both invasive and not very precise. Because the indexes did not offer clear advantages over other more general and conceptually powerful approaches, they were in large part abandoned. However, recent developments in pressure, flow, and dimension recording techniques may now enable PWR\textsubscript{max} to be noninvasively determined, rekindling interest in these measurements.

In the present study, we examined the preload, afterload resistance, and inotropic sensitivity of PWR\textsubscript{max}. Invasive pressure-volume and pressure-flow data were obtained in reflex-blocked anesthetized dogs to accurately assess each dependency. Based on recent studies of the load and inotropic sensitivities of stroke work,\textsuperscript{4} we anticipated that PWR\textsubscript{max} would display marked preload sensitivity but minimal change from afterload resistance. We further hypothesized that normalization of PWR\textsubscript{max} to the square of EDV (PWR\textsubscript{max}/EDV\textsuperscript{2}) would in large part eliminate the preload dependence and thus generate a reasonably specific index for contractile state. Our results, based on both experimental data and theoretical model analysis, are consistent with these predictions.

**Methods**

**Preparation**

Seven adult mongrel dogs (25–30 kg) were anesthetized with intravenous pentobarbital (20 mg/kg) and fentanyl (0.3–0.5 mg/kg), intubated, and ventilated on a volume respirator. The chest was opened via a lateral thoracotomy at the fourth intercostal space, and the pericardium was incised. The proximal periaortic fat was dissected free, and a 12- or 16-mm ultrasound flow probe (Transonic, Ithaca, N.Y.) was placed around the root. Two micromanometer-tipped catheters (Millar, Houston, Tex.) were positioned—one in the central aorta via the right brachial artery, and the other in the mid left ventricular chamber via a femoral artery. A large intravascular balloon occlusion catheter (Meditech, Billerica, Mass.) was placed in the proximal descending aorta. A large-bore cannula was introduced into the left atrium and attached to a reservoir primed with dextran and normal saline. Animals underwent autonomic reflex blockade before study (20 mg/kg hexamethonium, bilateral vagotomy).

An 11-electrode volume (conductance) catheter was placed in the left carotid artery and advanced to the ventricular apex. This catheter provided an online volume signal for pressure-volume loop analysis.\textsuperscript{13,14} The principles, design, use, and limitations of the conductance catheter have been reported elsewhere.\textsuperscript{15,16} All catheters were placed with the use of fluoroscopic guidance. In addition to inspection of volume catheter position, segmental volume signals from the catheter were individually digitized to determine those electrode segments within the ventricular chamber compared with those at or above the aortic valve. An electronic switch device enabled only those segments within the chamber to be used in the total volume signal. Volume catheter calibration was performed by determining the parallel conductance offset (average value of at least four separate estimates) using the hypertonic saline technique validated previously\textsuperscript{15} and a mean gain obtained from the ratio of integrated flow (stroke volume) from the ultrasound flow probe to catheter-derived stroke volume.

Pressures, volume, and flow data were digitized at 200 Hz using custom-designed analog-digital acquisition and signal display software. Raw data were stored on removable hard disks for subsequent analysis.

**Protocol**

**Preload and afterload sensitivity.** Preload was transiently reduced by gradual left atrial hemorrhage into a reservoir yielding an average of 27±3 sequential loops for analysis. Preload was defined by the diastolic volume obtained from each pressure-volume loop. Afterload resistance was transiently increased by inflation of the intra-aortic balloon, yielding an average of 13 differentially afterloaded beats for each heart. Afterload resistance was quantified by the effective arterial elastance (E\textsubscript{s})\textsuperscript{18,19} equal to the ratio of end-systolic pressure to stroke volume. At a constant heart rate, this ratio primarily reflects total vascular resistance. This can be seen by approximating mean arterial pressure with end-systolic pressure.\textsuperscript{18} Mean resistance (R) is equal to the following equation:

\[ R=\text{MAP}/\text{CO}=\text{MAP}/(\text{HR} \cdot \text{SV})=(\text{MAP}/\text{SV}) \cdot T \]  

(1)

where MAP is mean arterial pressure, CO is cardiac output, HR is heart rate, SV is stroke volume, and T is the cardiac cycle length (seconds). Therefore, E\textsubscript{s}=end-systolic pressure/SV=MAP/SV=R/T. Thus at constant T, which occurred in the present study because of autonomic blockade, E\textsubscript{s} primarily reflects total resistance (sum of peripheral resistance and characteristic impedance).

