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Physics of Invasive Blood Pressure Monitoring

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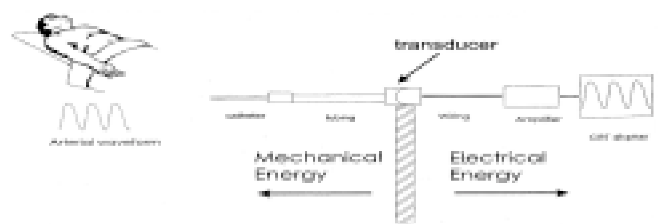
Study: Physics of Invasive Blood Pressure Monitoring
Introduction

Intra-arterial cannulation allows for continuous, beat-to-beat blood pressure measurement – it is considered the gold standard of blood pressure monitoring techniques. The quality of the transduced arterial pressure waveform depends on the dynamic characteristics of the catheter-tubing-transducer system. As clinicians, we strive to understand both the physiological and physical limitations of these measurements, judge the potential for error, and intervene appropriately in the face of these uncertainties.[1]

Discussion
Technical Aspects of Direct Blood Pressure Measurement

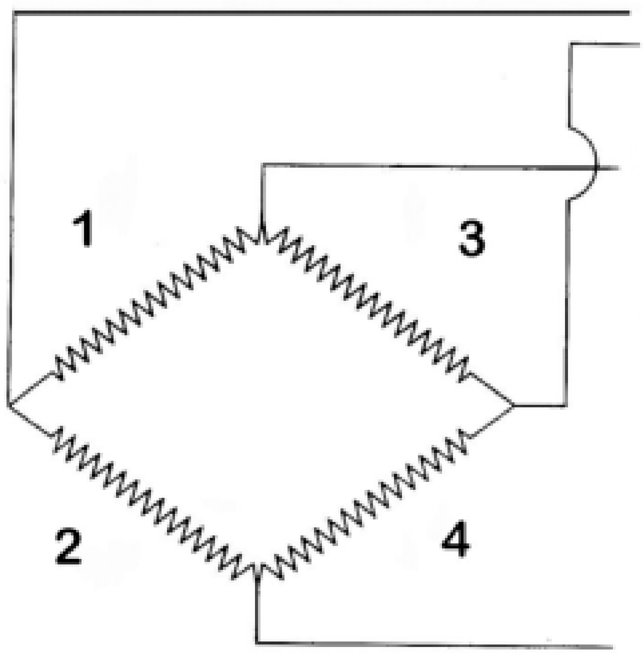
Catheter-transducer systems as used in the operating theatre and intensive care are characterized by an "underdamped, second-order dynamic system"[2] which is analogous to a bouncing tennis ball. Upon dropping the ball, it bounces several times and comes to rest on the floor. With each successive bounce, it does not rise as high as before. Each bounce has a characteristic frequency, and the time it takes the ball to come to rest is related to the damping coefficient.[3]

Figure 1: A common transducer in anesthesia changes mechanical energy (arterial pulse) into electrical energy



(From Miller RD (ed): Anesthesia, 5th ed, Churchill Livingstone 2000, fig 28-3)

Figure 2



A Wheatstone bridge in a strain gauge pressure transducer is connected to a deformable diaphragm. Stretching of a wire changes its electrical resistance. When pressure is applied to the diaphragm, strain on two of the resistors (no 2 and no 3) increases, whilst strain on no 1 and no 4 will decrease. The change in total resistance across the bridge is proportional to the change in blood pressure.

(From Morgan GE, Mikhail MS (ed): Clinical Anesthesiology, 2nd ed. Appleton & Lange, 1996, fig 6-15)

All pressure-monitoring systems attempt to accurately convert the physical energy of pressure-induced movements of a transducer diaphragm to electrical energy. The fidelity with which the system performs is dependant on the transducer (being the weakest link), its electrical components and the catheter-tubing system.

