

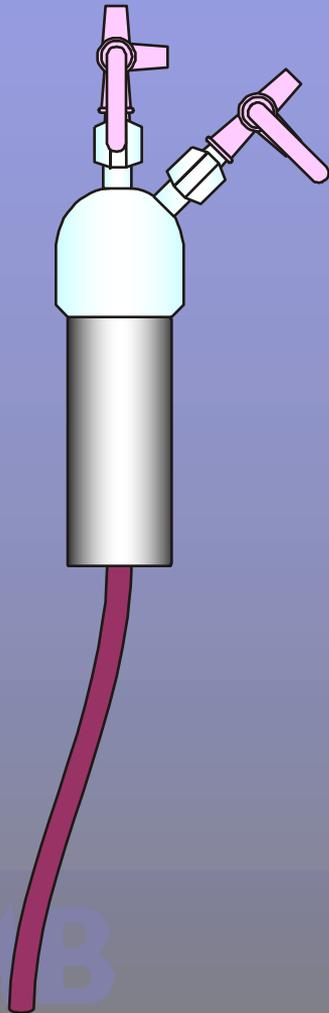
Medición de la presión



Transductor para medir presión por medio de un catéter de columna fluida. El catéter se conecta al extremo de una de las llaves incorporadas en el casquete superior. En el interior del casquete el fluido está en contacto con una membrana deformable, sensible a los cambios mecánicos, que son transformados en cambios de tensión eléctrica.

La lectura de la presión está “contaminada” por la longitud del catéter, la presencia de burbujas y las características elásticas de la pared del mismo catéter.

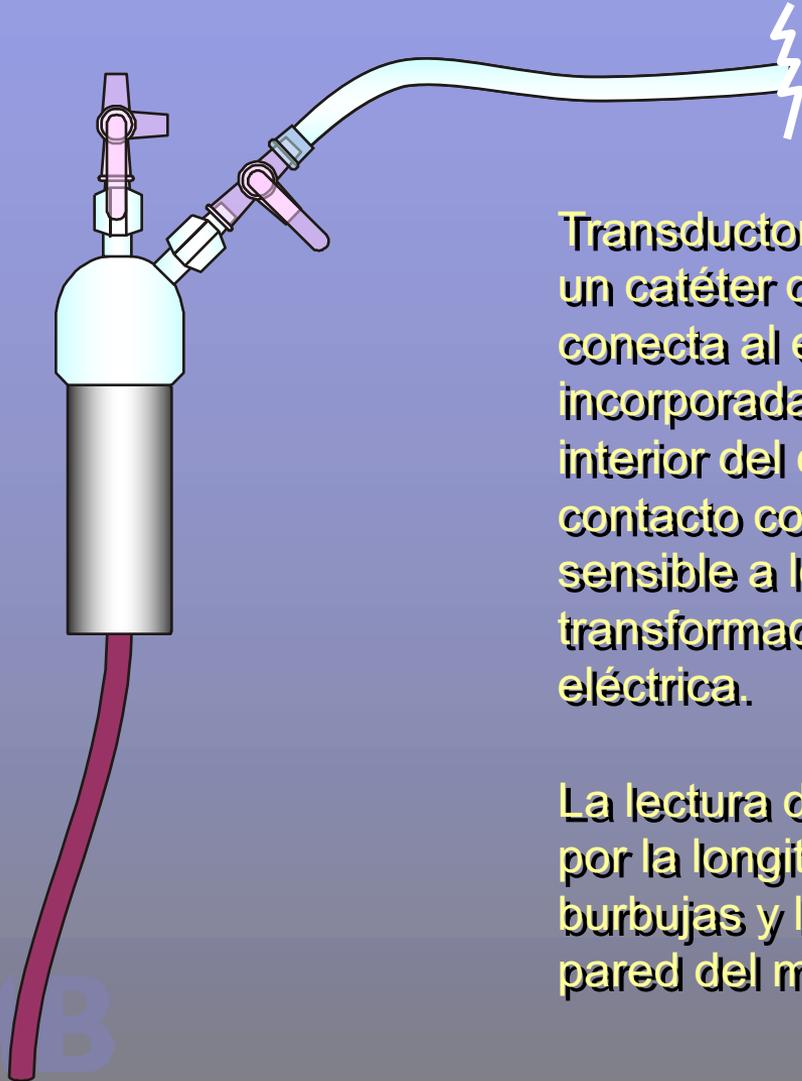
Medición de la presión



Transductor para medir presión por medio de un catéter de columna fluida. El catéter se conecta al extremo de una de las llaves incorporadas en el casquete superior. En el interior del casquete el fluido está en contacto con una membrana deformable, sensible a los cambios mecánicos, que son transformados en cambios de tensión eléctrica.

La lectura de la presión está “contaminada” por la longitud del catéter, la presencia de burbujas y las características elásticas de la pared del mismo catéter.

Medición de la presión

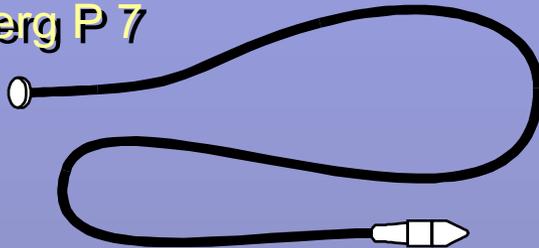


Transductor para medir presión por medio de un catéter de columna fluida. El catéter se conecta al extremo de una de las llaves incorporadas en el casquete superior. En el interior del casquete el fluido está en contacto con una membrana deformable, sensible a los cambios mecánicos, que son transformados en cambios de tensión eléctrica.

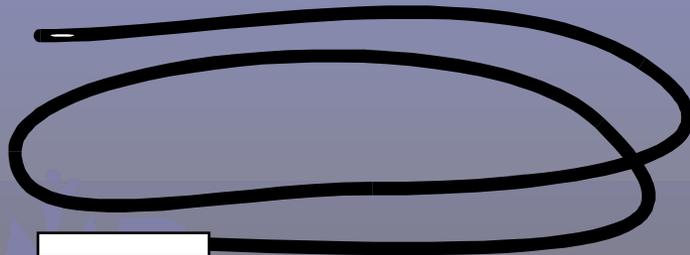
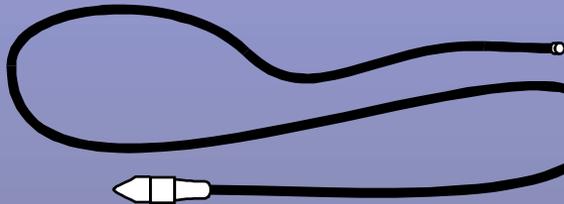
La lectura de la presión está “contaminada” por la longitud del catéter, la presencia de burbujas y las características elásticas de la pared del mismo catéter.

Medición de la presión

Konigsberg P 7



Konigsberg P 2.5



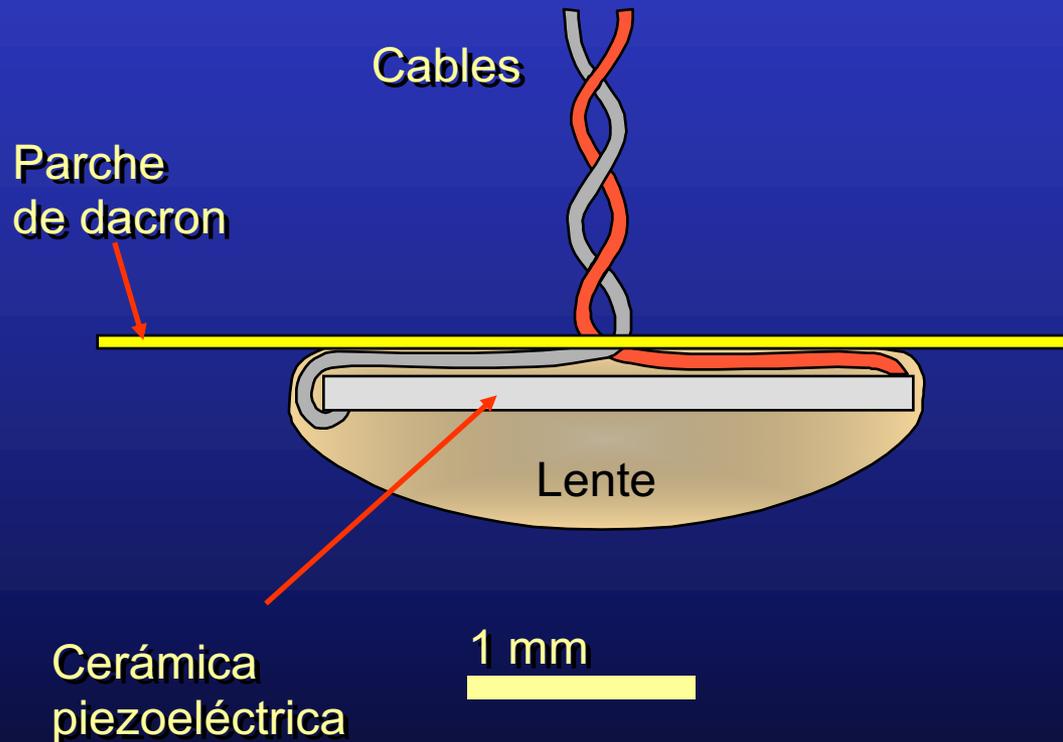
Catéter Millar

Los microtransductores de estado sólido (Konigsberg y Millar) para lectura directa intravascular proporcionan una señal de alta fidelidad.

Los microtransductores de presión Konigsberg están diseñados para uso en animales y puedan quedar implantados en forma crónica por varios meses.

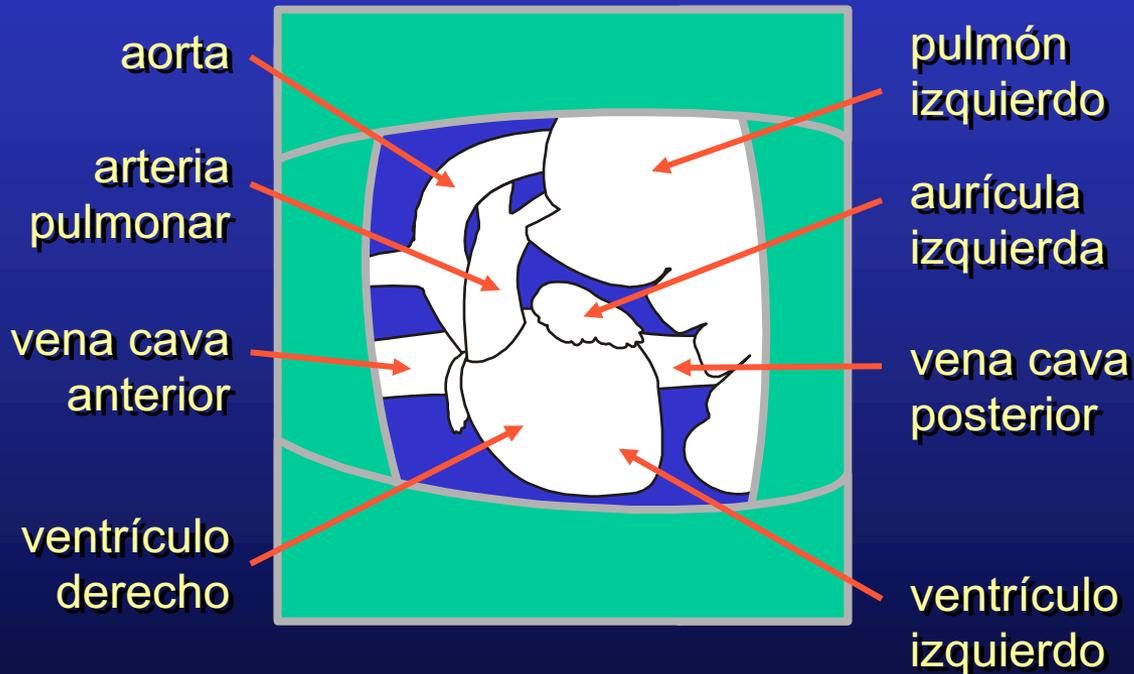
Los catéteres tipo “Millar” son implantados a través de la canalización de un vaso y **NO PUEDEN** quedar permanentes. Éstos pueden usarse en humanos. Algunos modelos tienen dos transductores separados por 10 cm.

Medición del diámetro



Esquema de un microcristal usado para medir diámetro vascular. En todos los casos la cerámica piezoeléctrica se halla soldada a un cable bipolar y recubierta por una lentilla de resina poliuretánica. El aparato que activa el cristal emisor y recibe la señal captada por el receptor se denomina **SONOMICRÓMETRO**, el cual permite calibrar cada señal en milímetros lineales en un rango de 2 a 123 mm.

Instrumentación quirúrgica



Esquema mostrando los órganos internos del tórax de un animal luego de realizada una toracotomía a nivel del quinto espacio intercostal. El pulmón izquierdo ha sido desplazado hacia el diafragma

Instrumentación quirúrgica

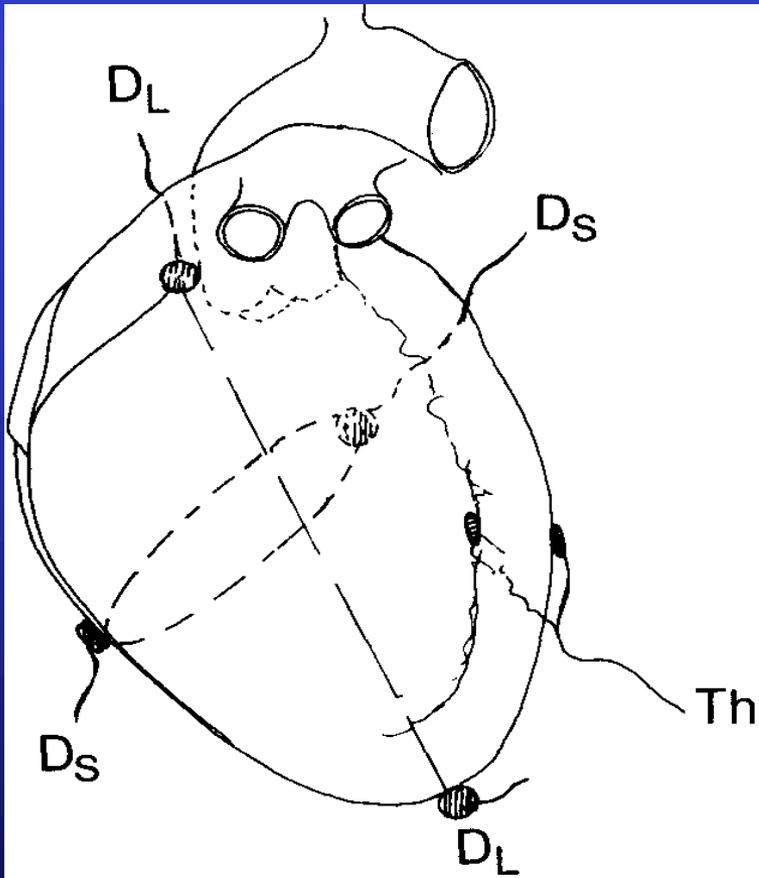
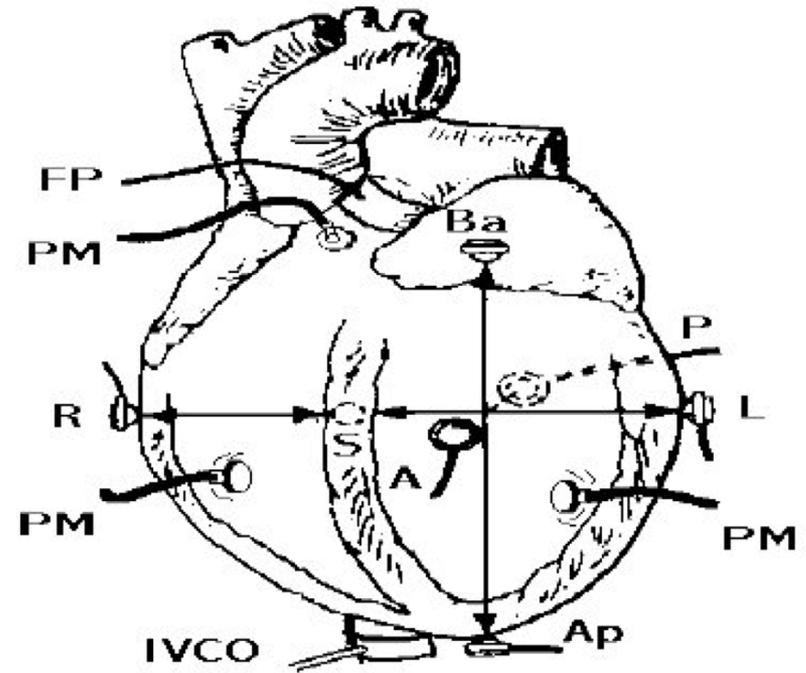
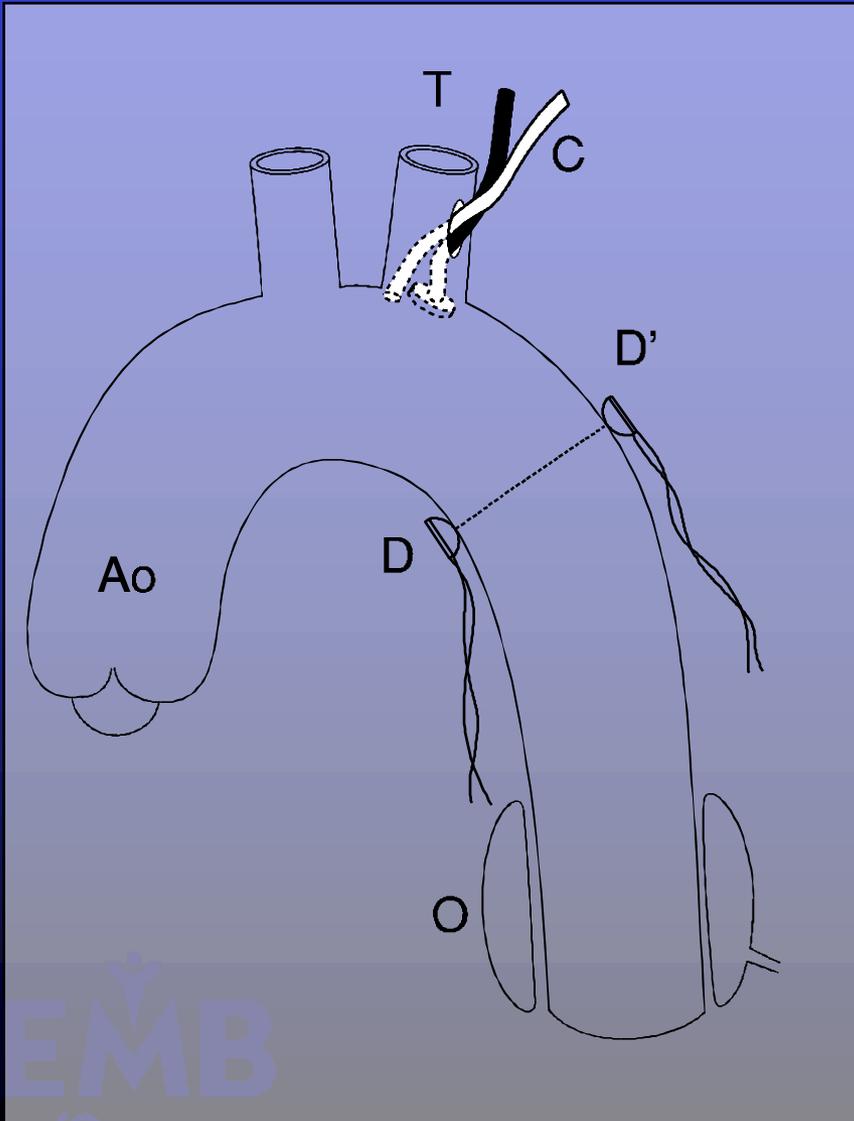