All data were collected with ventilation held at end expiration. Preload reduction data were obtained in all seven animals at two different contractile states (control and reduced via 4 mg/kg i.v. lidocaine). In two animals, additional preload reduction data were obtained with enhanced contractile state (dobutamine, 10–15 g/kg/min). Afterload increase data were determined at one contractility level in a total of six animals.

**Contractile sensitivity.** The seven animals described above provided a total of 16 different contractile states. To supplement these data and thereby better define power index sensitivity to contractile state, we reanalyzed data from a previously published study (six dogs) of contractility effects on pressure-volume relations.\textsuperscript{20} In this prior investigation, inotropic state was altered over a wide range by intravenous administration of propranolol and verapamil. Aortic flow had not been directly determined in these experiments; therefore, flow and thus power were calcu-
lated by differentiating the volume \( [V(t)] \) signal. \( V(t) \) was first filtered with a five-point Hanning window, and then \( dV/dt \) was digitally calculated using a five-point weighted slope. To verify the results of this approach, \( \text{PWR}_{\text{max}} \) values obtained from the direct-flow signal versus \( dV/dt \) were compared over a wide preload range in the present principal study group \((n=7)\) for which both measurements were obtained. The results (Figure 1A) demonstrated an excellent correlation \((y=0.98x-0.02, \ r=0.96, \ \text{SEE}=0.2, \ n=484, \ p<0.0001)\) that was not significantly different from the line of identity.

Combining both present and prior data sets yielded a total of 43 different contractile conditions that were used to assess power index inotropic sensitivity. Power index values were compared with the slope of the ESPVR \((E_{\text{es}})\) and the slope of the relation between \( dP/dV_{\text{max}} \) and EDV. Both of these relations (one, an end-ejection phase; the other, an isovolumic phase) have been extensively studied and found to be load-insensitive measures of contractile function.\(^{1,2,4,5}\)

### Calculations

Ventricular power \([\text{PWR}(t)]\) is the product of instantaneous left ventricular pressure \([P_{L,V}(t)]\) and rate of volume change \([dV/dt]\):

\[
\text{PWR}(t)=P_{L,V}(t) \cdot dV/dt
\]

(2)

In the absence of mitral regurgitation (accepting a small error from ignoring coronary blood flow), \( dV/dt \) during systole=atrial flow \([F_{A0}(t)]\); thus:

\[
\text{PWR}(t)=P_{L,V}(t) \cdot F_{A0}(t)
\]

(3)

\( \text{PWR}_{\text{max}} \) is determined as the peak value of \( \text{PWR}(t) \), which is digitally calculated from the instantaneous pressure-flow product.

An important consideration for noninvasive applications of \( \text{PWR}_{\text{max}} \) is the accuracy of its measurement from central arterial (versus ventricular) pressure. To test this, we compared \( \text{PWR}_{\text{max}} \) obtained from Equation 3 (i.e., \([P_{L,V}(t) \cdot F_{A0}(t)]_{\text{max}}\)) with the maximal product of central aortic pressure and flow:

\[
\text{PWR}_{\text{max}}=[P_{A0}(t) \cdot F_{A0}(t)]_{\text{max}}
\]

(4)

The comparison is shown in Figure 1B with individual points obtained from beats during transient preload and afterload resistance change. The two \( \text{PWR}_{\text{max}} \) values were highly linearly correlated \((p<0.0001)\), falling along the line of identity. Therefore, for the present study, \( \text{PWR}_{\text{max}} \) was calculated by using Equation 4.

In addition to \( \text{PWR}_{\text{max}} \), pressure-volume data were used to obtain other measures of systolic function such as ESPVR. The locus of points of maximal \([P/[V-V_{\text{GL}}]]\) for the multiple loops during preload change were fit by linear regression to yield the end-systolic elastance \((E_{\text{es}})\), the slope of the ESPVR. Another contractility index\(^{5}\) the slope of the relation between maximal pressure derivative and preload volume, was also determined from these beats.

