



Lecture Notes: Cardiac Physiology and Monitoring

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“Do not look for the key lost in the middle of the street under the street light at the corner just because the light is better there.”

Physiologists, and in particular physician physiologists, have often fallen into the trap of measuring certain cardiovascular parameters to explain cardiac performance because they could be measured, rather than because they should be measured. In classical cardiac physiology classes, one generally learns about cardiac performance through plots of cardiac output versus right atrial filling pressures. One learns how cardiac output varies with changes in right atrial pressure, and how various other variables affect these cardiac output/CVP plots. Even the venerable Starling conducted his analysis this way. Although certainly the measurements he made were accurate, they fall into what an engineer would describe as the black box approach to performance analysis; that is, one puts something in on one side of a black box, not knowing what's inside the box, and then measures what comes out at the other side. To an engineer, this seems unnecessary for a device like the heart, which in many ways is a simple pump. The proper analysis for a pump was described many years before Starling was born, and one has to wonder whether he avoided such analysis out of ignorance, or simply because the right measurements could not be made. In any case, in this lecture we will use engineering analysis to look at ventricular function. These descriptions were first made by **Covell** at UCSD, and later refined and emphasized by **Suga and Sagawa** at Johns Hopkins.

For any pump, one needs to know the volumetric output versus the input and output pressures that the pump is operating with. Also, for any pump that operates in a cyclical manner, one needs to first ask whether the desired output occurs over a sufficiently long period of time that one may ignore the duration of the pump cycle. Thus, to properly analyze cardiac function, we must plot the cardiac output versus the input and output pressures on a per beat basis.

One can look at how input pressures affect cardiac performance by plotting left ventricular intracavitary pressure versus volume. If one were to allow the heart to remain constantly in diastole (that is, not beating), one could fill up the heart like an empty balloon and get a curvilinear plot that shows an initial asymptotic line, a transitional zone, and then a second asymptote. The initial portion of the curve is caused by the increasing volume in the ventricular cavity distending the cardiac sarcomeres. Since the amount of pressure it takes to distend a sarcomere is roughly equivalent for all sarcomeres in the heart, the initial portion of pressure/volume plot displays a constant slope. Once the heart is stretched to a point where certain sarcomeres are fully expanded, one then must exert pressure to stretch the connective tissue attached to the sarcomeres. At this point, the compliance of the ventricle drops and the pressure builds more rapidly with increases in intracavitary volume. Finally, once the ventricle is filled to a point that all of the sarcomeres are fully stretched, additional volume only serve to distend connective tissue. The pressure/volume plot then displays a new asymptote that essentially reflects the compliance of the connective tissue. The total curve of diastolic pressure/volume relationship is called the diastolic function curve.

If one were to fill the heart to a particular point on the diastolic function curve and then allow it to contract, but not eject (that is, completely close the aortic valve during systole), the

pressure would build up in the left ventricular cavity to a maximum point and then would fall back to the diastolic line. If one does this for multiple starting points along the diastolic function line, one generates a series of maximum systolic pressures that happen to fall in a straight line that intersects zero. This line is called the systolic function curve, and represents the maximum pressure that can be generated by the heart for a given intracavitary volume and inotropic state. Of course, the heart wouldn't be very useful if it only generated pressure. The diastolic and systolic function curves provide the envelope in which the heart must work when it actually ejects blood. What does ejection look like? Initially, the heart is relatively empty, containing about 50 cc of blood at the beginning of diastole in a 70 kg subject. During diastole, volume flows into the ventricle, and pressure increases along the diastolic function curve. At the end of diastole, the ventricle begins to contract and builds pressure up towards the maximum point found on the systolic function curve. This is known as isovolumetric contraction. Once the pressure inside the ventricle exceeds the pressure inside the aortic root, the aortic valve opens and ventricle decreases in size as blood is ejected. Pressure continues to build however, as the ventricle continues to pursue its maximum pressure line. At the end of systole, the ventricular pressure has essentially run into the systolic function curve, and no further blood can be ejected. The ventricle relaxes, and pressure falls in a phase known as isovolumetric relaxation. The cycle then begins again as the heart fills during diastole. The graphical path followed by this cycle is known as the pressure/volume loop, or PV loop for short.