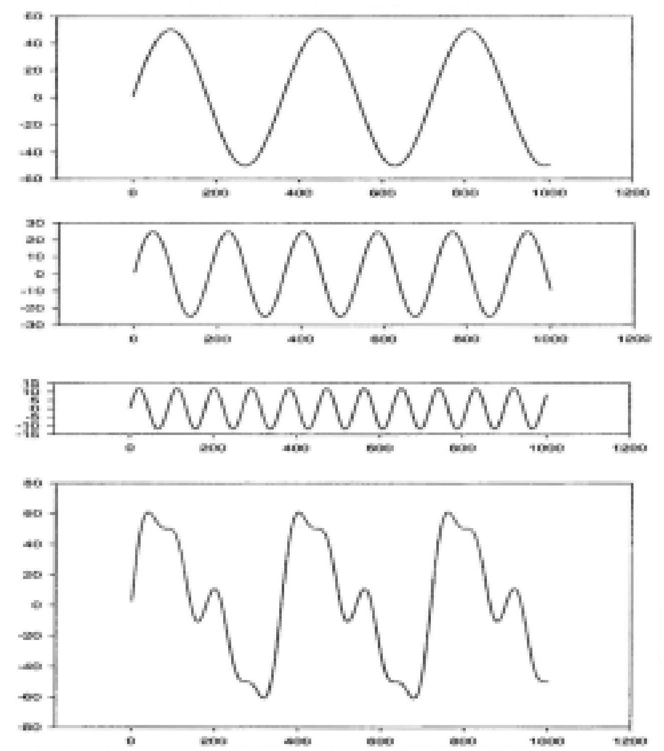
Frequency Content of the Arterial Pressure Waveform

An arterial pulse wave contains a fundamental frequency and series of harmonics.[4] The fundamental frequency is the lowest frequency sine wave and is equal to the pulse rate, thus the first harmonic. The second harmonic is a sinusoidal waveform with a frequency twice that of the fundamental harmonic. The waves are in phase, moving in the same direction and passing through zero amplitude together.[5]

The ideal measurement system should deal with all the harmonics of the input waveform in the same way. The amplitudes of the output harmonics will bear a constant ratio to the corresponding amplitudes of the input harmonics and there will be no phase difference between the output and input harmonics. It is rarely possible to avoid phase shift, but a phase shift proportional to frequency is acceptable.

The natural frequency of the measuring system must exceed the natural frequency of the arterial pulse (approximately 16-24 Hz).[6] The lower harmonics have the greatest amplitude. By reproducing

Figure 3: Fourier series



Any complex waveform, such as an arterial pulse wave, is the sum of simple sine and cosine waves

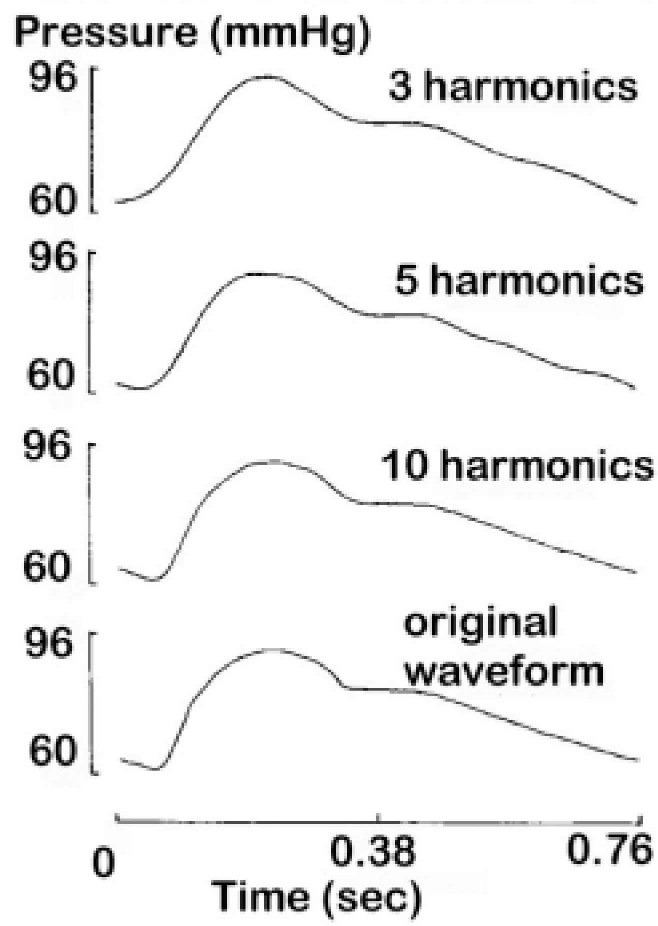
(From Miller RD (ed): Anesthesia, 5th ed. Churchill Livingstone, 2000, fig 28-6)

the fundamental and first to six ten harmonics, an approximation of the arterial pressure waveform can be obtained.[4]