### Statistical Analysis

The data were primarily in the form of multiple points from each heart relating load (or contractility) alteration to power index change; therefore, analysis employed linear regression. To combine data from all dogs and derive meaningful group statistics, a multivariate regression model was used that factored in other condition variables as well as provided for
interanimal variation. For preload (EDV) dependence, the regression model was the following equation:

\[
PWR_{\text{index}} = b_0 + b_1 \cdot \text{EDV} + b_2 \cdot E_\text{a} + b_3 \cdot \text{LIDO}
\]

\[+ b_4 \cdot \text{DOB} + \sum_{i=1}^{6} d_i D_i + \sum_{i=1}^{6} e_i D_i \cdot \text{EDV} \]  \hspace{1cm} (5)

where LIDO and DOB are dummy variables coding for lidocaine or dobutamine (0 for control, and 1 for drug), \(E_\text{a}\) is the afterload parameter obtained at steady state before preload change, and \(D_i\) represents dummy variables for each dog (\(D_i = 1\) for dog \(i\) and \(D_i = 0\) for dog \(j\) (for \(i = 1-6\)), and \(D_7 = -1\) for dog \(7\). The regression output provided the mean regression of power index on EDV (\(b_0\) and \(b_1\)), allowing for individual variation in each animal's regression (\(d_i\), \(e_i\)) as well as for effects of afterload (\(b_2\) and \(b_3\)) and contractility (\(b_4\) and \(b_5\)). The model was simplified for afterload analysis because only one contractile state was used, and \(E_\text{a}\) and EDV covared during transient aortic occlusion; thus, only one was included. To assess the correspondence between power and contractility indexes (i.e., ESPVR and \(dP/dt_{\text{max}} = \text{EDV}\) relations), terms were included to account for heart rate, preload volume, and afterload \(E_\text{a}\).

Statistical analyses were performed on an AT-compatible 286 computer using the SYSTAT (Evans-

\[\text{to, Ill.}) software package.\]

**Results**

**Preload Sensitivity**

A typical recording (signal versus time) obtained during transient preload reduction is shown in Figure 2A. Power was digitally calculated from the aortic pressure-flow product (measured in watts); all of the other channels were obtained on-line. These data demonstrate a strong dependence of \(PWR_{\text{max}}\) on preload, a consequence of combined changes in arterial pressure and peak flow.

The relation between \(PWR_{\text{max}}\) and EDV is displayed directly in Figure 3 (left upper panel), showing data for a representative dog. Dividing \(PWR_{\text{max}}\)
by EDV (not shown) reduced the preload dependence; however, dividing PWR\text{max} by EDV^2 virtually eliminated it (see figure; note PWR\text{max}/EDV^2 is multiplied by 1,000 so it can be displayed on the same axis). The left lower panel of Figure 3 shows the same data but with each index (and EDV) normalized to its respective baseline value; thus, all start at 1.0. This reveals that PWR\text{max} is reduced by nearly 60% for a 30% decrease in EDV. Dividing PWR\text{max} by EDV^2 yields an index that deviates little from 1.0 despite EDV change, which is consistent with minimal preload dependence.

Individual regressions for normalized PWR\text{max} versus EDV (as in Figure 3, left lower panel) are provided in Table 1. Each relation revealed strong preload dependence, with an average slope of 1.71±0.48 and $r^2$ of 0.981. Data from all 16 runs were graphically combined by averaging values over equally spaced normalized volume ranges (Figure 4). Dividing PWR\text{max} by EDV reduced the preload dependence of PWR\text{max} but did not eliminate it. In contrast, PWR\text{max}/EDV^2 was only minimally influenced by marked preload change.