FIGURE 1. Schematic representation of the orientation of the ultrason-



Instrumentación quirúrgica



Instrumentación quirúrgica de la aorta torácica descendente.

Ao: aorta.

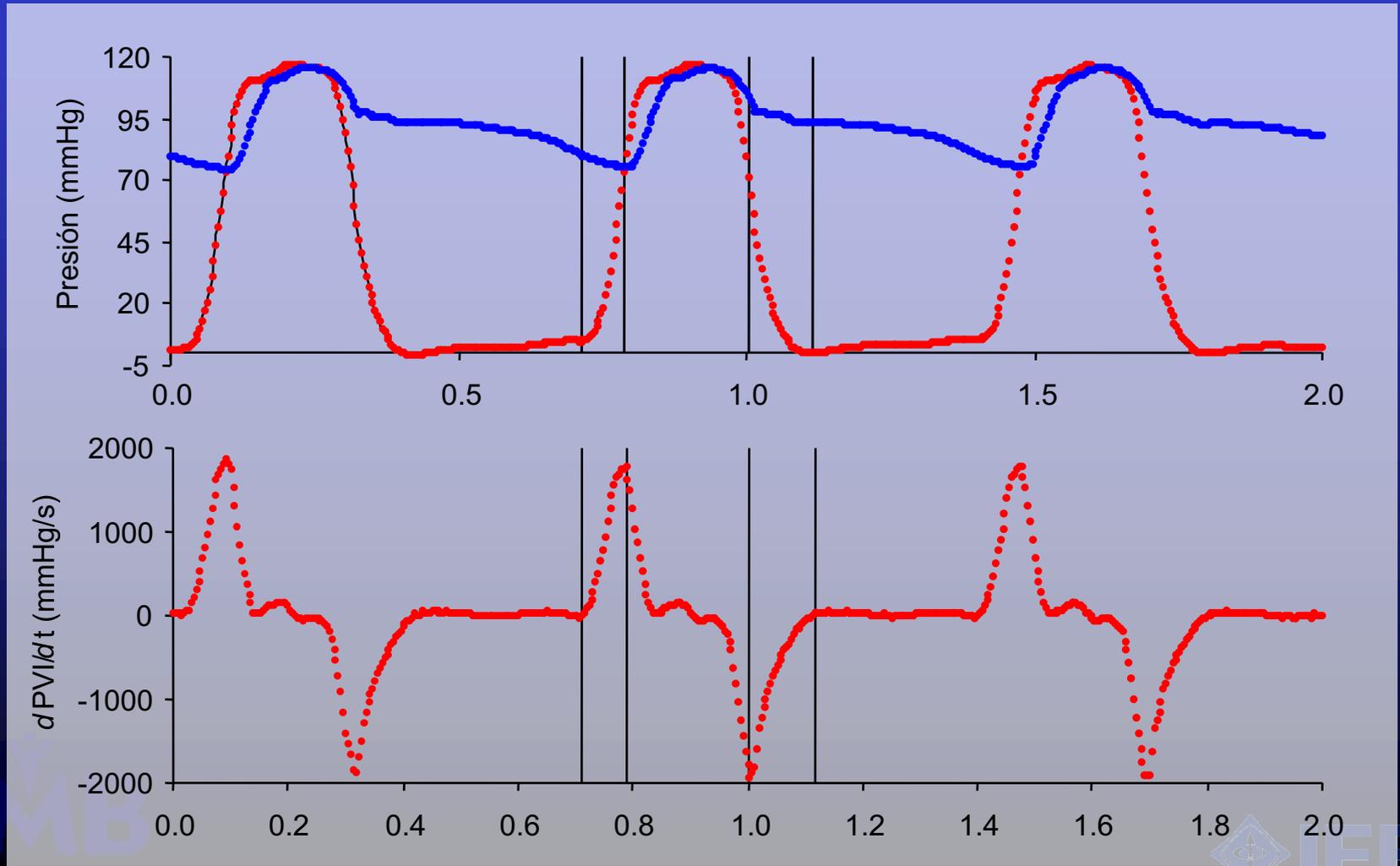
T: transductor de presión de estado sólido (Konigsberg P7).

C: catéter de columna fluida.

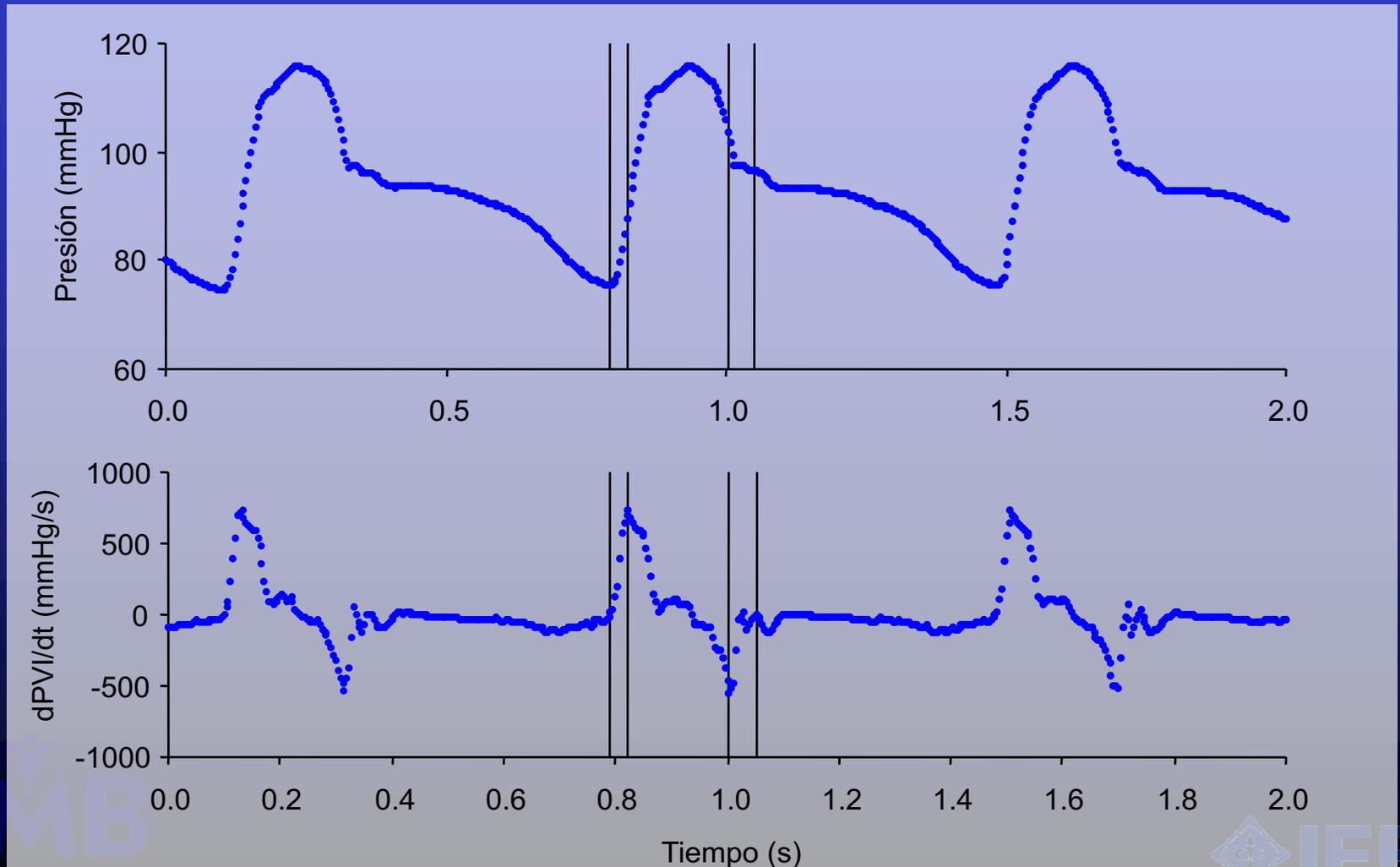
D-D': cristales piezoeléctricos ultrasónicos para medir el diámetro externo de la arteria.

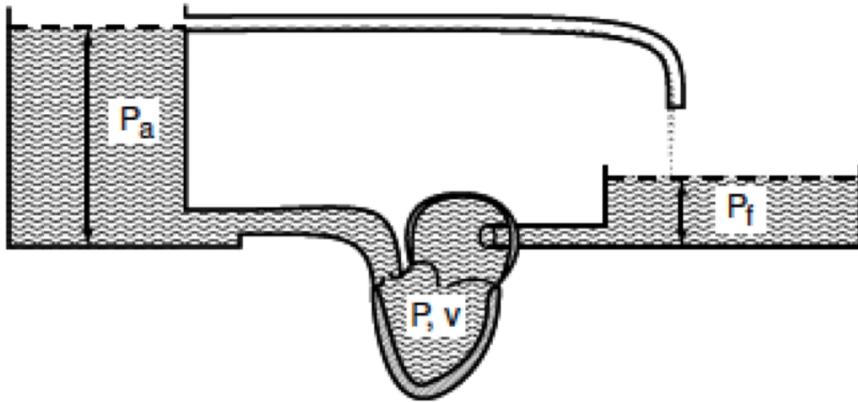
O: ocluidor neumático.

Análisis temporal

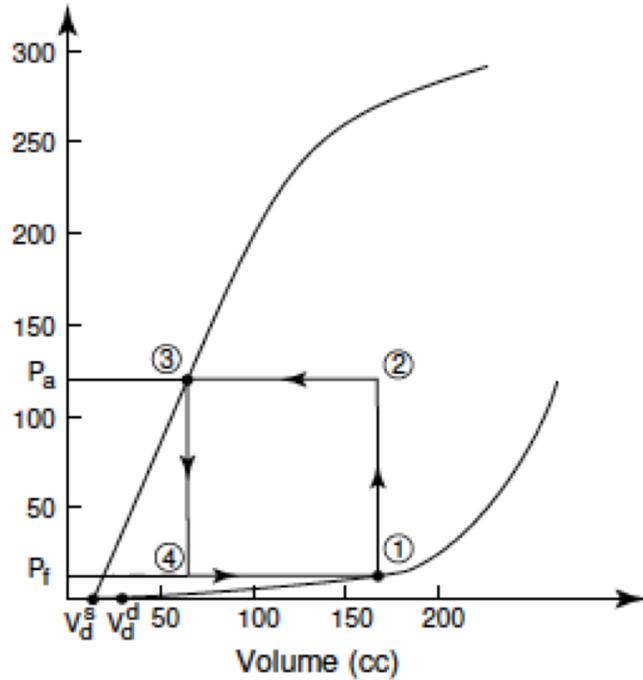


Análisis temporal

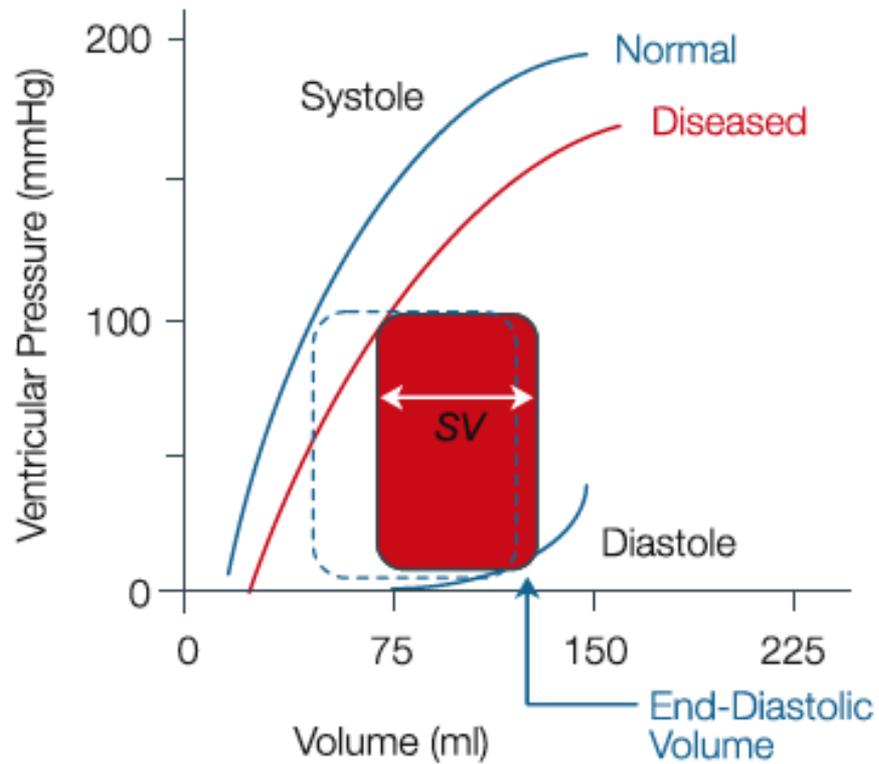




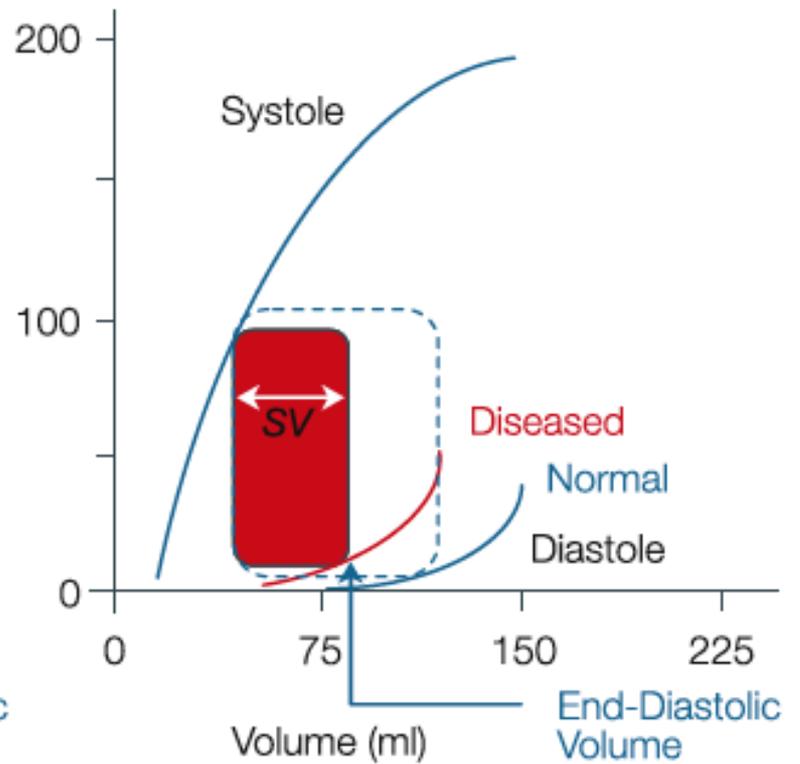
Pressure (mmHg)