A pulse rate of 70 beats per minute (bpm) would require a frequency response which is undistorted up to (70 X 10)/60=11.7 Hertz (Hz). To reproduce pulse rates up to 140 bpm, a flat frequency response of up to 20 Hz is required. The faster the heart rate and the steeper the systolic pressure upstroke, the greater the dynamic re-

continued on page 36

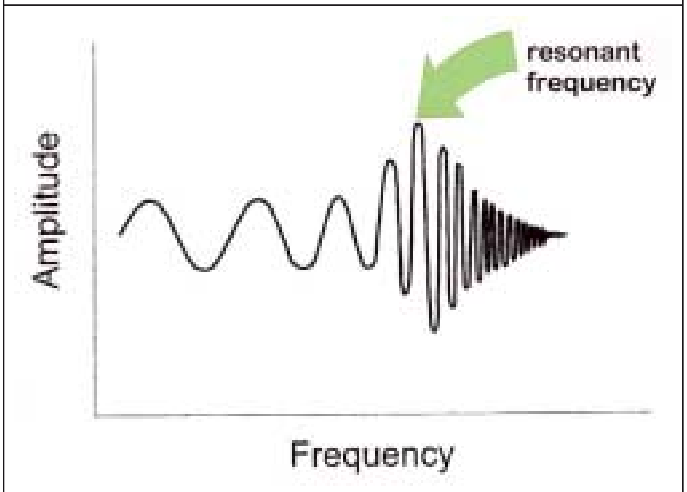
Figure 4: Pressure waveform from the ascending aorta



Increasing numbers of harmonics of a Fourier series are used to achieve progressively closer approximation to the original waveform, in this case a pressure waveform from the ascending aorta.

(From Scurr C, Feldman S, Soni N (ed): Scientific Foundations of Anaesthesia, 4th ed. Heineman Medical Books, 1990, fig 5-9)

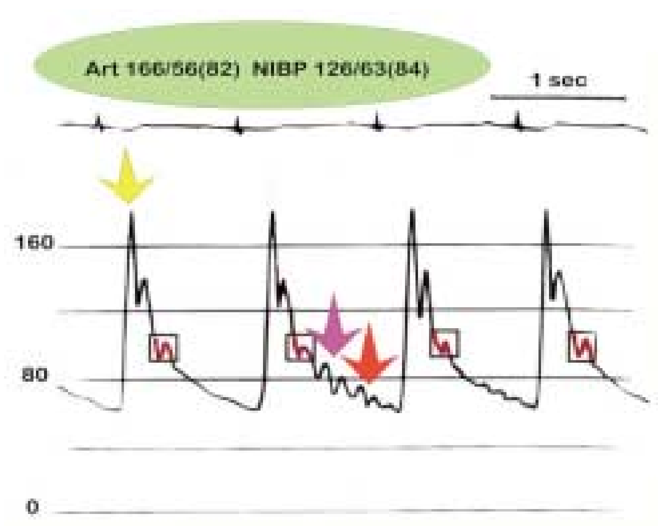
Figure 5: Resonant frequency



Amplitude of oscillation of a diaphragm in a catheter-transducer system as the applied frequency is increased. The amplitude is maximal at the resonant frequency of the system but at higher frequencies the diaphragm fails to follow the applied pressure.

(From Sykes, Vickers (ed): Principles of Measurement & Monitoring in Anesthesia and Intensive Care 3rd ed, fig 13-8)

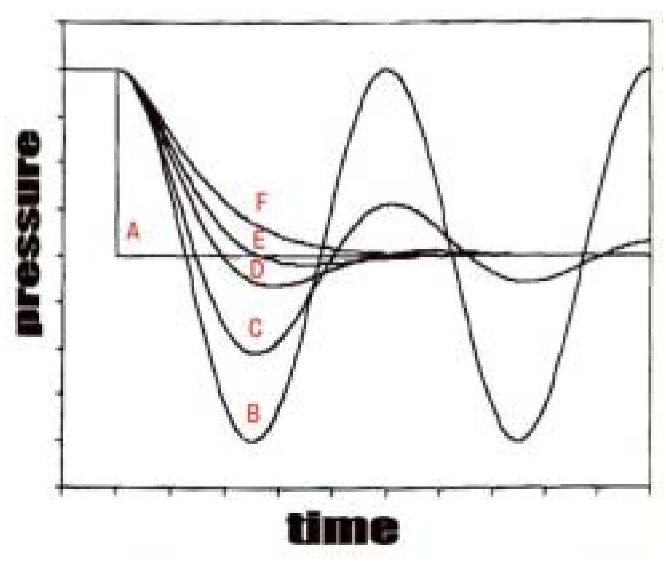
Figure 6: Underdamped arterial pressure waveform