Some of the more commonly known cardiovascular parameters are contained in the PV loop. The pressure at the beginning of the ejection, where left ventricular pressure exceeds the aortic root pressure, is diastolic blood pressure. The highest pressure obtained during ejection is systolic blood pressure. The difference between end-diastolic volume and end-systolic volume is the stroke volume. Another key point on the PV loop is where the loop contacts the systolic function line. This known as the end-systolic pressure/volume point, and will be discussed in more detail later.

All of left (or right) ventricular performance can be described by examining the PV loop. The initial loading conditions are set by the diastolic function curve, the maximum performance of the ventricle is set by the systolic function curve, and the actual performance is described by the ejection portion of the PV loop. To see how this analysis fits in with more classical descriptions of cardiac performance, we will now discuss preload, afterload, contractility, heart rate and rhythm, and see how these variables affect cardiac performance.

Preload was originally defined as the amount of weight placed upon an isolated papillary muscle. Thus, although this preload was measured in units of force, it actually represented the initial stretch of papillary muscle sarcomeres in a single dimension. Since for the intact ventricle we are concerned about performance in three dimensions, the appropriate parameter that describes the initial stretch of the ventricle is the end-diastolic volume. Thus, the volume at which the ventricle begins its contraction is the most accurate definition of ventricular preload. If preload is increased (and afterload, contractility, rate and rhythm are held constant), the heart begins its PV loop at a higher end-diastolic volume, and follows an ejection curve concentric with the initial PV loop, but at a higher level of pressure. The net result is an increase in blood pressure and stroke volume, exactly what

one would expect with a large rapid infusion of volume into a patient. Conversely, decreases in preload produce a concomitant lowering of pressure and stroke volume.

Although easy to define, preload is relatively difficult to measure. When the heart can be directly observed, as during cardiac cases when the right ventricle is visible in the field, one can make qualitative statements about the extent of ventricular filling. During your rotation on the cardiac service, you will learn to distinguish between hypovolemia, normovolemia, and overdistension of the ventricle. Usually, however, we must depend upon our monitors to measure preload. Transesophageal echocardiography is probably the next best visual mode of estimating ventricular filling, but likewise only provides only a qualitative estimate. If one assumes a normal ventricular diastolic compliance, then the left ventricular end-diastolic pressure can be used to indicate left ventricular end-diastolic volume. Of course, it is rare that a ventricular catheter is inserted. However, by measuring the left atrial pressure during the time that the atrio-ventricular valve is open, one can get left ventricular pressures for LA line measurements. Such lines are also not typical outside of the cardiac room; rather, LA pressure is usually obtained by the use of a pulmonary artery catheter that is wedged into the pulmonary artery. Although there are artifacts that can effect this measurement, there generally is very close correlation between the wedge pressure and left atrial pressure. In fact, in the presence of congestive heart failure, left ventricular hypertrophy on chest x-ray, or ejection fraction lower than 40%, the CVP is a notoriously poor correlate of LAP. One should also note that left atrial pressure and, indeed, left ventricular end-diastolic pressure, can be misleading as a measure of left ventricular end-diastolic volume. Changes in diastolic compliance can alter LVEDP with no change in LVEDV. Such compliance changes are common during myocardial ischemia, but also occur frequently after cardiopulmonary bypass.

Changes in preload are most easily produced by rapid volume infusion, although changes in venous capacitance can also increase left ventricular filling. Thus, agents like ephedrine increase preload by increasing venous tone, while a spinal anesthetic will empty the heart because of an increase in venous capacitance. A MAST suit is a physical means of decreasing venous capacitance and thereby increasing ventricular preload.