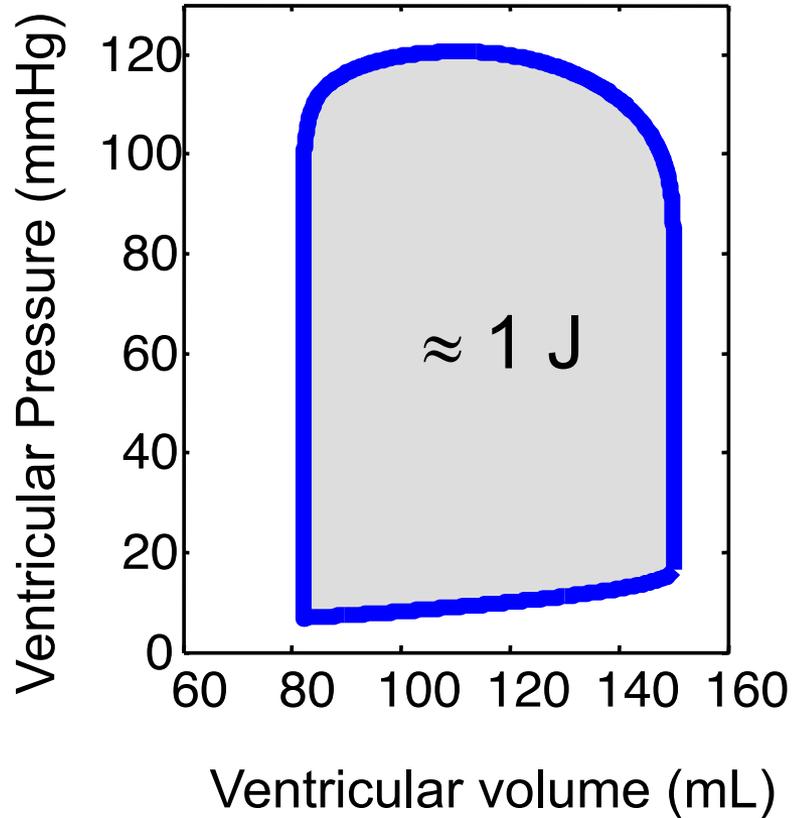
SYSTOLIC DYSFUNCTION



DIASTOLIC DYSFUNCTION



Ventricular work

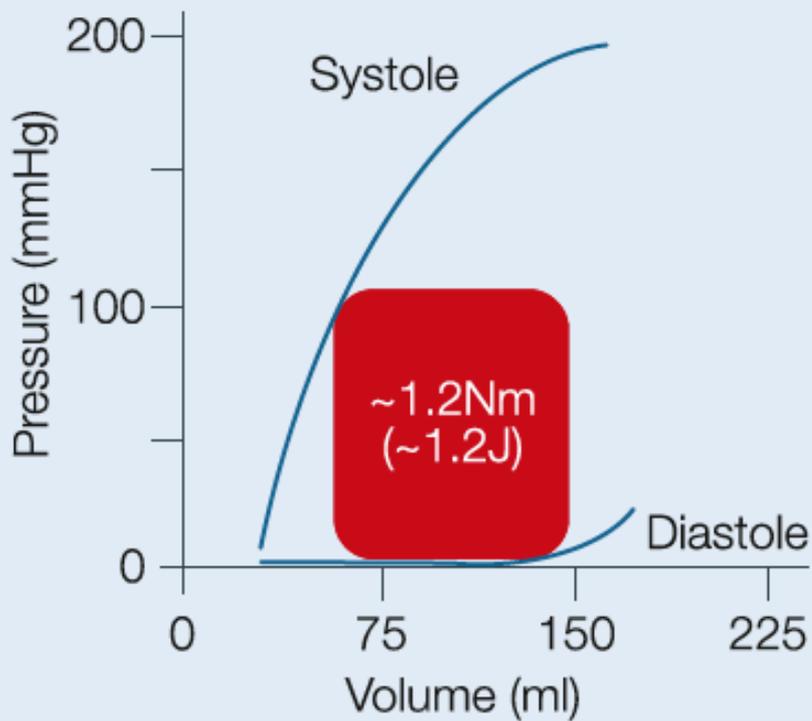


$$\int P(t) dV$$

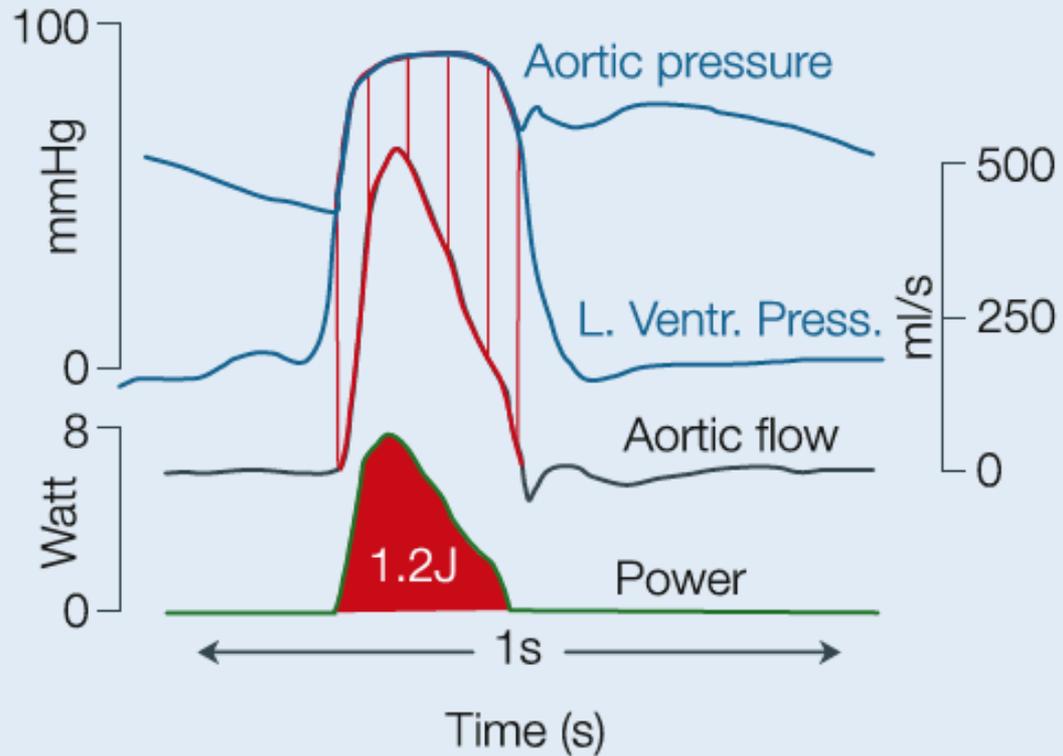
P : Ventricular pressure

V : Ventricular volume

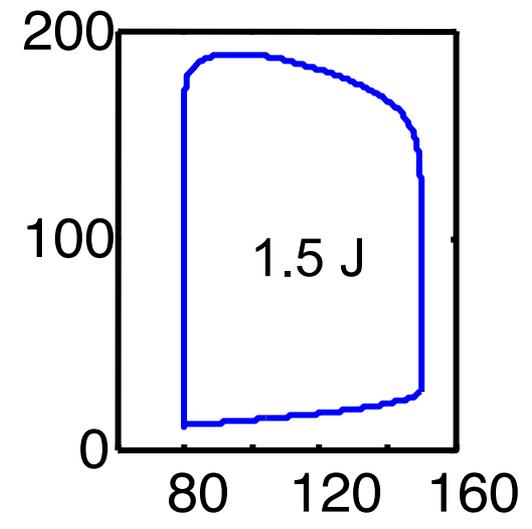
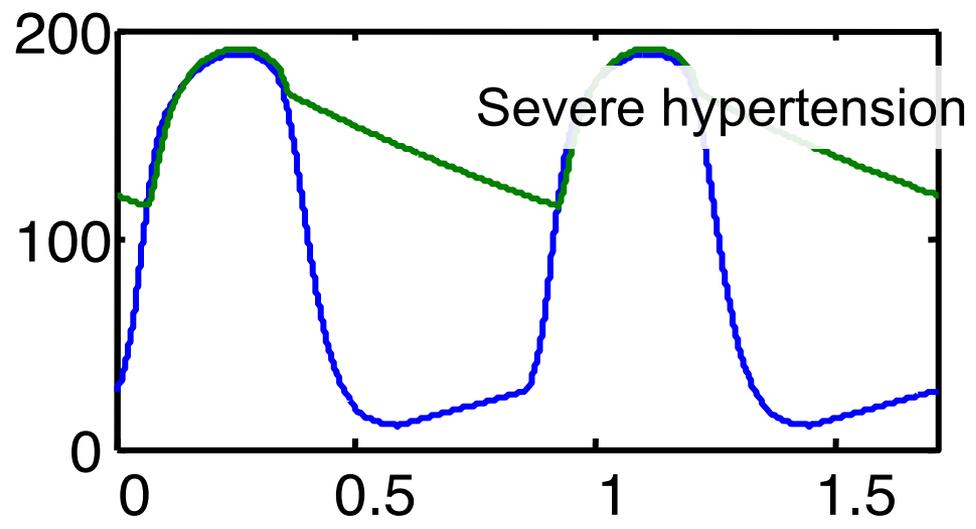
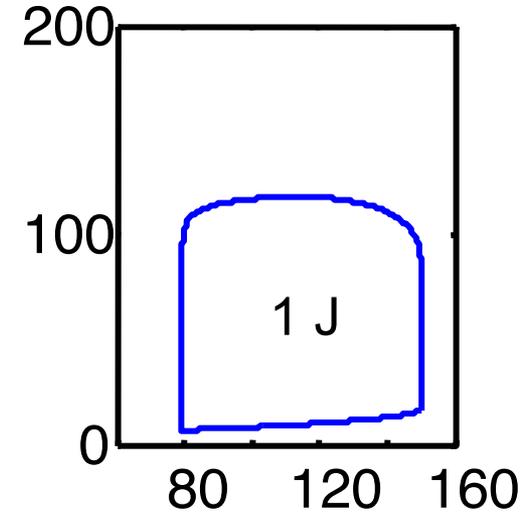
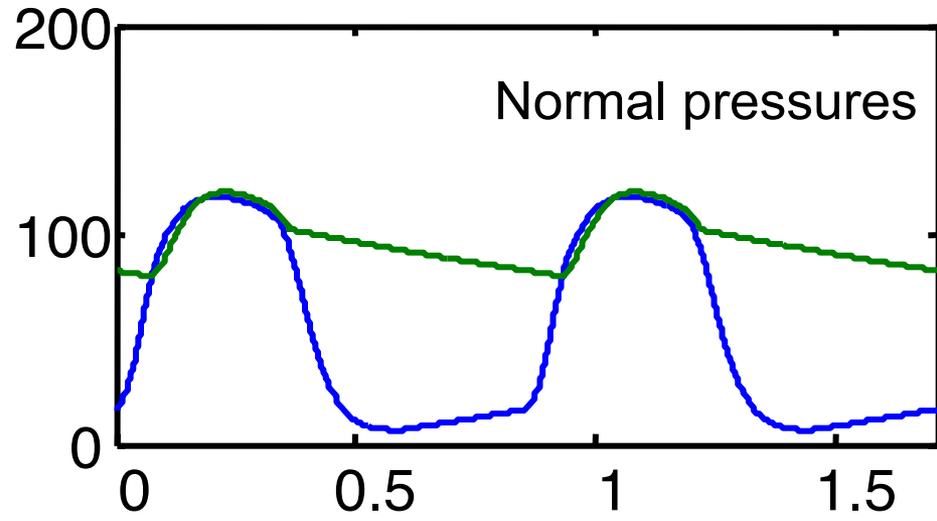
Area within P - V loop equals external work per beat



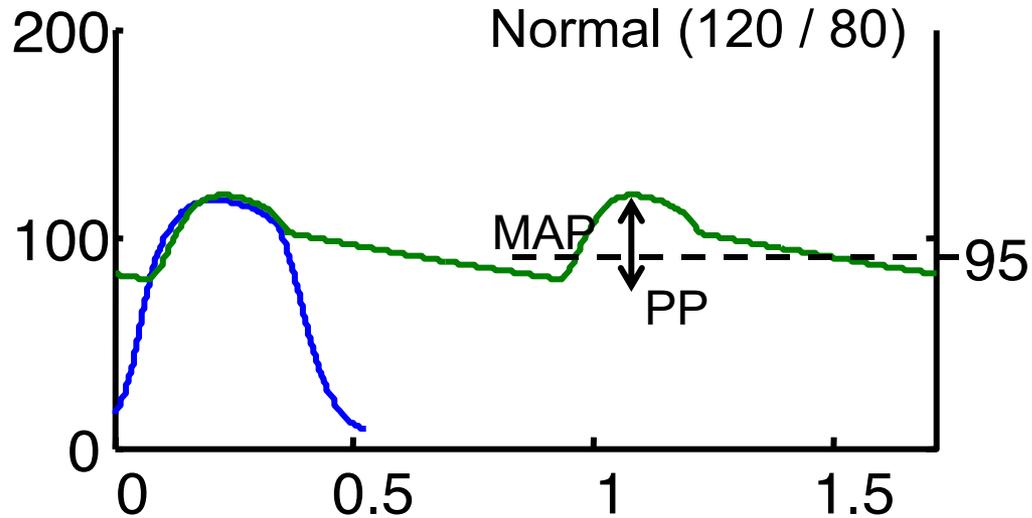
The product of P and Q , equals external power as function of time