- systolic pressure overshoot
- waveform distortion (red arrow)
- dicrotic notch (box) difficult to discern
- poor correlation between NIBP and IBP

(From Mark JB: Atlas of Cardiovascular Monitoring. New York, Churchill Livingstone, 1998, fig 9-4)

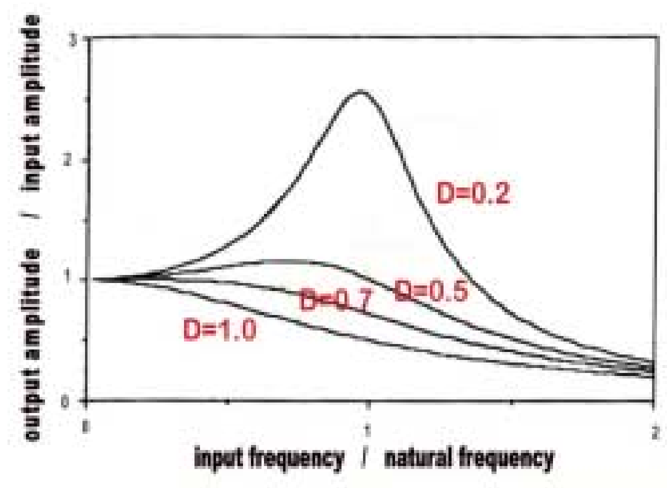
Figure 7: The effect of damping on the system response to a step change in pressure



A. the input signal
B. response of a system with zero damping
C. response of a system with a damping coefficient of 0.2
D. response of a system with a damping coefficient of 0.5
E. response of a system with an optimal damping coefficient of 0.7
F. response of a system with a critical damping coefficient of 1.0

A damping coefficient around 0.7 yields the best compromise between fast response time and small overshoot. Small damping factors allow amplitude overshoot; large damping coefficients result in low relative amplitudes below the natural frequency.

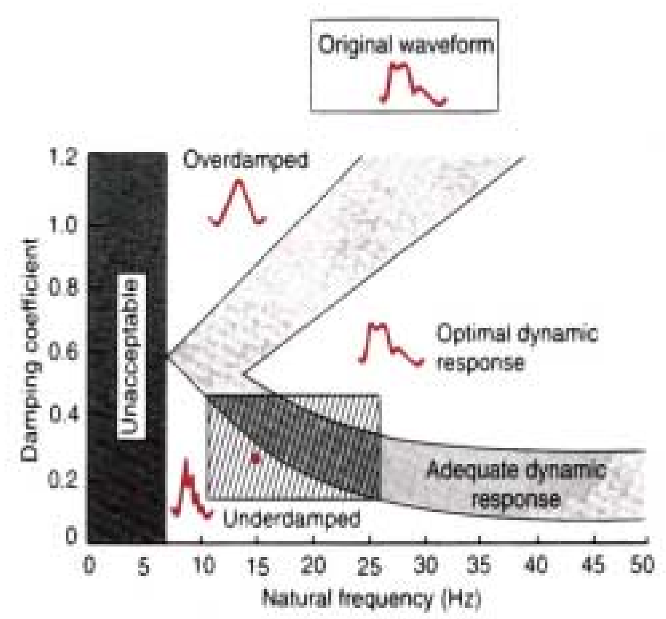
Figure 8: The effect of damping on the frequency response



Increasing the damping coefficient (D) results in less overshoot from oscillations near the natural frequency of the system. It also decreases the oscillating frequency. A damping coefficient around 0.7 is optimal, because it yields an output-to-input amplitude ratio close to unity over the widest frequency range.