Afterload, for an engineer, is relatively easy to define; in the human system however, it is somewhat difficult to measure. Unfortunately, multiple definitions of afterload exist that may not be correct from an engineering standpoint. In the papillary muscle model, afterload is defined as the weight placed on the papillary muscle after the preload weight has been applied. It is hard to develop an exact analogy for this model in the intact ventricle. Several sources prefer to define afterload by measuring the wall tension in the ventricle. In the opinion of this author, afterload is always determined by the system into which a pump is trying to eject, therefore measurements of afterload that use ventricular parameters are misleading. Proper analysis defines the afterload as the complex impedance facing the ventricle trying to eject blood. By complex, one is including those aspects resisting ejection that are caused by vascular resistance, vascular compliance, and blood flow inertia. Inertia represents the portion of afterload due to the need to accelerate a certain quantity of blood from zero velocity inside the ventricle to its peak velocity through the aortic valve. This is analogous to the force the shot putter must apply to the shot in order to get it down the field. Compliance takes into account the force that must be expended in order to distend the vessel walls; as such, it is analogous to the force one

must exert to blow up a balloon or compress a shock absorber. Finally, resistance is a measure of the force necessary to overcome the natural tendency of any fluid to stick to the vessel wall. This is the same force that tries to deter us from sliding furniture along the floor, and is the force that holds tires on the road. Depending upon the physical system, any of these three components, resistance, compliance or inertia, may predominate in determining how much force a pump must need to eject its volume. For the left ventricle, 70% of its effort is spent overcoming resistance, an additional 20% overcoming inertia. Although it is difficult to come up with a numerical estimate of this complex impedance, one can see its effect on ventricular ejection by examining the PV loop between the start of ejection and end-systole. A vascular system that is highly compliant has a very bowed ejection pattern, with a peak systolic pressure that is significantly higher than the end-systolic pressure. Atherosclerotic vascular trees that have limited compliance show a relatively straight line between the diastolic pressure point and end-systole. An increase in afterload, from whatever cause, makes it more difficult for the ventricle to eject. Increases in afterload cause the peak pressure generated by the ventricle to increase, while at the same time forcing the stroke volume to fall.

Although it is possible to estimate the compliance of the systemic vascular tree and the energy needed to accelerate blood, it is sufficient to use the systemic vascular resistance as a measure of left ventricular afterload. This is because, as noted before, SVR is the predominant determinant of the complex impedance facing the left ventricle. Further, changes in left ventricular afterload are almost entirely due to changes in SVR. Systemic vascular resistance can change by an order of magnitude, whereas vascular compliance can change only as much as a factor of two; the same for inertia. The equation for calculating SVR is:

$$(\text{Mean blood pressure} - \text{mean CVP}) / \text{cardiac output} = \text{SVR}$$

Methods to change this resistance are quite familiar to you. Alpha beta and delta agonists, as well as a host of neurohumoral agents and direct muscular dilators, are either increasing or decreasing vascular resistance.

Contractility is the most difficult parameter to define, as it has no clear engineering or physical correlate. Most physiologists feel that the term contractility refers to the innate strength of the ventricle, independent of its loading conditions. Measurements that have been used to estimate this term include cardiac output, maximum systolic pressure, dp/dt , maximum dp/dt , stroke work, stroke work index, phase of the moon and sign of the zodiac. Luckily, although difficult to define succinctly, one can measure contractility with reasonable exactness. The systolic function curve defines the upper limit of pressure performance for a ventricle. The slope of this line is thus an accurate indication of how powerful a given ventricle is. As this line is defined as the maximum pressure when the heart is not ejecting, it is difficult to measure both in-vivo or in-vitro. A more acceptable way of finding this line is to generate several PV loops for a ventricle while altering either the afterload or preload, while simultaneously preventing any reflex endogenous or exogenous changes in inotropic state. This produces several end-systolic pressure/volume points which lie in a line that is almost exactly equal to the systolic function curve. When derived in this manner, the line is called the end-systolic pressure/volume relationship (ESPVR). Currently, this is the most accurate measure of contractility, and is unique in its total independence of loading

conditions. **In fact, when perusing the literature to determine the effects of a given drug on myocardial contractility, if the ESPVR was not measured, I would personally ignore the article.** An agent which increases contractility rotates the PVR curve to the left. If both preload and afterload are held constant, this produces an increase in blood pressure and stroke volume. Likewise, negative inotropes rotate the ESPVR to the right, resulting in lowered blood pressure and lower stroke volume for the same loading conditions.