Effect of systemic hypertension

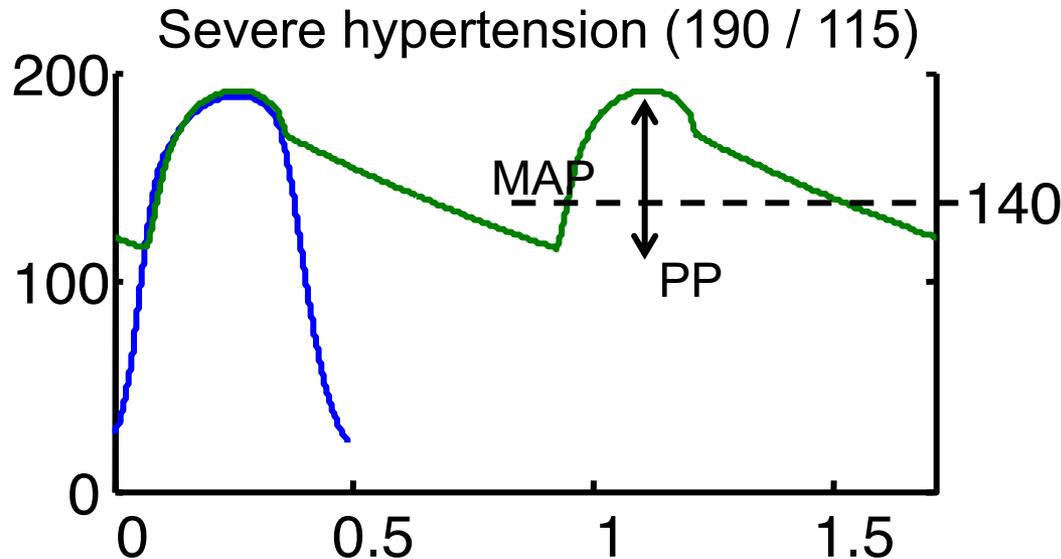


Hypertension and aortic pressure



$$R = 1.1 \text{ mmHg.s / mL}$$
$$C = 2.0 \text{ mL / mmHg}$$

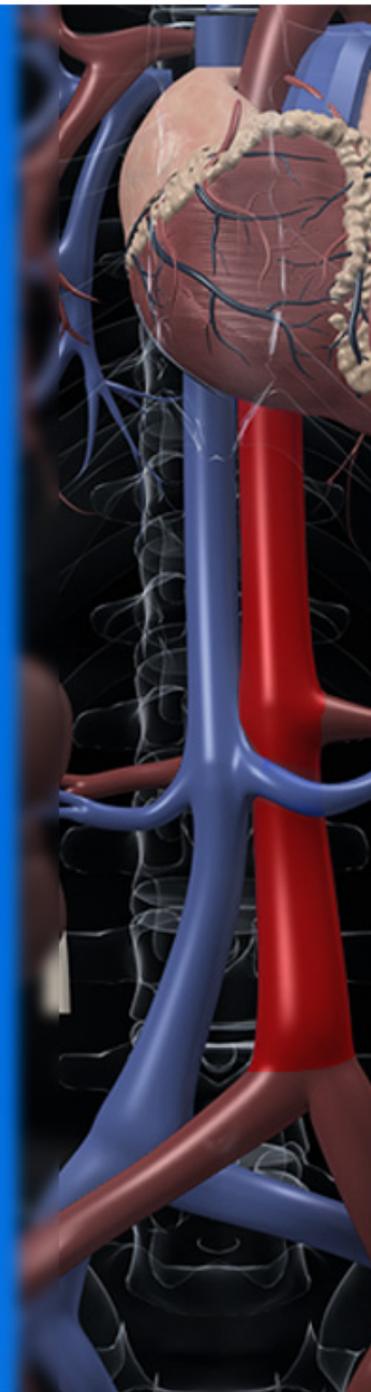
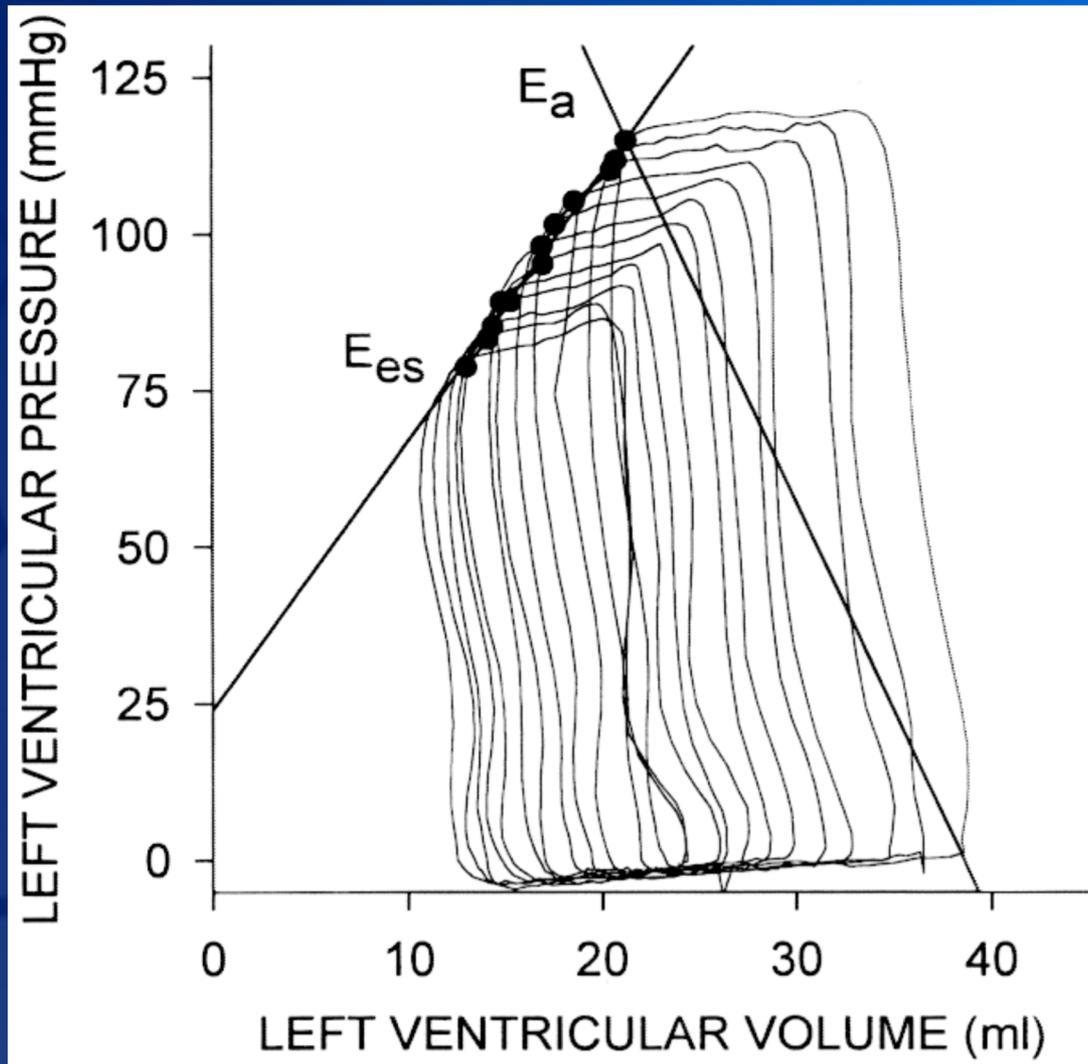
$$\text{MAP} = 95 \text{ mmHg}$$
$$\text{PP} = 40 \text{ mmHg}$$



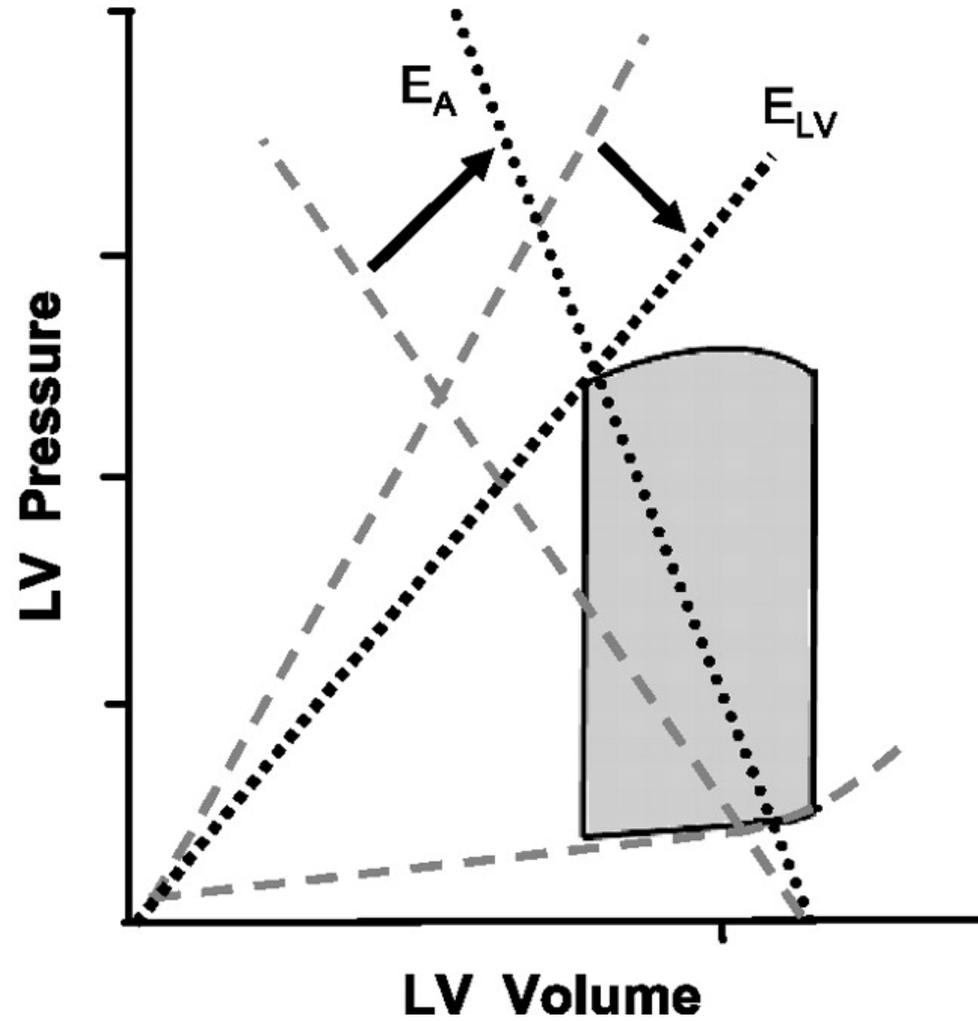
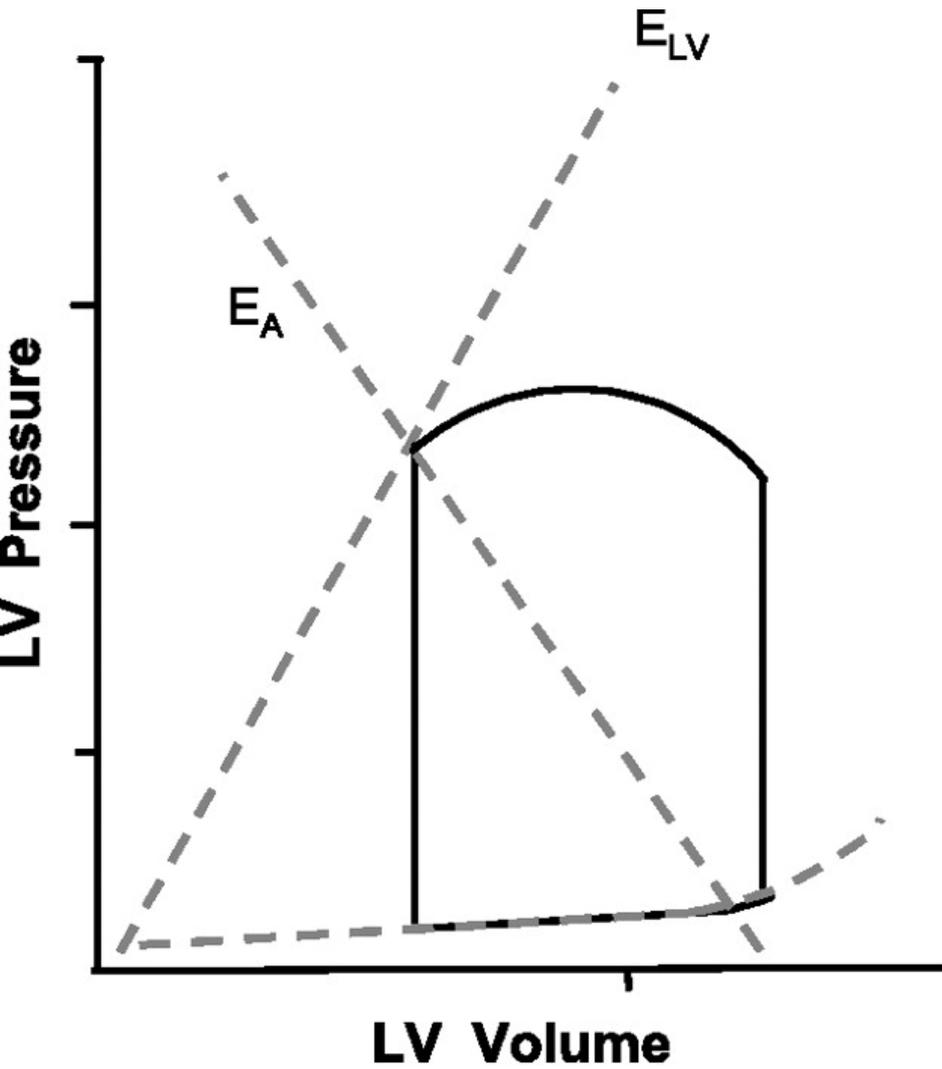
$$R = 1.7 \text{ mmHg.s / mL}$$
$$C = 0.8 \text{ mL / mmHg}$$

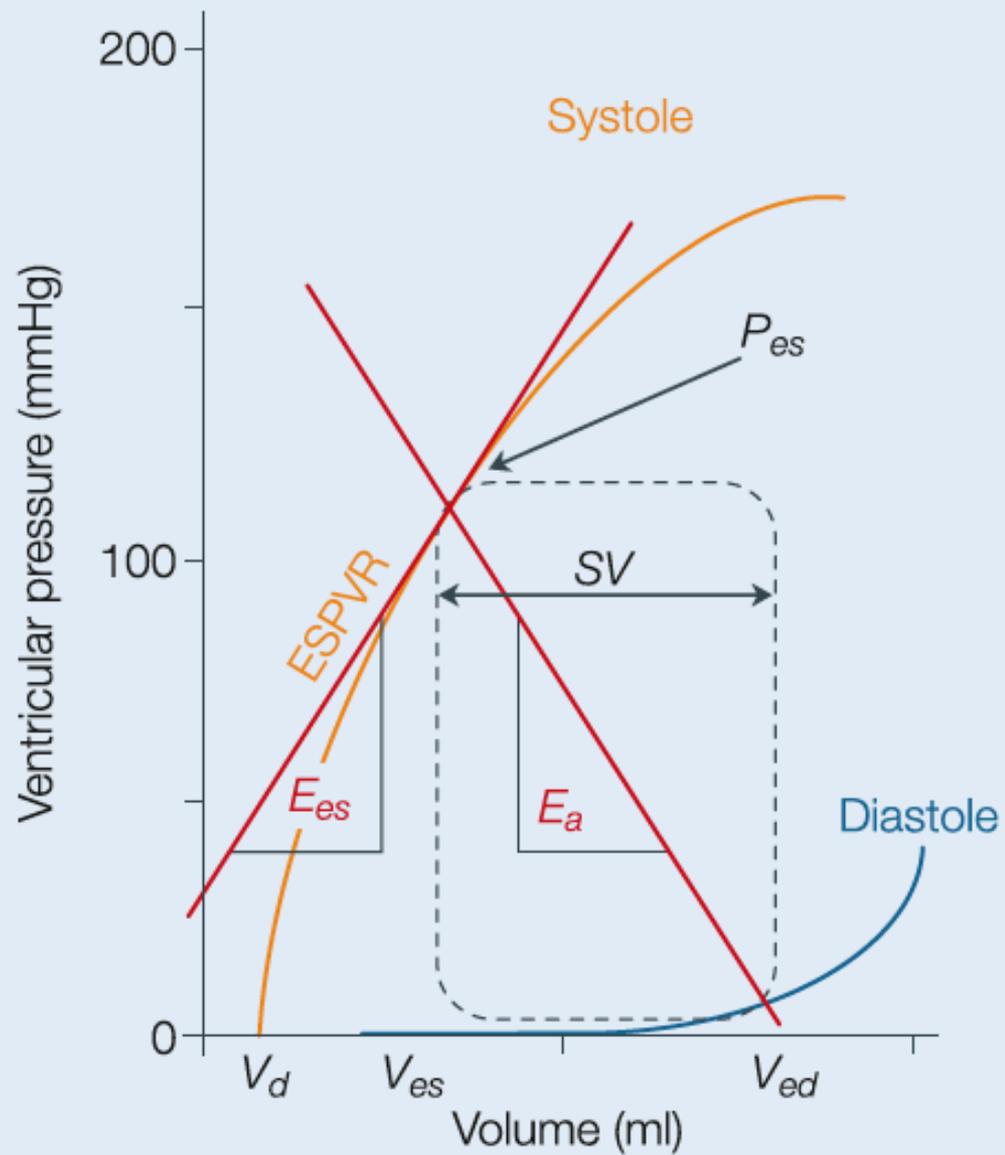
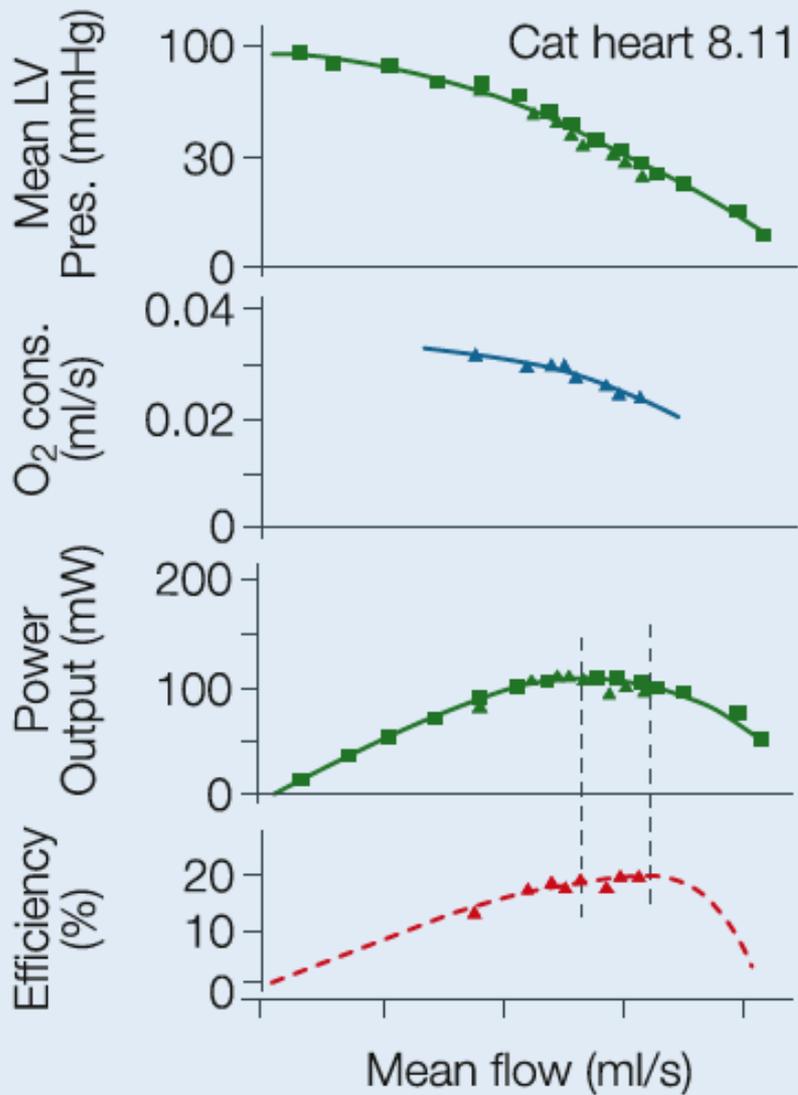
$$\text{MAP} = 140 \text{ mmHg}$$
$$\text{PP} = 75 \text{ mmHg}$$