(From Lake CL(ed): Clinical Monitoring for Anesthesia and Critical Care, 2nd ed. WB Saunders, 1994: fig 3-2)

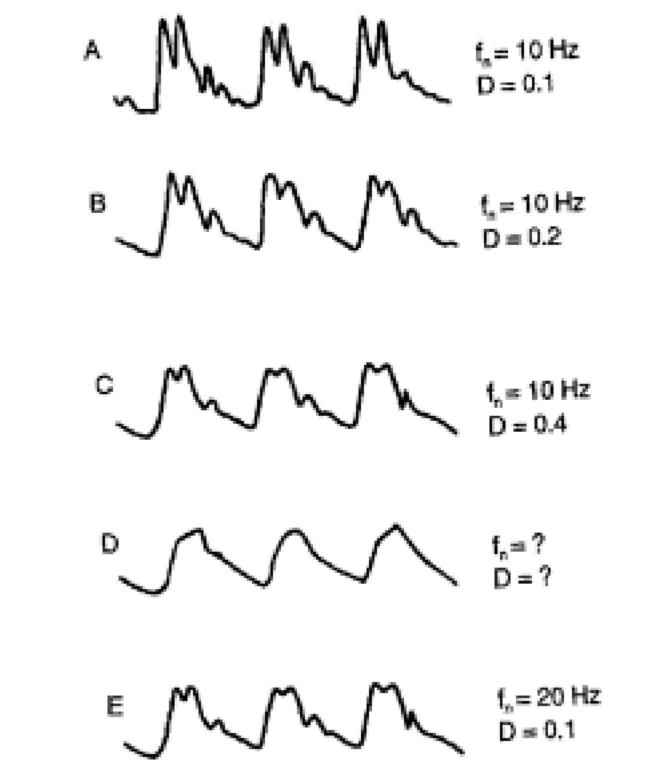
Figure 9: Interaction between damping coefficient and natural frequency. Catheter tubing-transducer systems fall into one of five different response ranges:



1. adequate dynamic response – accurate recording of most pressure waveforms seen in clinical practice
2. overdamped
3. underdamped
4. unacceptable – system with natural frequency <7Hz
5. optimal faithful recording

Rectangular box: ranges seen in clinical measurement systems

Figure 10: Interaction between damping coefficient (D) and natural frequency (fn)

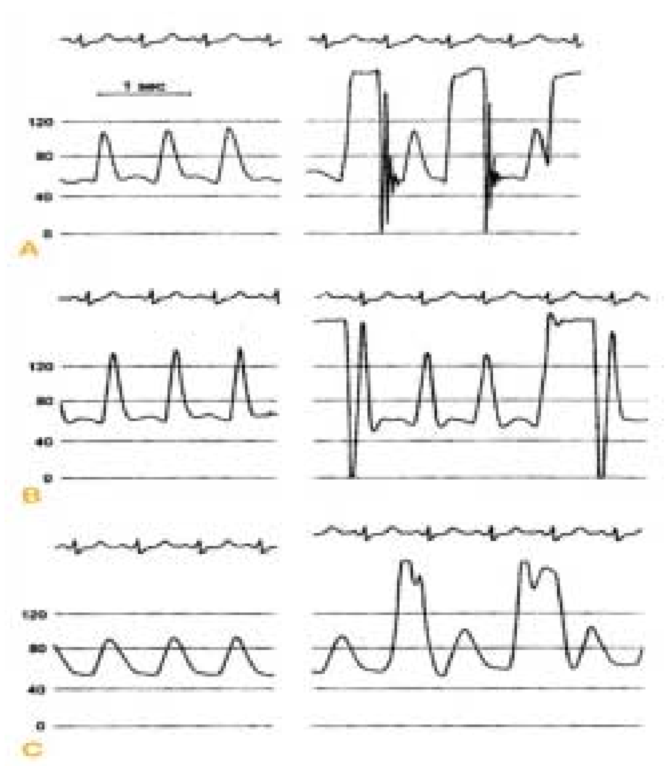


A. underdamped: small artifacts, systolic overshoot
B. damping coefficient increased: artifacts diminished
C. critical damping: accurate pressure wave, even though fn is low
D. overdamping: detail lost and determination of fn or D precluded
E. increased fn allows a low damping coefficient to have very little impact on waveform morphology