Multiple regression analysis of preload dependence (using the raw data) is provided in Table 2. EDV significantly influenced both PWR\text{max} and PWR\text{max}/EDV ($p<0.001$ for both), whereas this was not so for PWR\text{max}/EDV^2 ($p=0.624$). Contractile state

### Table 1. Preload Dependence of Maximal Ventricular Power

<table>
<thead>
<tr>
<th>Dog</th>
<th>Slope</th>
<th>Intercept</th>
<th>$r^2$</th>
<th>SEE</th>
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<tr>
<td>4a</td>
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<td>0.02</td>
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</tr>
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<td>0.012</td>
<td>13</td>
</tr>
</tbody>
</table>

Results of linear regression analysis of individual maximal ventricular power (PWR\text{max})–end-diastolic volume (EDV) relations are for each dog at varying contractility (a, b, c). Data were first normalized so that both PWR\text{max} and EDV started at 1.0 before preload reduction. Slope, intercept, correlation index $r^2$, SEE, and number of points per regression ($n$) are provided.
Afterload Sensitivity

Figure 2B displays signal-versus-time plots during acute descending aorta balloon inflation. Despite an increase in systolic pressure and \( E_a \), \( PWR_{max} \) was little altered. This occurred because there was a small but significant decrease in peak flow as pressure increased. Because \( PWR = P \cdot F \), it is proportional to \( R \cdot F^2 \), where \( R \) is resistance, \( P \) is pressure, and \( F \) is flow. For the group data, resistance increased by 57%, whereas peak flow decreased by 24%. Heart rate was not significantly altered. Thus, the net power change during aortic occlusion could be predicted as 
\[
(1.57 \cdot 0.76^2) = 0.91, \text{ or } 91\% \text{ of baseline power with an almost } 60\% \text{ increase in afterload resistance.}
\]
In the observed data, \( PWR_{max} \) actually increased slightly with aortic occlusion; however, this was more likely due to simultaneous increases in EDV during aortic occlusion (see Figure 2B). Figure 3 (right panels) also shows an example (from the same dog) of the \( PWR_{max} \)-afterload relation (panel C shows raw data, and panel D shows the same data normalized to baseline). Again, dividing \( PWR_{max} \) by \( EDV^2 \) minimized load dependence.

Group data, with \( PWR_{max} \) indexes and afterload resistance (\( E_a \)) normalized to their respective baseline control values, are shown in Figure 5. None of the indexes displayed much change for an initial 20% increase in afterload. However, with further resistance increase, concomitant EDV increase led to significant increases in \( PWR_{max} \). Normalization to EDV reduced this effect, but the changes were in large part eliminated by dividing by \( EDV^2 \). Multiple regression results for the afterload change (again, based on the raw data) are provided in Table 3 and were consistent with the graphic analysis.

Contractile State Sensitivity

Although \( PWR_{max}/EDV^2 \) was relatively insensitive to preload and afterload resistance change, it correlated well with several standard measures of contractility. Comparisons were made to two indexes: the slope of the ESPVR (\( E_a \)), and the slope of the \( dP/dt_{max} \)-EDV relation (A). Both relations were derived using multiple pressure-volume loops obtained during preload reduction under each contractile state.
state. PWR_max/EDV^2 (y) correlated with both slopes (y=0.025·A+0.02, 2=0.86, p<0.001 for dP/dt_max−EDV; y=0.31·E_a−0.04, r=0.82, p<0.001 for E_a). Figure 6 shows the results of these comparisons as well as the 95% prediction intervals for the regressions.

To test whether simultaneous changes in heart rate, preload, or afterload resistance that could accompany drug-induced alterations in contractility influenced the relation between PWR_max/EDV^2 and E_a or A, we again used a multivariate regression model that included these variables. The only factor with a significant influence on PWR_max/EDV^2 was EDV, and this was true only for the regression of PWR_max/EDV^2 versus E_a. In this instance, the dependence had a small negative slope (−0.036±0.014) (i.e., power index decreased with increasing EDV), which is consistent with higher volume at low contractilities but opposite to a direct preload effect as defined earlier.

Discussion

The purpose of the present study was to assess the load and inotropic sensitivity of PWR_max and to determine whether a reasonably load-independent contractile index could be obtained by dividing PWR_max by EDV^2. These data revealed a very strong dependence of PWR_max on preload volume with much less effect from varied afterload resistance. Dividing PWR_max by EDV^2 minimized both load effects over a broad range; however, PWR_max/EDV^2 was sensitive to inotropic change, directly correlating with E_a and the slope of the dP/dt_max−EDV relation. Finally, PWR_max values obtained from central aortic versus ventricular pressures (or aortic flow versus chamber volume derivative) were very similar, supporting potential noninvasive applications of this index.