ELASTANCIA de FDS

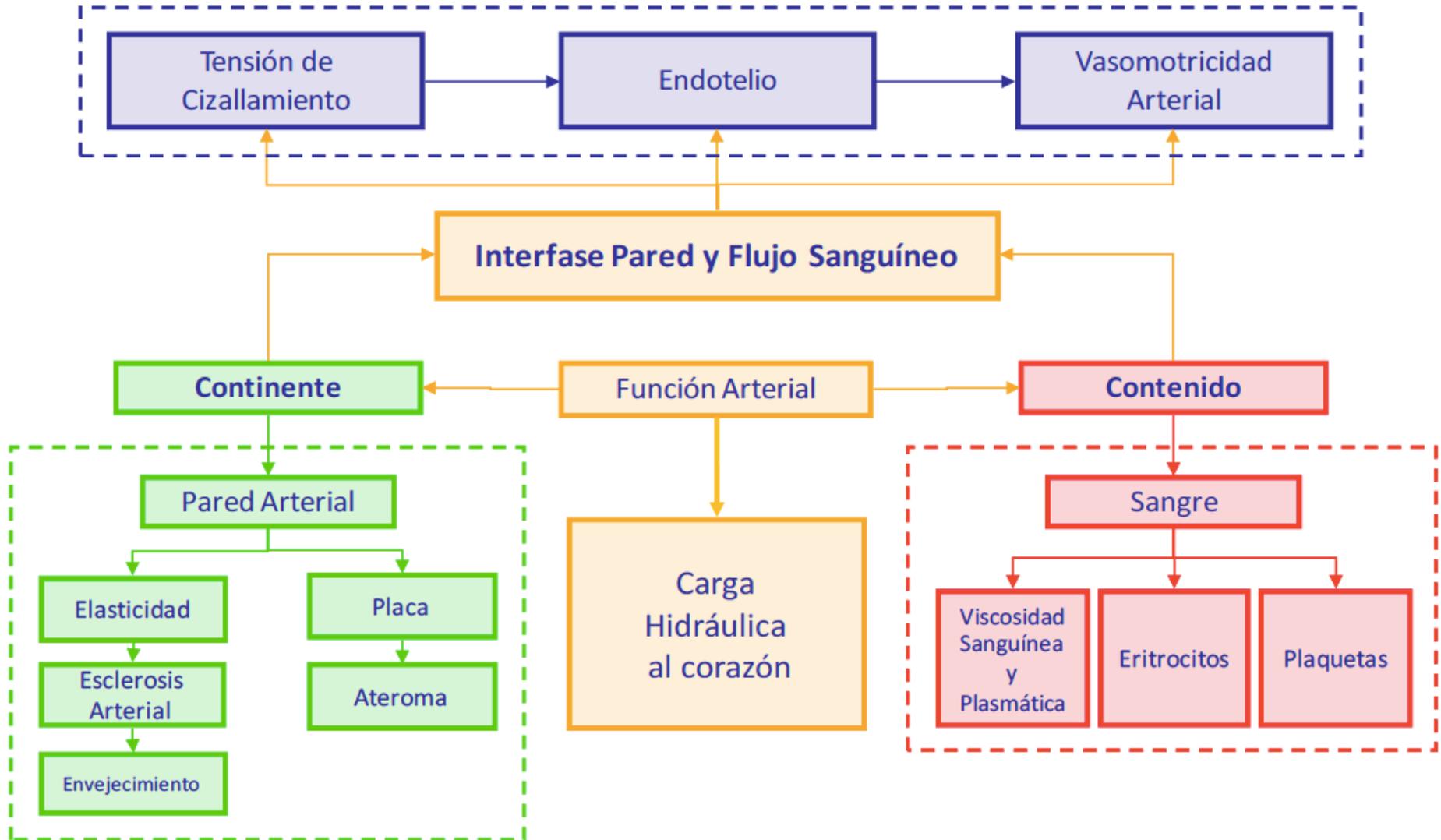


ACOPLAMIENTO





ELASTANCIA ARTERIAL?

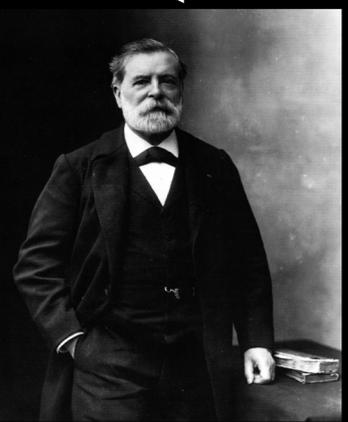
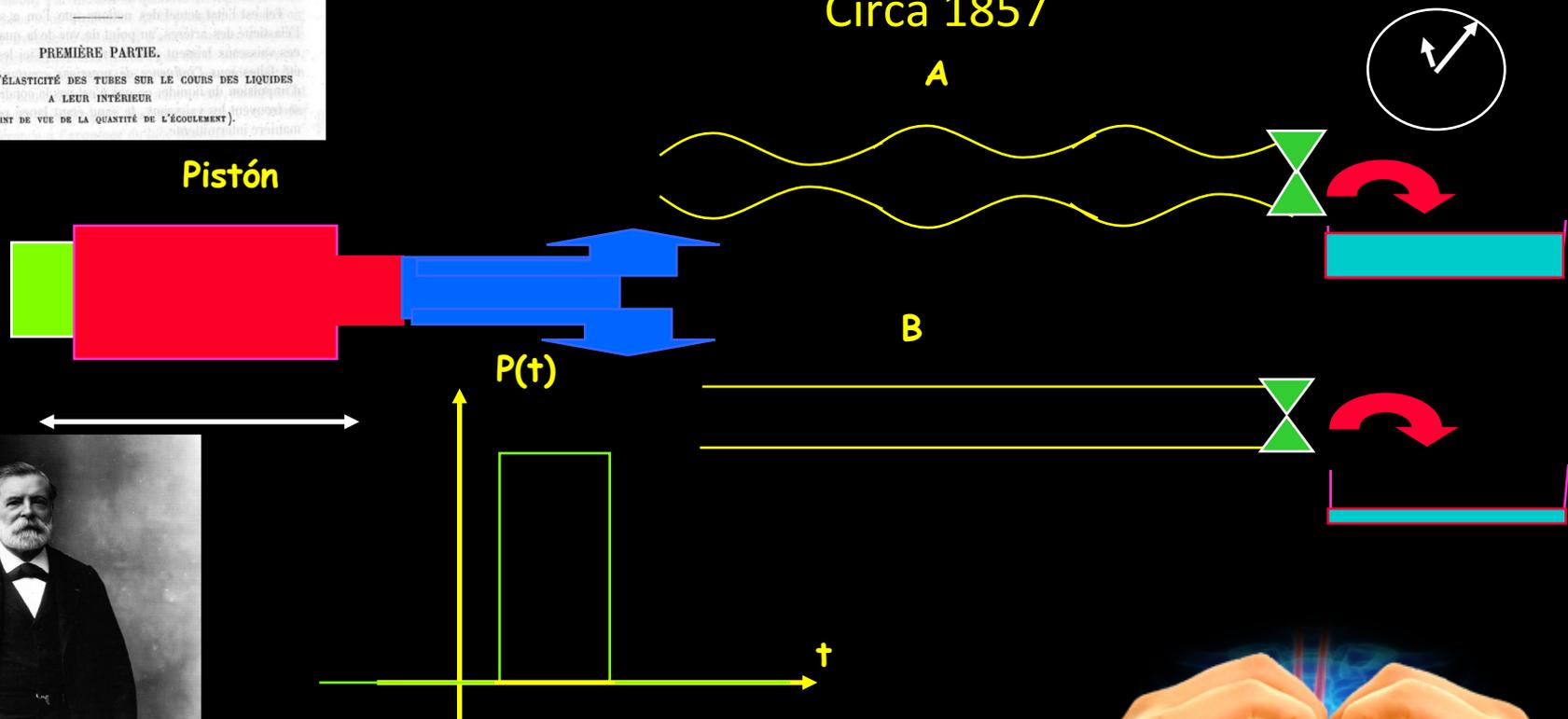


RECHERCHES HYDRAULIQUES
 SUR
 LA CIRCULATION DU SANG,
 Par M. J. MAREY,
 Interne à l'Hôpital Cochin.
 PREMIÈRE PARTIE.
 INFLUENCE DE L'ÉLASTICITÉ DES TUBES SUR LE COURS DES LIQUIDES
 A LEUR INTÉRIEUR
 (AU POINT DE VUE DE LA QUANTITÉ DE L'ÉCOULEMENT).

ETIENNE-JULES MAREY

Recherches Hydrauliques sur la circulation du sang

Circa 1857

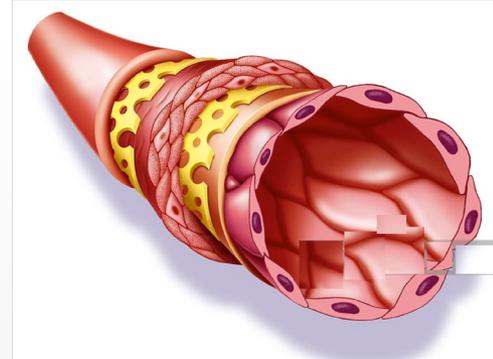
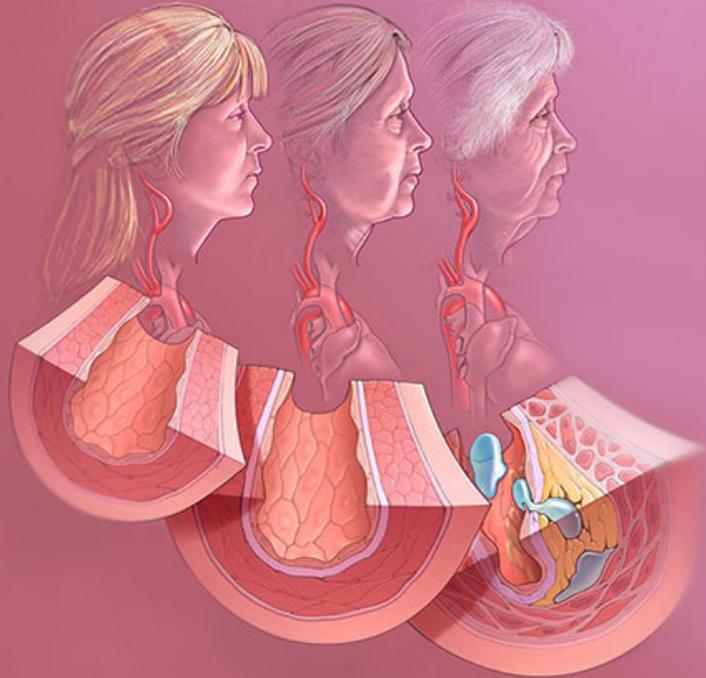


Sobre una Idea Dr. Carlos Taquini, Buenos Aires



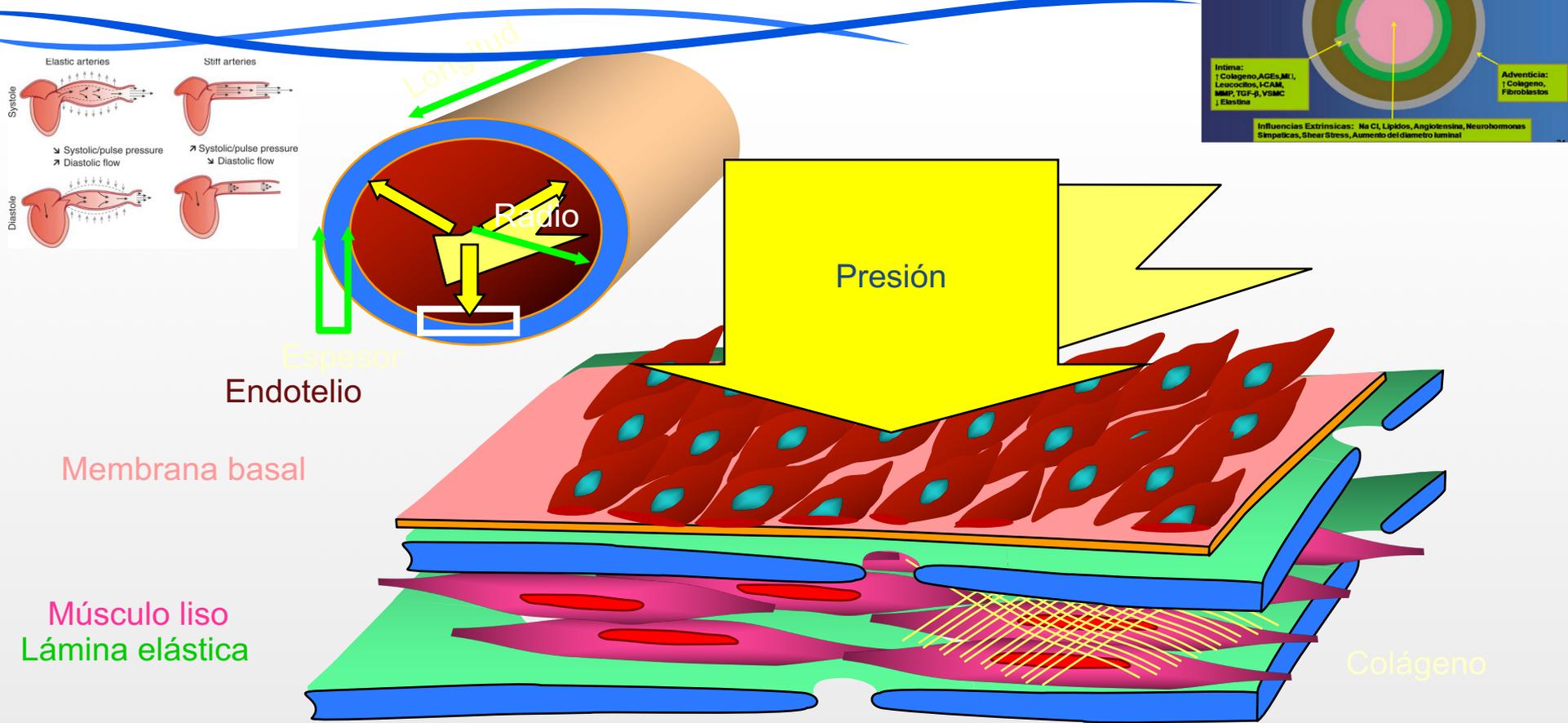
CONTINENTE

La rigidez de la pared vascular tiene importante valor predictivo y pronostico en relación a la aparición de episodios CV, incluso tal vez superior al de los FRCV clásicos.

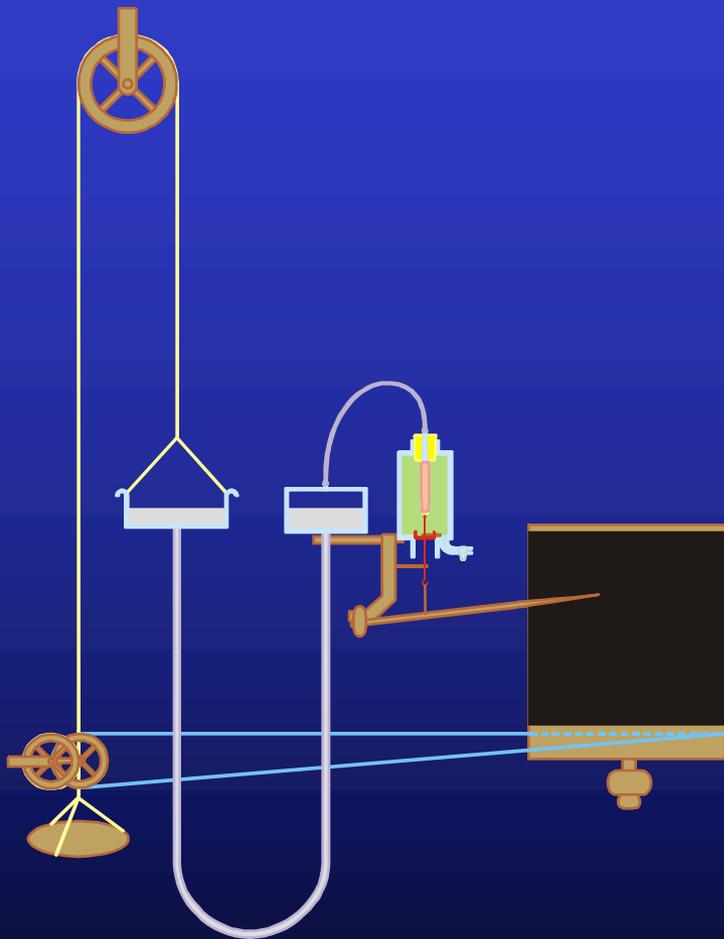


PARED VASCULAR: STIFFNESS

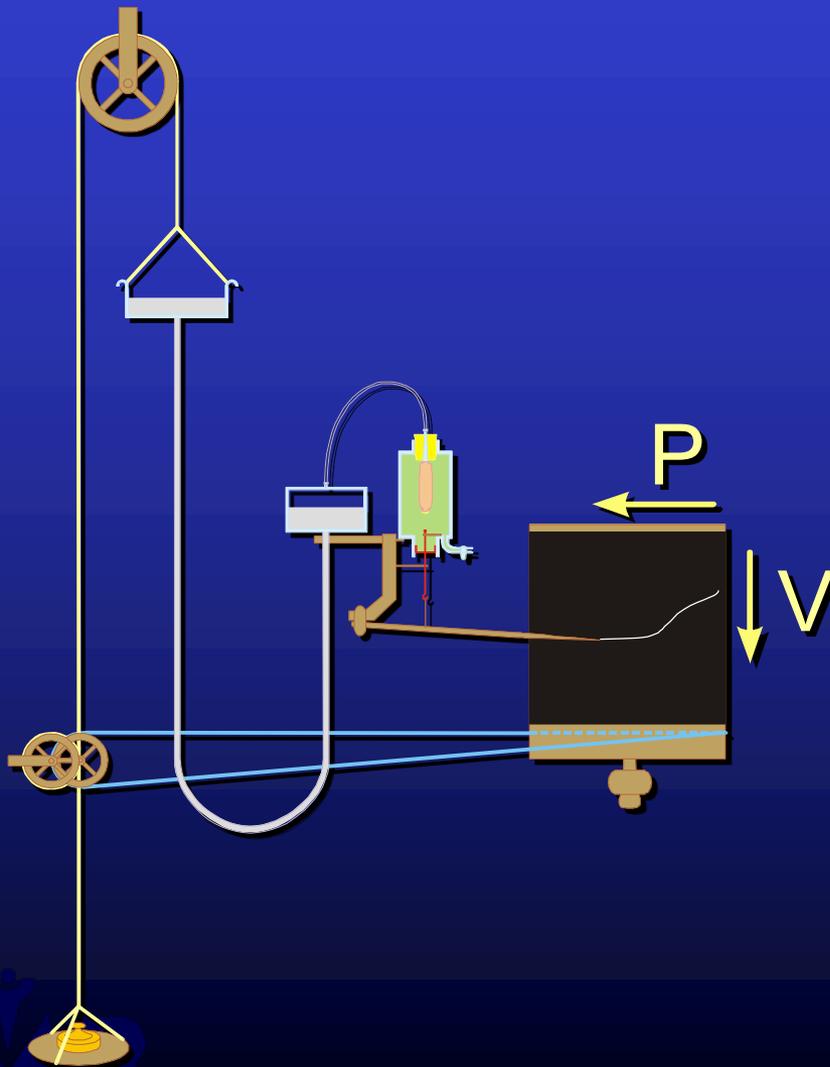
Mecanismos de degeneración y rotura de fibras elásticas con su reemplazamiento por colágeno la hipertrofia de la intima, la necrosis del musculo liso medial y fenomenos de fibrosis e inflamacion.