Note: waveform C and E are very similar

(From Mark JB: Atlas of Cardiovascular Monitoring. New York, Churchill Livingstone, 1998: fig 9-7)

Figure 11: The effect of small air bubbles within arterial pressure monitoring systems



A. adequate dynamic response (fn 17 Hz, D 0.2)
B. small air bubble in the monitoring system
 · decreases natural frequency
 · paradoxic increase in arterial blood pressure
C. big air bubble in the monitoring system
 · degrades dynamic response
 · spurious arterial hypotension

(From Mark JB: Atlas of Cardiovascular Monitoring. New York, Churchill Livingstone, 1998: fig 9-14)

sponse that is required from the monitoring system. Venous pressure waveforms, on the other hand, do not have steep waves or high frequency response.[7] To reproduce the first ten harmonics, transducers should have resonant frequencies, not at 15 Hz, but at 10 times that level i.e. 150 Hz.[8] Such a system is not practical and another physical characteristic, namely damping, must be added to the system.

Natural Frequency and Damping Coefficient

The transducer and pressure tubing is an oscillating system with its own natural frequency. When the signal rate (heart rate) approaches this natural frequency, the system will resonate and recorded pressure waveforms will be an exaggeration of true intra-arterial pressures.

Damping is added to keep the system from approaching its natural frequency and therefore "ringing."[8] All the systems are damped by friction and the viscosity of the fluid filling the catheter. Damping factors between 0.64 and 0.77 are optimal for blood pressure monitoring. The phase lag is also linear over the widest frequency range with a damping factor of 0.64.[5]

A major requirement of any catheter transducer recording system is that it has a high natural frequency to allow for the largest possible latitude in damping coefficient. This is achieved by limiting the length of tubing and using stiff tubing designed for pressure monitoring. Normal extension tubing, which is too compliant, will cause excessive damping because the natural frequency is low. Blood clots and air bubbles within the tubing-stopcock will adversely affect the dynamic frequency in the same way.

Clinical pressure monitoring systems are typically underdamped, displaying some degree of systolic pressure overshoot. Some clinicians attempt to increase damping by introducing a small air bubble into the tubing. Problems with this practice are:

- the patient is at risk of arterial air embolism
- the potential for retrograde flushing of the air bubbles into the cerebral circulation
- too large a bubble will overdamp the system, distort the pressure waveform and decrease systolic blood pressure
- a small air bubble increases system resonance and worsens systolic pressure overshoot

Alternative to deliberately placing air bubbles in the monitoring system, devices that increase damping without lowering natural frequency, can be added.[2] These devices eliminate wave reflection and prevent resonance in the system by impedance matching.[10,11]

The limitations of these devices include inability to adjust the monitoring system to provide the most accurate in vivo pressure recording.[12]

Clinical Measurement of Natural Frequency and Damping Coefficient

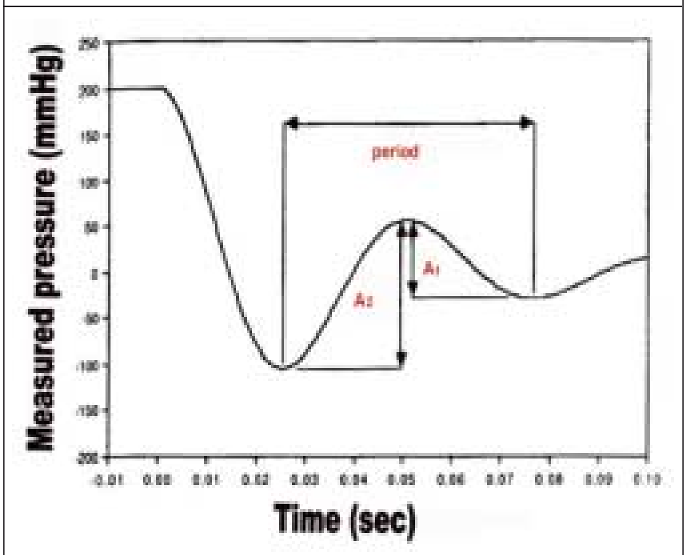
The fast flush test is the method used most often to evaluate the dynamic response of the monitoring system.[13] This method yields results comparable to standard laboratory square wave testing.[14]

Advantages of the fast flush test are:

- it can be performed at the bedside without additional equipment
- the entire monitoring system is tested, from catheter to transducer[15]

To perform the test, the fast flush valve is opened briefly and repetitively. The resulting flush artifact is examined.