Load Dependence of PWR_max: Prior Studies

PWR_max incorporates many aspects of ventricular performance, including magnitude and rate of pressure development, ejection rate, and ventricular work. However, prominent preload dependence limits its usefulness. Previously, investigators have addressed normalization of PWR_max in several ways. In an early clinical study of patients with a wide variety of disease conditions, Russell et al. found a good correlation between PWR_max/EDV and ejection fraction. However, this study did not systematically test loading sensitivity, nor did it compare this ratio with other load-insensitive measures.

Stein and Sabbah reported in a canine study that the maximal instantaneous rate of power rise [d(PWR)/dt]_max was preload and afterload independent, despite the fact that PWR_max itself varied with load. However, although preload increase (dextran infusion) led to little change in d(PWR)/dt_max, systolic pressure also changed little in this study and actually decreased as much as 25 mm Hg in some cases. This is opposite to what is expected from a pure preload increase, suggesting that complex loading and/or reflex activation occurred. Furthermore, the previously documented preload dependence of stroke work as well as PWR_max shown in the present study strongly suggest that d(PWR)/dt_max should also be volume sensitive. We confirmed this by determining the maximal average rate of power rise for each beat during the preload reduction runs. This mean rate
power was calculated by dividing PWRmax by the time interval from onset of flow to PWRmax. The data (Figure 7) show nearly the same preload sensitivity observed with PWRmax. Furthermore, d(PWR)/dtmax has the disadvantage of requiring further differentiation, which can amplify signal noise.

Last, another study normalized mean power (versus PWRmax) to estimated diastolic wall stress. Incorporation of wall mass and geometry considerations via a stress formula may be useful for contrasting absolute values of power between subjects with markedly different heart sizes or thicknesses; however, this requires modeling assumptions. The present data were obtained in normal canine hearts with a fairly narrow range of cardiac masses, but in clinical disease states, particularly those with substantially increased chamber volume or mass, heart geometry could be important. Stress normalization would be less critical for predrug and postdrug intervention measurements in the same patient or for studies combining power measurements with exercise.

Why PWRmax/EDV2?

The notion that PWRmax/EDV2 should be fairly free from load dependence yet sensitive to inotropic state can theoretically be supported on several grounds. Mean power is the product of mean pressure and flow and thus resistance multiplied by flow squared. Power divided by volume squared is therefore proportional to resistance divided by seconds squared. Mean arterial resistance is only minimally altered (in the absence of reflexes) with steady-state changes in circulating volume. Thus, at a constant heart rate and contractile state such as during preload reduction in our protocol, PWRmax/EDV2 should be little changed.

Another way to consider the preload dependence of PWRmax is to express power in terms of a time-varying elastance [E(t)]:

\[
\text{Power} = \frac{P(t) \cdot F(t) \cdot dV/dt}{E(t) \cdot [V(t) - V_0] \cdot dV/dt}
\]

(6)

where P(t) is instantaneous ventricular pressure and F(t) is instantaneous flow. Thus, power divided by volume squared has units of elastance divided by seconds. At a constant heart rate, it can be shown that PWRmax/EDV2 should be proportional to Ees (see "Appendix 1").

It is somewhat more difficult to predict the afterload dependence of PWRmax with a simple equation. Because PWRmax occurs early in systole ("Appendix 2"), one might expect it to be more influenced by changes in characteristic impedance (IZo) than peripheral resistance. However, these two parameters are difficult to vary independently in vivo. Therefore, we assessed this issue by computer simulation. In our model, the ventricle was represented by a time-varying elastance, and the arterial system was repre-
sented by a three-element windkessel. The simulation calculated both PWRmax and PWRmax/EDV \(^2\) for pure changes in preload (EDV), peripheral resistance (R\(_p\)), and |Z\(_c\)|. Model output was obtained (using typical values for E\(_{es}\), R\(_p\), |Z\(_c\)|, compliance, EDV, and heart rate taken from our experimental data) and then normalized to baseline so that the results could be compared with the experimental data shown in Figures 4 and 5.