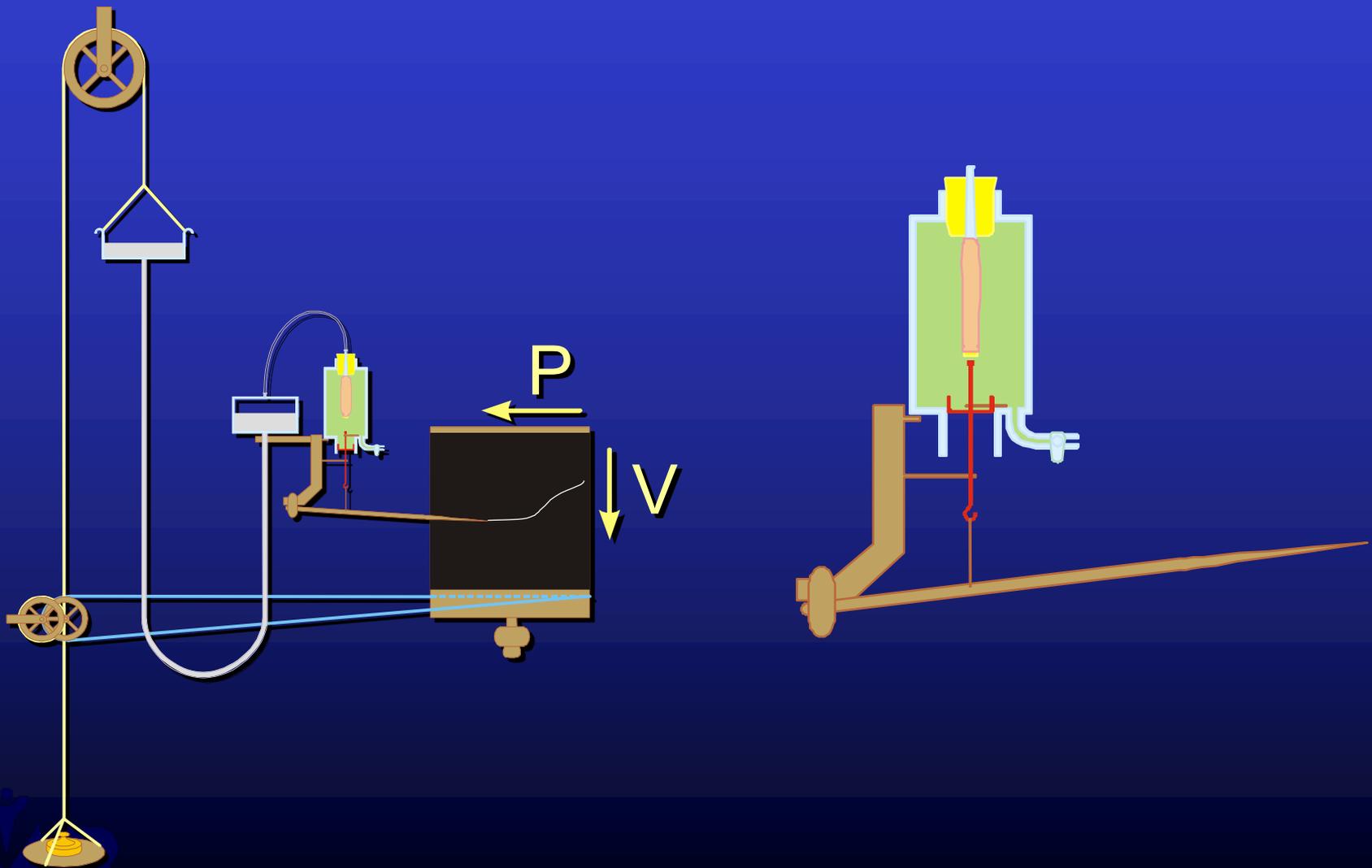
Roy, 1881



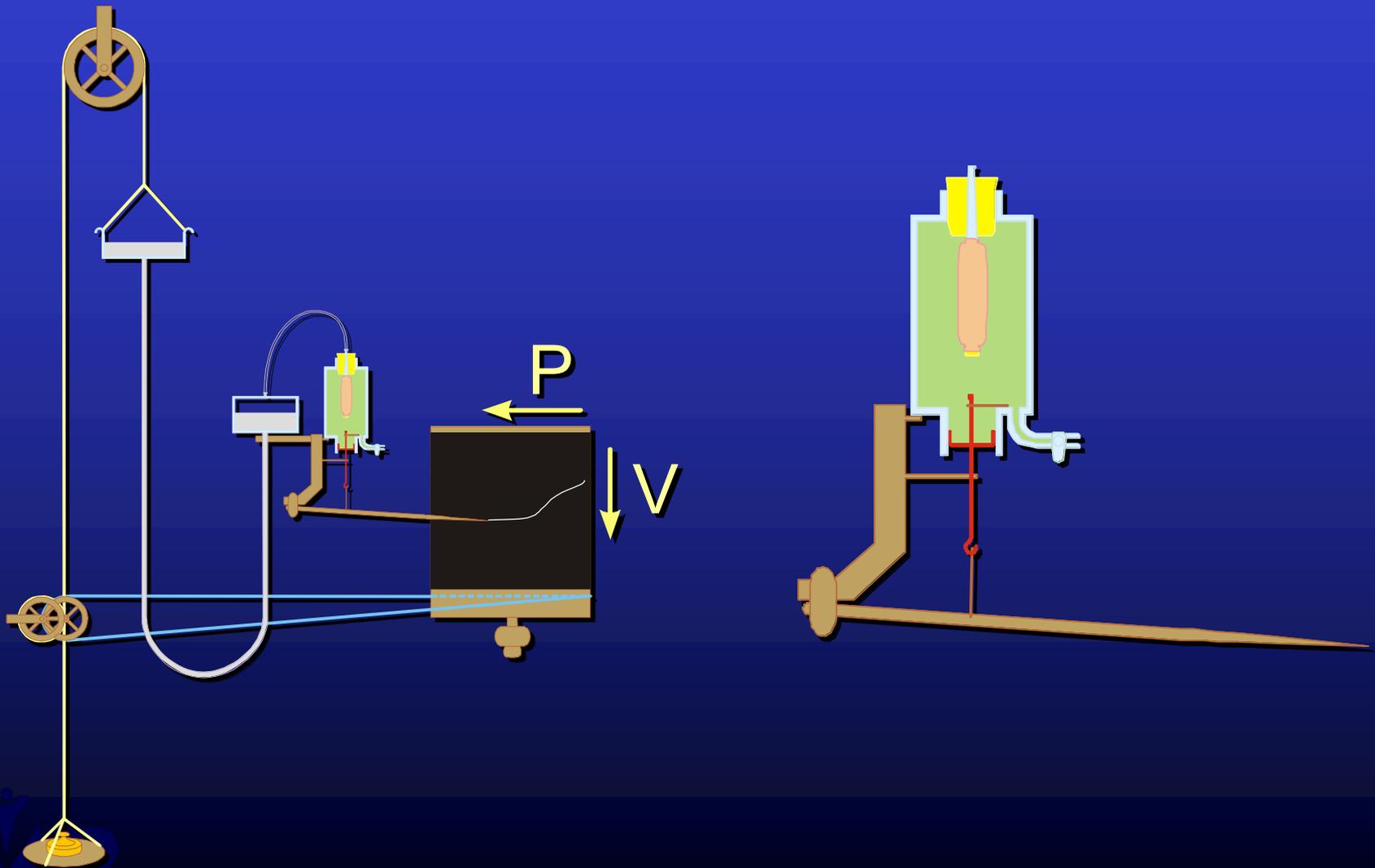
Antecedentes históricos (Roy, 1881)



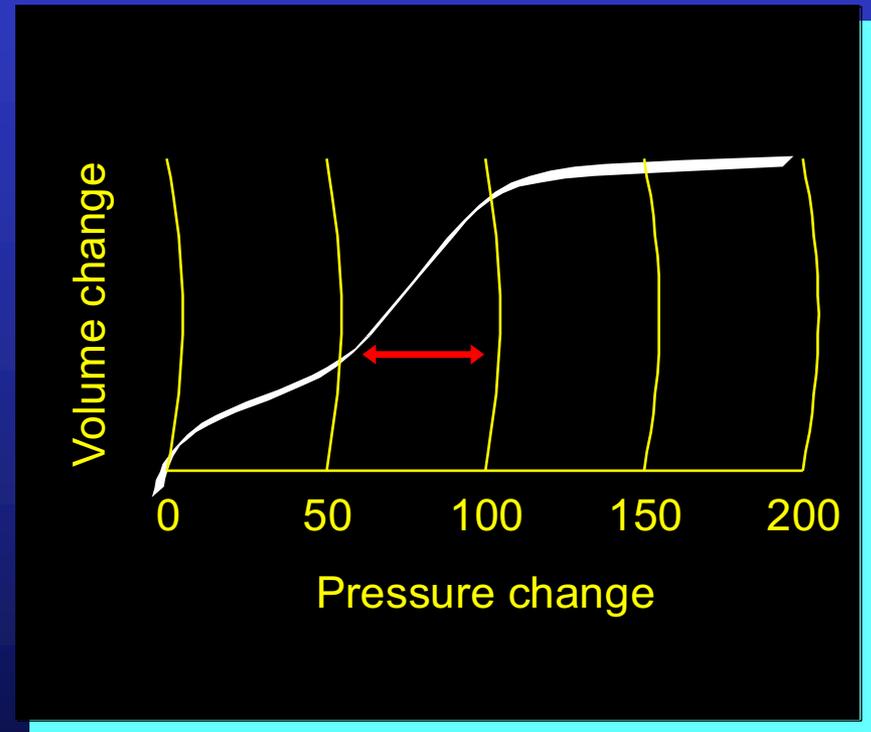
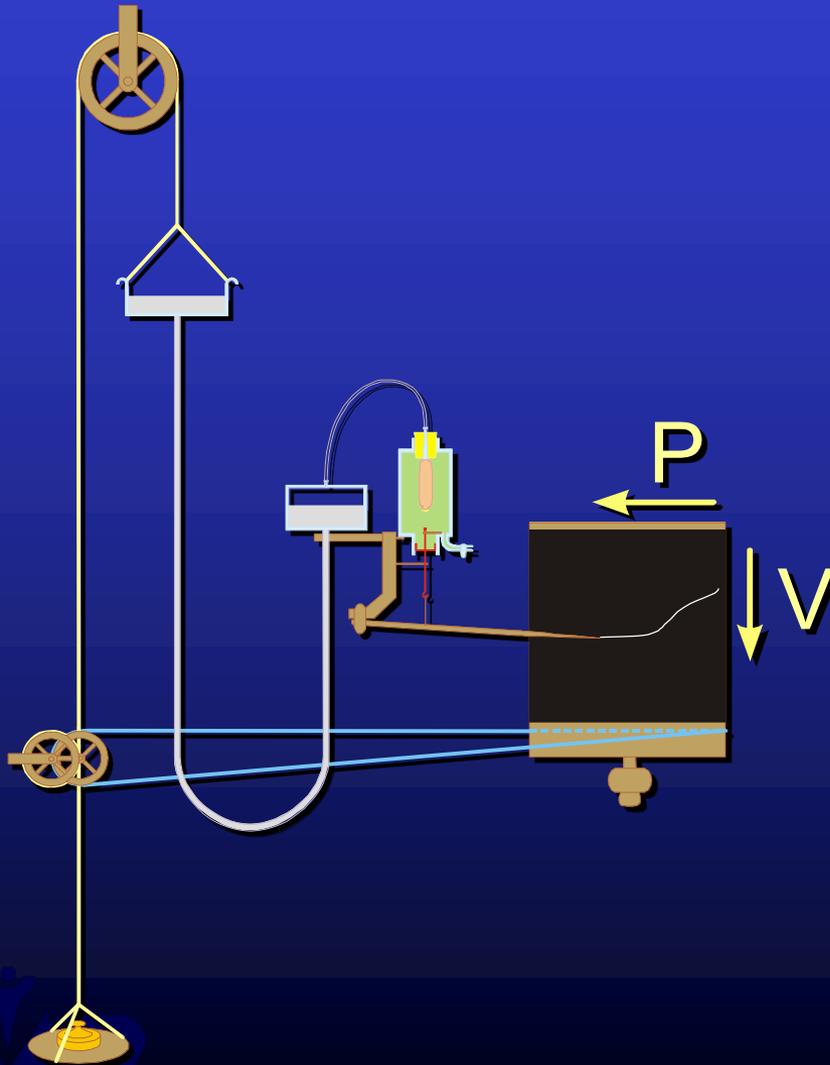
Antecedentes históricos (Roy, 1881)



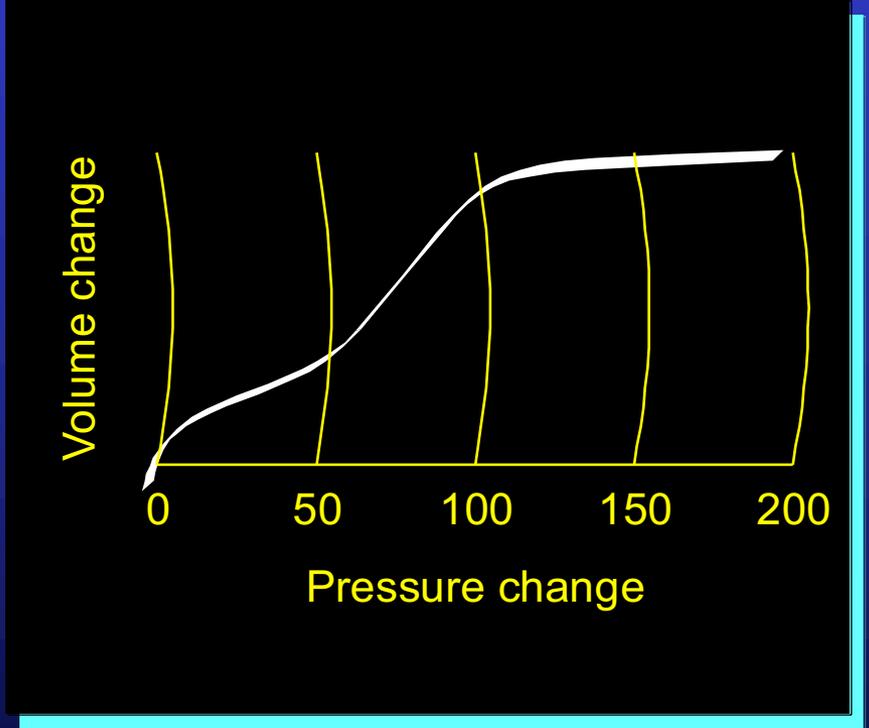
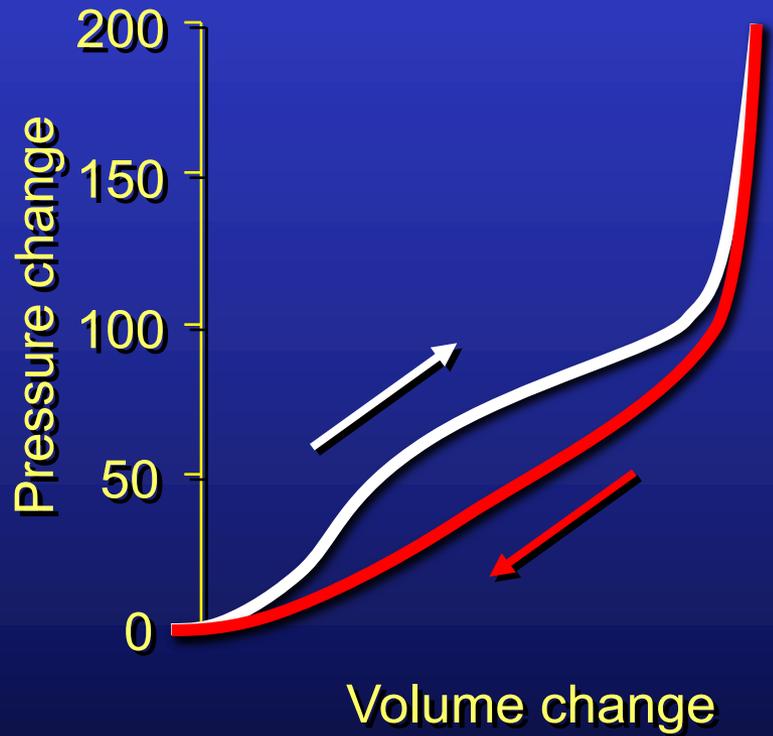
Antecedentes históricos (Roy, 1881)