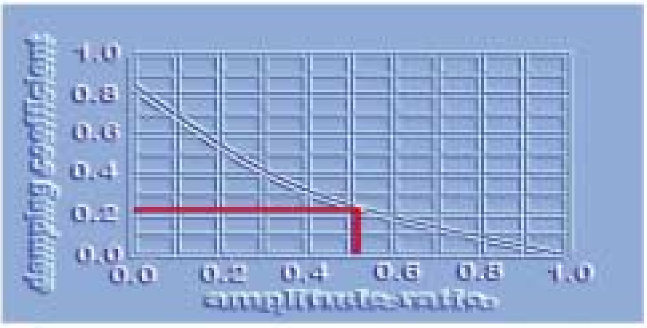
Figure 12: Square wave method for determining dynamic response



By quickly opening and closing the fast flush valve of a continuous flush mechanism, a square wave is generated. The oscillating frequency (inverse of the period) and the amplitude ratio (A1/A2) are recorded. The damping coefficient and natural frequency can then be derived. (see fig 13)

(From Lake CL (ed). Clinical Monitoring for Anesthesia and Critical Care, 2nd ed. WB Saunders, 1994: fig 3-7)

Figure 13: Two methods of determining the damping coefficient from the square wave method amplitude ratio



	0.00	0.01	0.02	0.03	0.04	0.05	0.06	0.07	0.08	0.09
0.0	1.000	0.826	0.780	0.745	0.716	0.690	0.667	0.646	0.627	0.608
0.1	0.591	0.575	0.559	0.545	0.531	0.517	0.504	0.491	0.479	0.467
0.2	0.456	0.445	0.434	0.424	0.414	0.404	0.394	0.385	0.376	0.367
0.3	0.358	0.349	0.341	0.333	0.325	0.317	0.309	0.302	0.294	0.287
0.4	0.280	0.273	0.266	0.259	0.253	0.246	0.240	0.234	0.228	0.221
0.5	0.215	0.210	0.204	0.198	0.192	0.187	0.181	0.176	0.171	0.166
0.6	0.160	0.155	0.150	0.146	0.141	0.136	0.131	0.126	0.122	0.117
0.7	0.113	0.108	0.104	0.100	0.095	0.091	0.087	0.083	0.079	0.075
0.8	0.071	0.067	0.063	0.059	0.055	0.052	0.048	0.044	0.041	0.037
0.9	0.034	0.030	0.027	0.023	0.020	0.016	0.013	0.010	0.006	0.003
1.0	0.000									

The conversion can be read from the graph or, if greater accuracy is desired, from the table. For example, an amplitude ratio of 0.51 yields a damping coefficient of 0.21

(From Lake CL (ed). Clinical Monitoring for Anesthesia and Critical Care, 2nd ed. WB Saunders, 1994: fig 3-8)

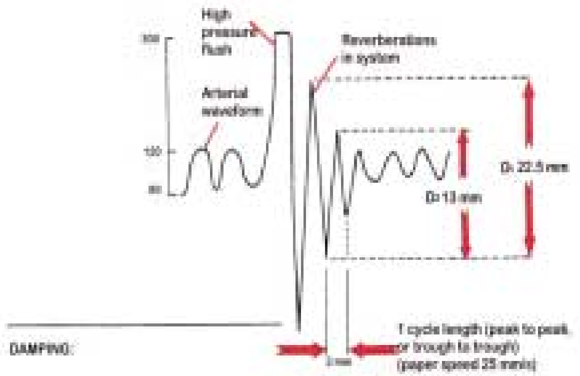
Practical consideration for the optimal set-up