The results (Figure 8) were remarkably similar to those obtained experimentally. For preload (top panel), the model predicted a strong EDV dependence of PWRmax but little-to-no dependence for PWRmax/EDV \(^2\). Changing only peripheral resistance (middle panel) over a twofold range altered both power indexes only slightly (model outputs are superimposable as EDV was held constant and the data normalized to baseline). PWRmax eventually declines to zero at either extreme of load (flow, and thus PWRmax equal 0 when R\(_p\) is \(\infty\), and pressure, and thus PWRmax equal 0 when R\(_p\) is 0), but plateaus in the physiological loading range, which is similar to that previously reported for stroke work.\(^{19}\)

Change in |Z\(_c\)| (lower panel) led to a somewhat larger change in PWRmax compared with pure change in peripheral resistance; however, this effect was still fairly small. Furthermore, as |Z\(_c\)| is relatively difficult to vary acutely pharmacologically or with exercise,\(^{24}\) this factor is somewhat less critical. Aging alters |Z\(_c\)|\(^{25}\) and one might anticipate differences in PWRmax values as a function of increased proximal aortic dimension and stiffness due to age; this is under investigation.

Limitations

Several experimental limitations should be considered. Aortic flow was measured by flow probe at the aortic root; therefore, a small error was incurred in not including coronary flow in this measurement. Heart rate was maintained constant in these studies, and as with dp/dt\(_{max}\) heart rate change alone could alter PWRmax. For studies in which heart rate significantly changed, multiplying PWRmax/EDV \(^2\) by cardiac cycle length (seconds) should help normalize for pure rate effects. These studies were conducted in autonemically blocked animals to prevent reflex activation from interfering with the interpretation of load or contractility dependency relations. Reflexes that stimulate (or lower) the inotropic state of the heart would be expected to alter PWRmax/EDV \(^2\), or any contractile index, for that matter. It is generally impossible to separate this factor in an integrated system unless reflexes are expressly blocked. This should be remembered when studies are performed in intact animals or a clinical setting.

The conclusions presented here for both experimental and theoretical model data depend on V\(_0\), the volume at zero chamber pressure, being relatively small compared with EDV (see "Appendix 1"). In conditions in which V\(_0\) may increase, such as with chronically dilated hearts with chamber remodeling, PWRmax/EDV \(^2\) may not be as load insensitive. It is in

![Figure 8. Plots of computer estimation of preload (upper panel), peripheral resistance (middle panel), and characteristic impedance (lower panel) influence on maximal ventricular power (PWRmax) and PWRmax/end-diastolic volume squared (EDV)\(^2\). The simulation used a time-varying elastance heart model (E\(_{es}\), 6.0 mm Hg/ml; V\(_0\)=0 ml) coupled to a three-element windkessel vascular model (baseline peripheral resistance (R\(_p\)=4.0 mm Hg/ml · sec\(^{-1}\)), characteristic impedance (|Z\(_c\)|=0.2 mm Hg/ml · sec\(^{-1}\)), and compliance (C\(_a\)=1.4 ml/mm Hg) at a heart rate of 100 beats/min and baseline EDV of 45 ml. These model parameters were derived from and typical of the experimental data. To compare model with group experimental results, model data were also normalized as in Figures 4 and 5. The model predicted a dependence of PWRmax on EDV (upper panel) and a minimal effect of EDV on PWRmax/EDV \(^2\) that were very similar to the actual experimental data (compare with Figure 4). In response to either 50% reductions or 100% increases in either R\(_p\) or r |Z\(_c\)| (middle and lower panels), PWRmax and PWRmax/EDV \(^2\) changed relatively little (latter response identical to PWRmax due to constant EDV and normalization). Of these two afterload parameters, |Z\(_c\)| change had a slightly larger effect, which is consistent with the fact that PWRmax occurs early in ejection.](image_url)
these instances, as noted above, that additional consideration of mass or wall stress would probably be needed.