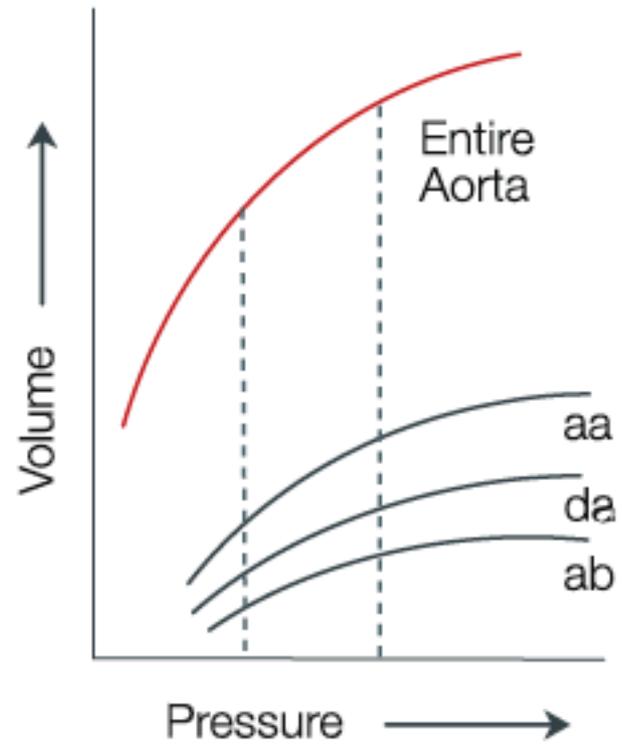
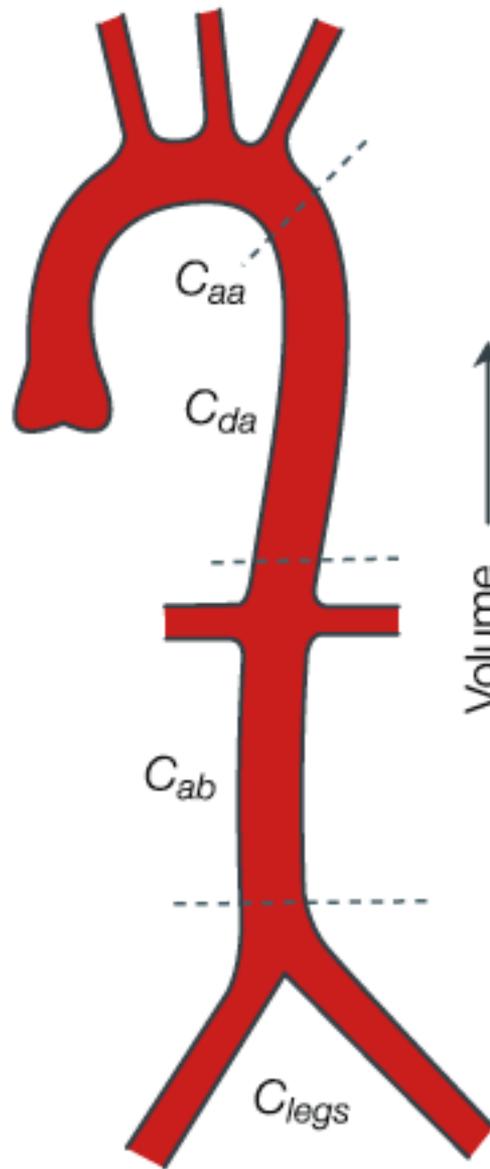


Antecedentes históricos (Roy, 1881)



Antecedentes históricos (Roy, 1881)

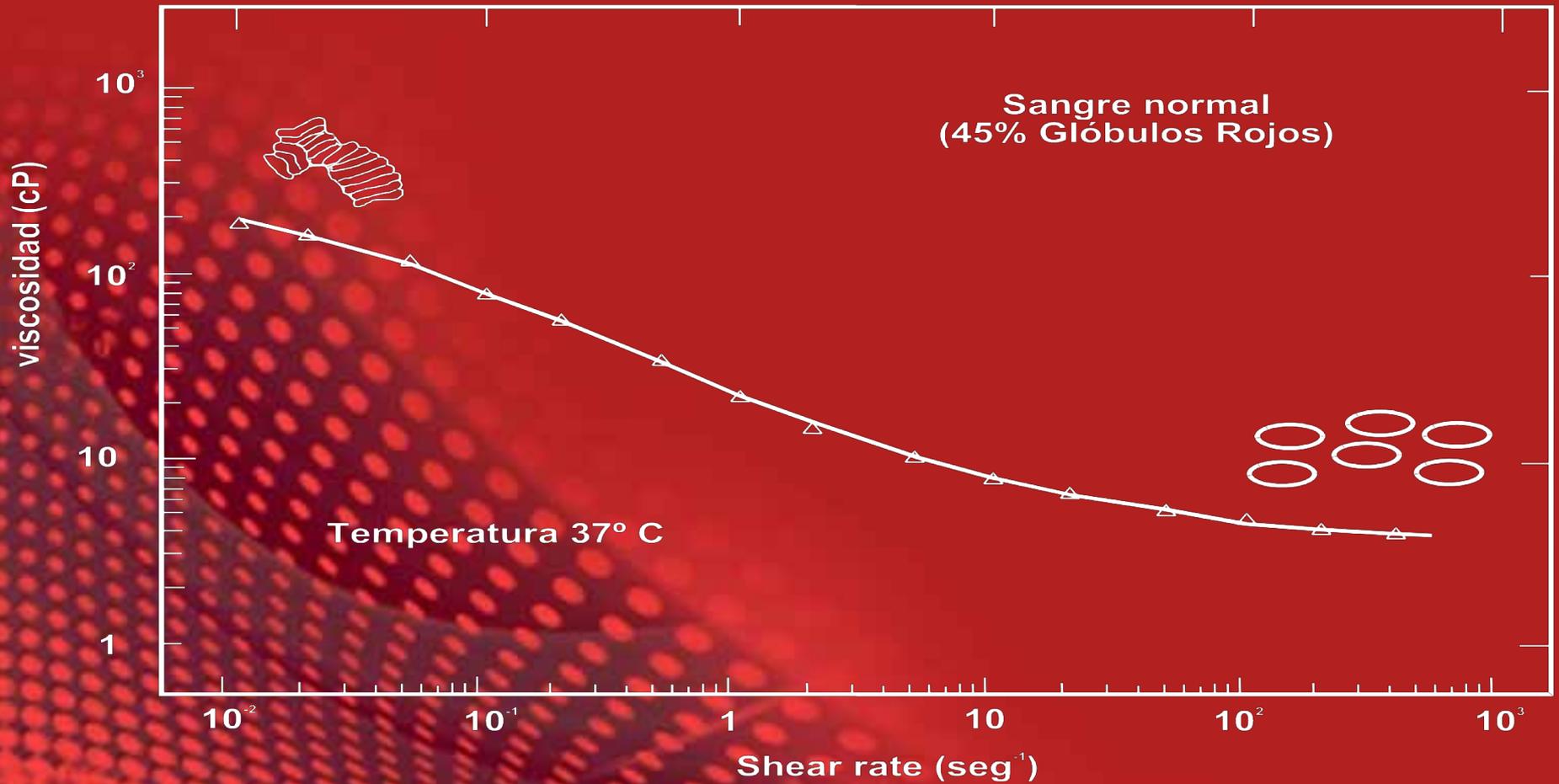




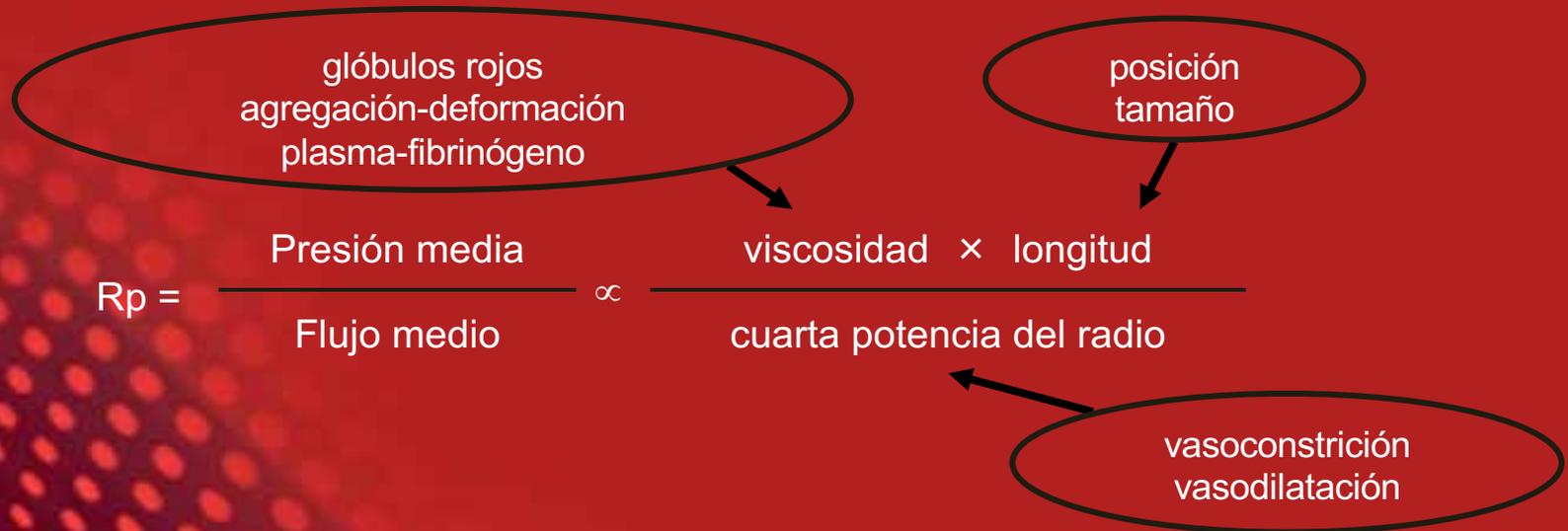


Contenido: SANGRE

Viscosidad y cizallamiento



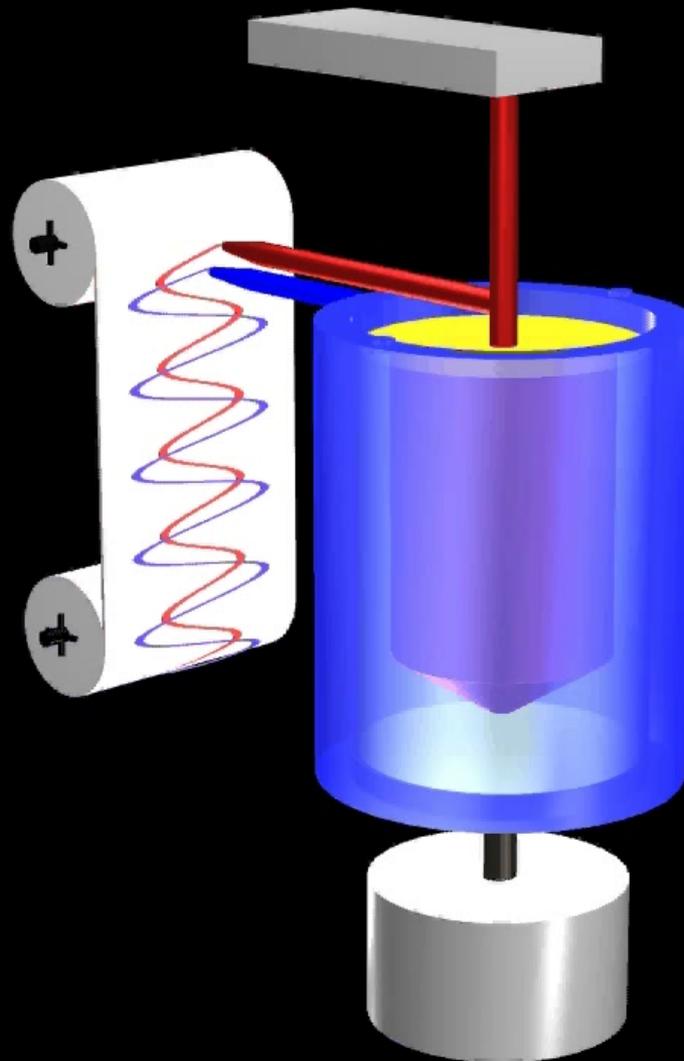
Resistencia periférica

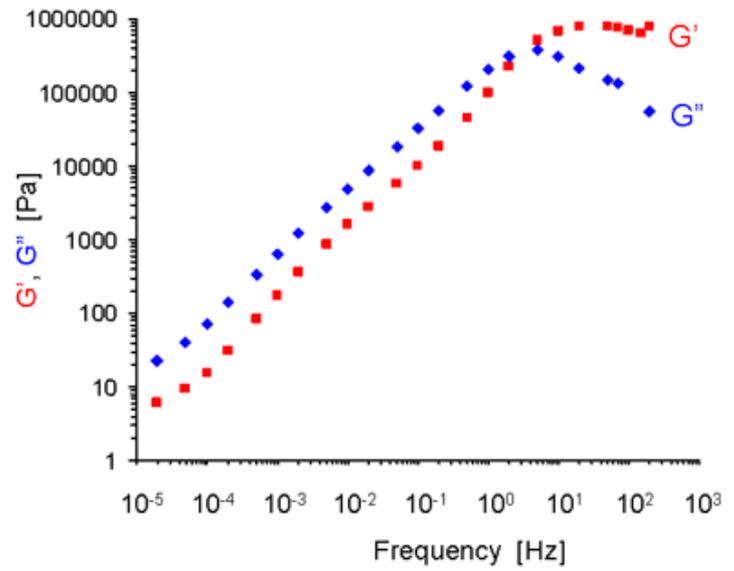
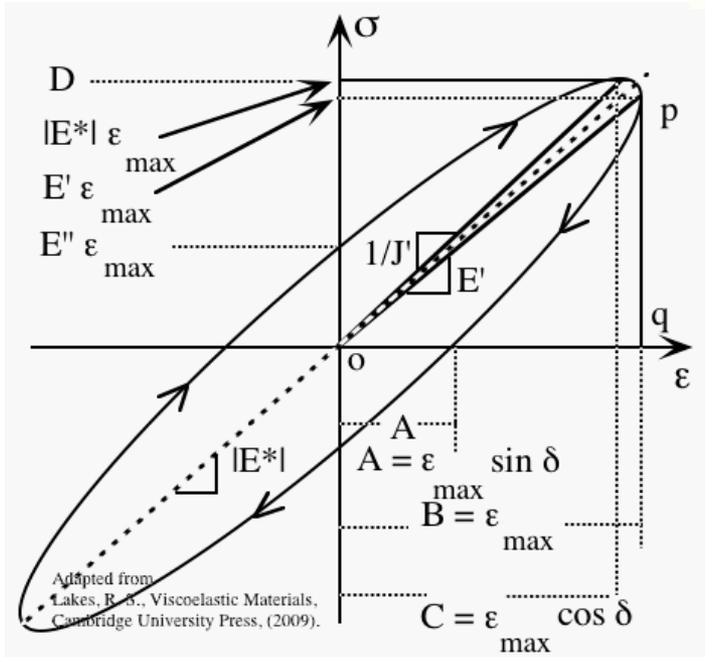
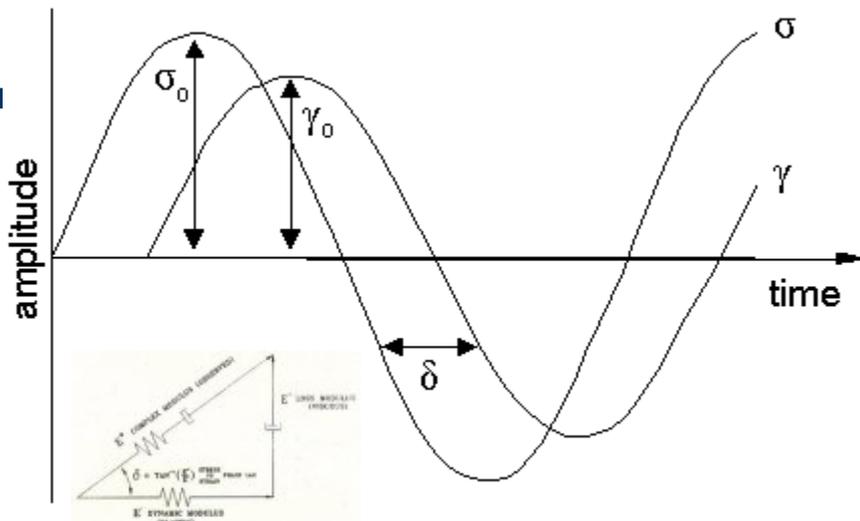
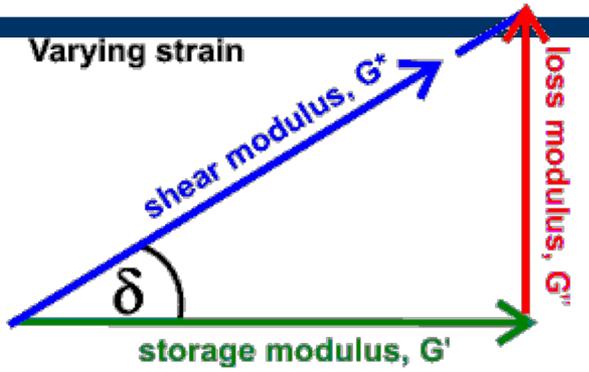