- no device in itself can substitute careful set-up of equipment
- high frequency response and sensitivity are mutually exclusive
- choose a system according to what should be measured
  - CVP: low frequency response (10Hz) but high sensitivity
  - arterial pressure: higher frequency response (20Hz)
  - dP/dt max: very high frequency response (>30Hz)
- choose wide bore arterial cannulas (minimum 20 SWG in adults) - small enough to minimize arterial damage, yet large enough to improve the frequency response of the system
- use wide bore taps and as few stopcocks as possible
- choose a short, wide (>1-2mm) and rigid catheter
- undamped natural frequency (fo) = $^1/_2\ \pi \sqrt{\ \ }\ ^s/_m$
- rid the system of air bubbles, which will alter M (fluid velocity)
- prevent blood clots by regularly flushing with heparin; clots decrease the lumen, increase velocity and M, therefore decreasing fo and increasing damping (through increased resistance $R = 8\eta L/\pi r^4$)
- do regular zero point calibration
- ensure optimal damping
- systolic pressure is overestimated by an underdamped system
- diastolic pressure is less sensitive to sub-optimal dynamic response, but is underestimated by underdamped systems and overestimated by overdamped systems
- mean pressure is least affected by the dynamic response of the measuring system
- most pressure monitoring systems have dynamic response limitations, thus it is expected that direct measurement of systolic arterial pressure often exceed indirect noninvasive measurement, simply because of underdamping and resonance[12]
- the display system should have a higher frequency response than

Figure 14: Damping and natural frequency of a transducer system can be determined by a high-pressure flush test

Figure 14.1: High-pressure flush test

- the distance between successive oscillations should be identical, because this is the natural frequency
- the lighter the oscillation cycles, the higher the natural frequency
- choose a fast recording speed, examine several flush cycles and calculate an average



$$\text{Amplitude ratio} = \frac{D_2\ (\text{mm})}{D_1\ (\text{mm})} = \frac{13}{22.5} = 0.58$$

$$\text{Damping coefficient } \beta = \sqrt{\frac{\left(\ln \frac{D_2}{D_1}\right)^2}{\pi^2 + \left(\ln \frac{D_2}{D_1}\right)^2}} = 0.17$$

Amplitude ratio (D_2/D_1)	Damping coefficient
.9	.034
.8	.071
.7	.113
.6	.160
.5	.215
.4	.280
.3	.358
.2	.456
.1	.591

- amplitude ratio indicates how quickly the system comes to rest˝
- a low amplitude ratio (system comes to rest quickly) corresponds to a high damping coefficient
- a high amplitude ratio (system resonates) corresponds to a low damping coefficient

NATURAL FREQUENCY:

$$\text{Natural frequency} = fn = \frac{1}{2\pi}\sqrt{\frac{\pi D^2\, \Delta P}{4\rho\, L \Delta V}} = \frac{\text{Paper speed (mm/sec)}}{\text{Length of 1 cycle (mm)}}$$

$$= \frac{25 \text{ mm/sec}}{2 \text{ mm}} = 12.5 \text{ HZ}$$

D = Internal diameter of tubing

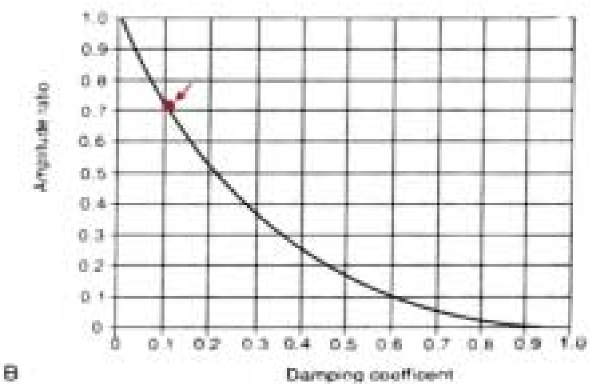
ρ = Density of blood

L = Length of tubing

$\frac{\Delta P}{\Delta V}$ = Compliance (stiffness) of system

(Modified from Blitt CD (ed): Monitoring in Anesthesia and Critical Care Medicine, 2nd ed. Churchill Livingstone, New York, 1990, p110)

Figure 14.2:The damping coefficient can be calculated or determined graphically from the measured amplitude ratio



(From Mark JB: Atlas of Cardiovascular Monitoring. New York, Churchill Livingstone, 1998: fig 9-10)

the catheter transducer system e.g. a heated stylus recorder for arterial pressure, but a photographic recorder for dP/dt max (LVEDP)

Conclusion

The quality of the transduced arterial pressure waveform depends on the dynamic characteristics of the catheter-tubing-transducer system.[2] We strive to understand the physiological and physical limitations of these measurements, judge the potential for error and, equipped with practical considerations for a more optimal set up, intervene appropriately in the face of uncertainty.

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