In isolated heart preparations in-vitro, the ESPVR is relatively easy to measure. In-vivo, one can accomplish such measurements with continuous echocardiographic estimates of left ventricular volume while simultaneously altering the preload or afterload with volume or vasopressors, respectively. This cannot be done on line however, and is rarely done clinically. An excellent estimate of contractility can be obtained by observing the ventricle in the open chested patient. A ventricle with normal contractility is said to have a “snappy” contraction. That is, its rate of contraction increases during ejection. A normal contraction is also notable for a twisting of the ventricle along its long axis, along with the translation of the apex towards the ventricular base. Another interesting way to estimate ESPVR is to observe the change in blood pressure during a rapid infusion of volume into the patient from the cardiopulmonary bypass machine. An infusion of 100cc's of fluid that produces a 30 to 40 mmHg increase in blood pressure is indicative of good contractility. Significant increases in stroke volume during volume induced increases in preload are also a sign of good inotropic state. Multiple agents that can change myocardial contractility and anesthetics have effects on inotropic state.

How does heart rate affect ventricular performance? If preload, afterload, and contractility are held constant, then heart rate has no effect on myocardial performance.. That is, during a given contraction, the heart does not care how long it was since the last beat, or how long it will be until the next beat. It is certainly true that in-vivo, almost all agents that affect heart rate also affect either preload, afterload or contractility, and therefore it becomes difficult to separate the affects of changes in heart rate from changes in the other hemodynamic variables. There also is certainly a range of heart rates in which the ventricle works most efficiently. Going below or above this rate does affect ventricular performance (although the purist would say that this is due to a change in contractility at the outlying heart rates and not a direct effect of the heart rate itself). Usually, increases in heart rate from 60 to approximately 120 produce increases in cardiac output, although the stroke volume may fall during this change, so cardiac output does not increase on a percentage basis as much as heart rate does. In general, analysis of ventricular performance, particularly in the anesthetized patient, should avoid focusing on heart rate and rather should focus on loading conditions and contractility. Heart rate is easy to measure, but remember that direct observation of the heart may settle confusion when the EKG is scrambled or an arterial waveform is not available.

Cardiovascular rhythm does not directly affect ventricular performance, but rather alters preload. For a given mean pulmonary artery pressure, sinus rhythm produces a larger left ventricular end-diastolic volume than any rhythm in which the atrium does not contract before the ventricle. The mechanism is analogous to a skyscraper which uses pumping stations in series approximately every 6 to 10 stories, in order to minimize the pressure needed on the ground floor in the primary pumping station. Increases in mean pulmonary

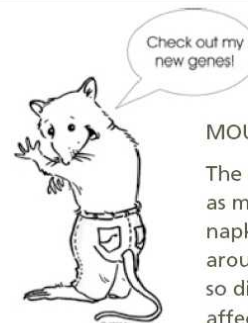
pressure (or mean CVP pressure for the right side) can always compensate for a non-sinus rhythm, with the limitations of pulmonary edema for the left side and venous congestion on the right. Although electrocardiographic analysis is the gold standard for defining rhythm, once again direct observation of the heart should provide even more clues. If atrial contraction is followed by ventricular contraction on a one to one basis, it is sinus rhythm regardless of what the EKG shows. CVP or LA pressure trends can also be useful for determining rhythm. Sudden loss of atrial kick produces an immediate increase in atrial pressures, which can be observed if one is continuously recording CVP or LAP.



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