All prior studies of loading influences on power or $PWR_{max}$ examined steady-state data at one or two loading states. In this regard, the present study is unique by assessing load dependence on a beat-by-beat basis over a broad range of loads within each heart and among hearts. It is always possible that a component of the observed response stems from the specific type of load intervention used. It is impractical to test all possible load maneuvers; however, the similarity between our experimental and model results (the latter representing idealized load changes) suggests this was not a significant limitation. Although the influence of right heart loading itself (via right ventricle–left ventricle interaction) was not directly tested, it should be noted that left atrial hemorrhage rapidly leads to lowered right heart filling pressures, so interaction effects were not eliminated.

**Clinical Implications**

Recent advances in noninvasive recording technology have renewed interest in power indexes. Doppler echocardiography and nuclear ventriculography can each provide reasonable estimates of aortic outflow and volumes. Central aortic pressure has been estimated using tonometers applied to either the carotid or subclavian pulse and calibrated with peripheral cuff pressures.

A more recent and quite promising pressure-recording technique uses an automated cuff device. By appropriate recording of cuff pressure and time to the onset of flow at the cuff (by ultrasound sensor) gated off the electrocardiogram, the calibrated ascending portion of the central aortic pressure waveform can be determined. The pressures obtained by this device compare favorably with invasive micromanometer measurements. Because power occurs early in ejection, it is only the ascending aortic waveform that is required to determine maximal power. Power estimation using this device and simultaneous nuclear gated scintigraphy was recently reported in a human exercise study of normal and postinfarction patients. Certainly, a practical advantage of $PWR_{max}/EDV^2$ lies in its assessment at steady state rather than requiring loading interventions as for ESPVR and related measures.

Our data should encourage future studies testing the ratio of $PWR_{max}$ to $EDV^2$ for noninvasive clinical systolic function assessment. Although absolute values may be influenced by complex adaptive changes that often accompany chronic disease states, this index should be useful as an adjunctive measurement for assessing relatively acute drug responses and for exercise testing where relative changes at preset levels of exertion are important. Future clinical studies are needed to define the ultimate clinical usefulness of this index.

**Appendix 1**

Modeling the heart by a time-varying elastance, we obtain:

$$P(t) = E(t) \cdot [V(t) - V_0]$$  \hspace{1cm} (7)

Substituting into Equation 2 for power, we obtain:

$$PWR(t) = P(t) \cdot dV/dt = E(t) \cdot [V(t) - V_0] \cdot dV/dt$$  \hspace{1cm} (8)

Because $PWR_{max} = [PWR(t_{max})]$ occurs early into ejection, we can approximate $V(t)$ at $t_{max}$ by $EDV$. If $V_0$ is small relative to $EDV$, Equation 8 becomes:

$$PWR(t) \alpha E(t) \cdot EDV \cdot dV/dt$$  \hspace{1cm} (9)

$PWR_{max}$ also occurs very near $dV/dt_{max}$ (see Figures 3 and 6), and $dV/dt_{max}$ directly varies with preload volume at a constant heart rate (i.e., $dV/dt_{max} \alpha dV/dt_{max}$). Thus,

$$PWR(t) \alpha E(t) \cdot EDV \cdot \lambda EDV \alpha E(t) \cdot EDV^2$$  \hspace{1cm} (10)

$PWR/EDV^2$ will be proportional to chamber elastance; thus, it is not surprising that there is a correspondence between $PWR_{max}$ and $E_{cs}$.

**Appendix 2**

Maximal power occurs just after flow deceleration. This can be shown as follows:

$$PWR(t) = P(t) \cdot dV/dt = P(t) \cdot F(t)$$  \hspace{1cm} (11)

$PWR_{max}$ occurs when $d[PWR(t)]/dt = 0$. Thus:

$$dP/dt = F(t) + P(t) \cdot dF/dt = 0$$  \hspace{1cm} (12)

Early in ejection (until $P_{max}$), $dP/dt$, $P(t)$, and $F(t)$ are all greater than zero. Thus, $PWR_{max}$ occurs when $dP/dt$ is less than 0 or when flow is starting to decelerate. This is important because systolic pressure wave reflections generally occur after this time and thus would not influence $PWR_{max}$.

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