La disipación de la energía está generalmente asociada a fenómenos viscosos y por lo tanto a la disipación de calor por fricción. En el caso de sistema arterial la resistencia está concentrada en el lecho arteriolo-capilar y dos magnitudes físicas tienen una acción destacada sobre ella: la viscosidad de la sangre y la dimensión de los vasos



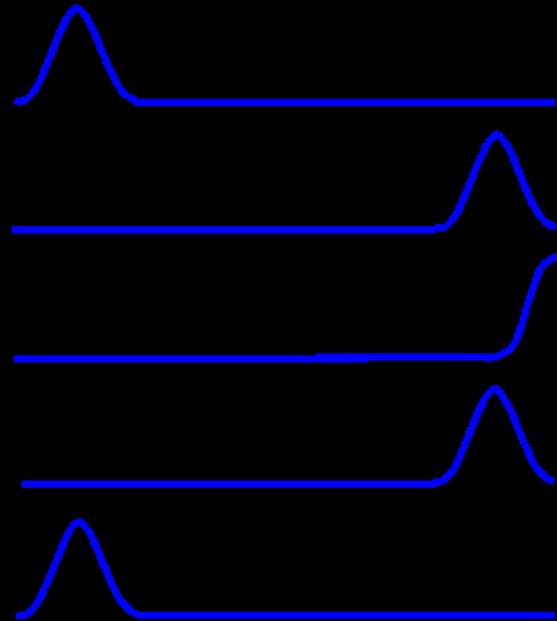
Bio Reología

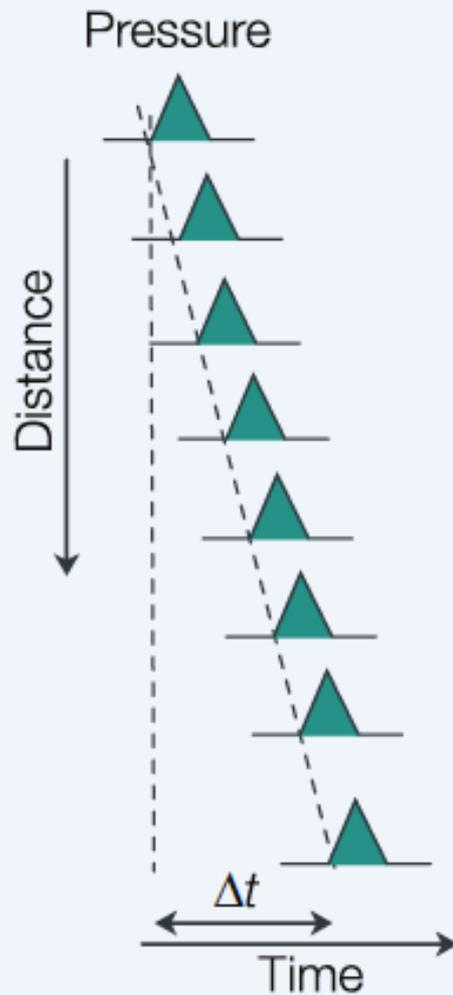




LA ONDA REFLEJADA

y sus efectos sobre la forma del pulso





Wave speed, c , is

$$c = \Delta x / \Delta t$$

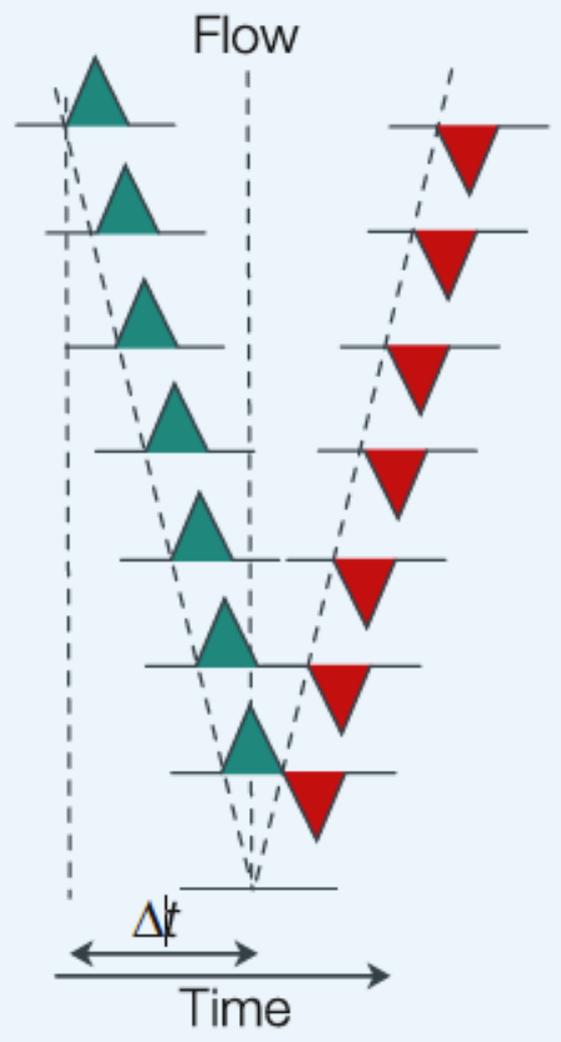
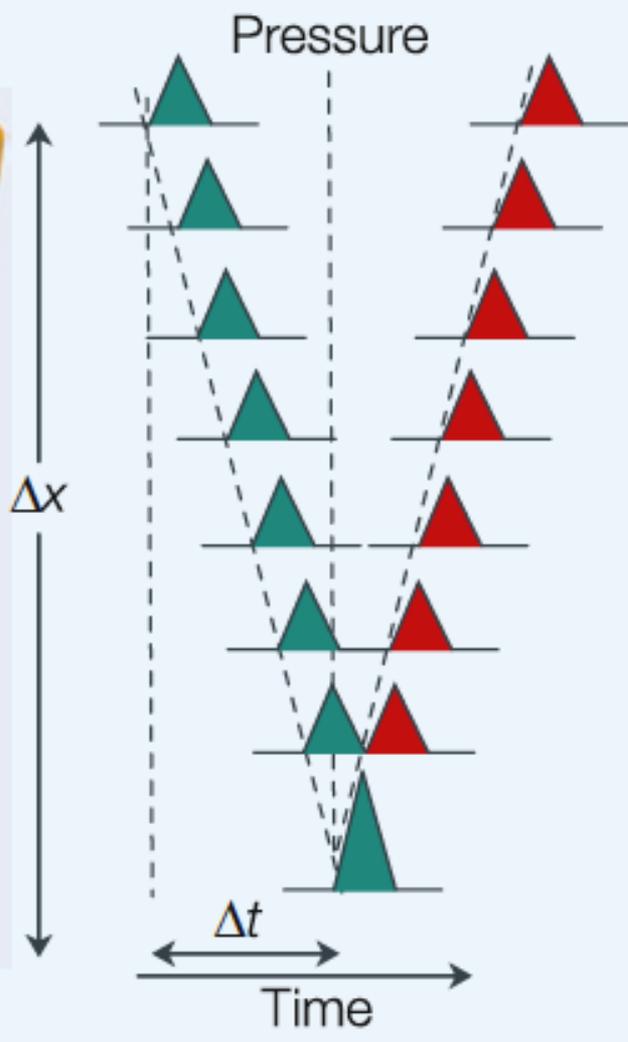
Wave speed as a function of area compliance, $C_A = \Delta A / \Delta P$, wall elastic modulus, E_{inc} , wall thickness, h , and blood density is:

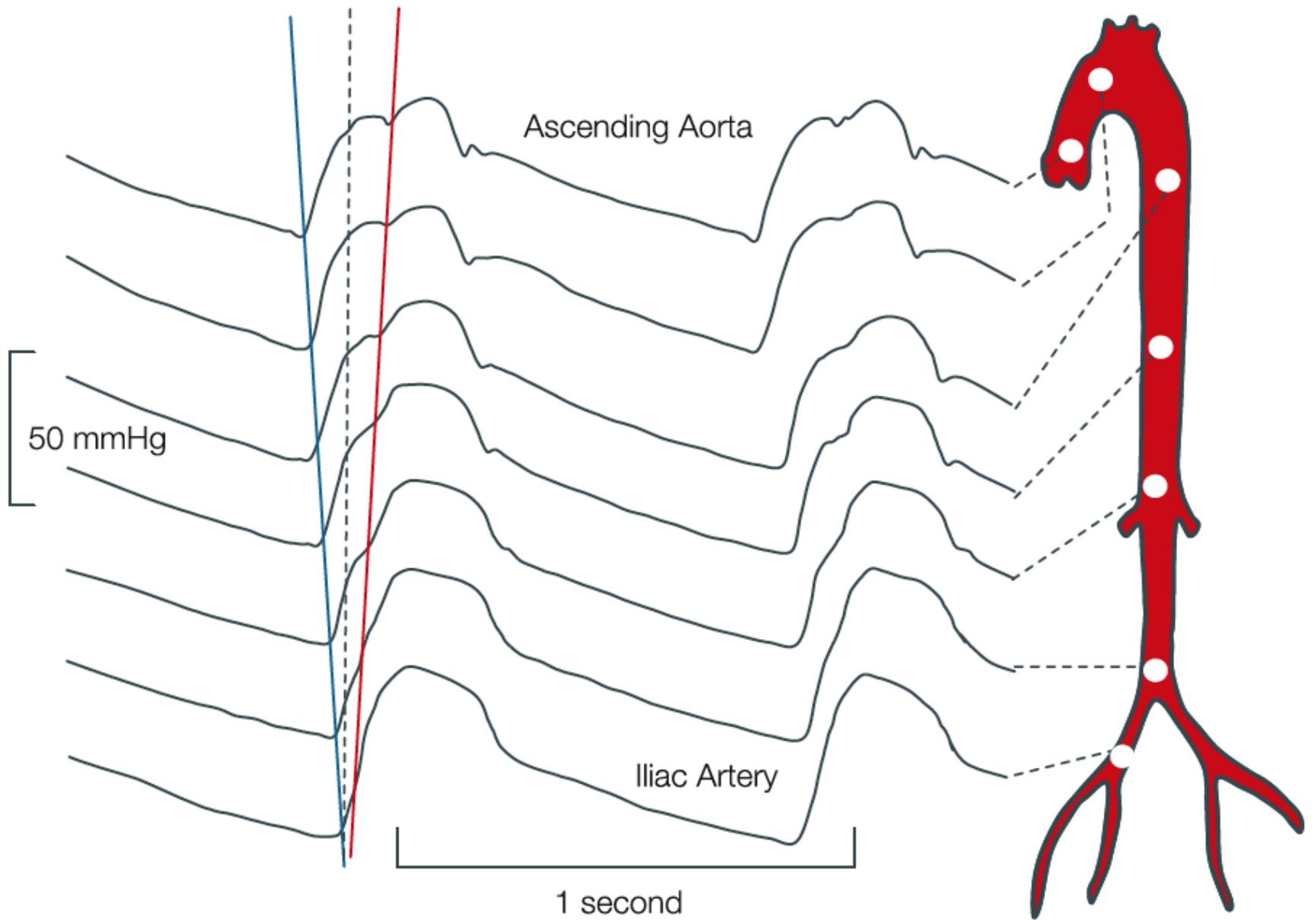
$$c = \sqrt{A / \rho C_A}$$

(Newton-Young, or Frank, or Bramwell-Hill equation)

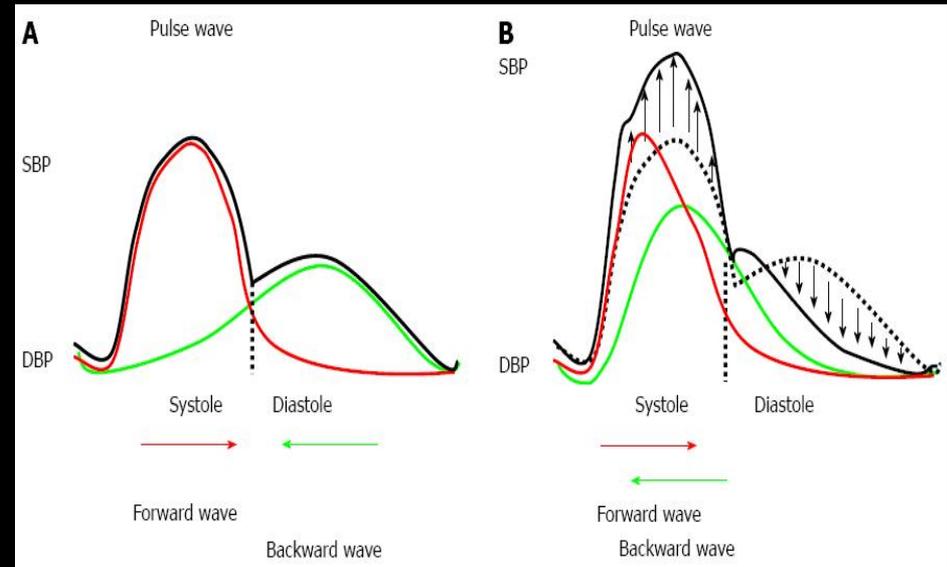
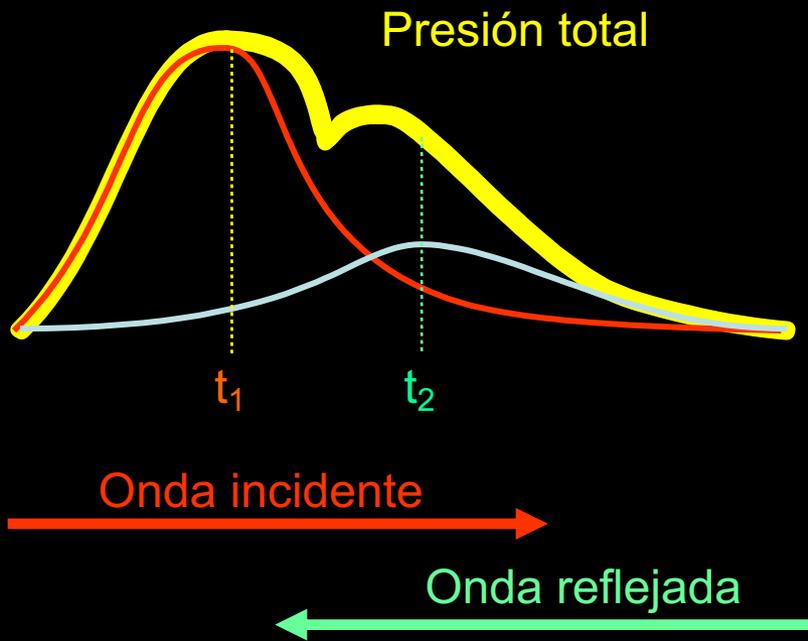
$$c = \sqrt{E_{inc} h / \rho D}$$

(Moens-Korteweg equation)



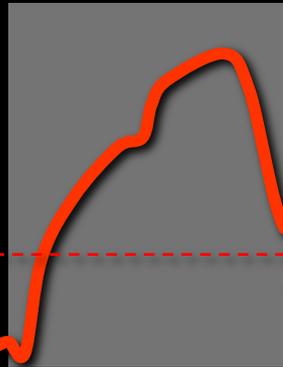


Onda incidente y onda reflejada

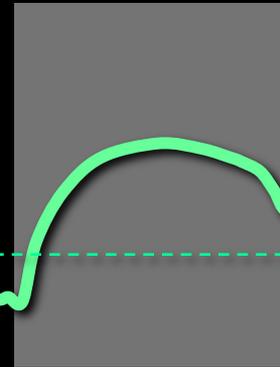


Tipos de ondas

Tipo A (aged)



Tipo C (child)

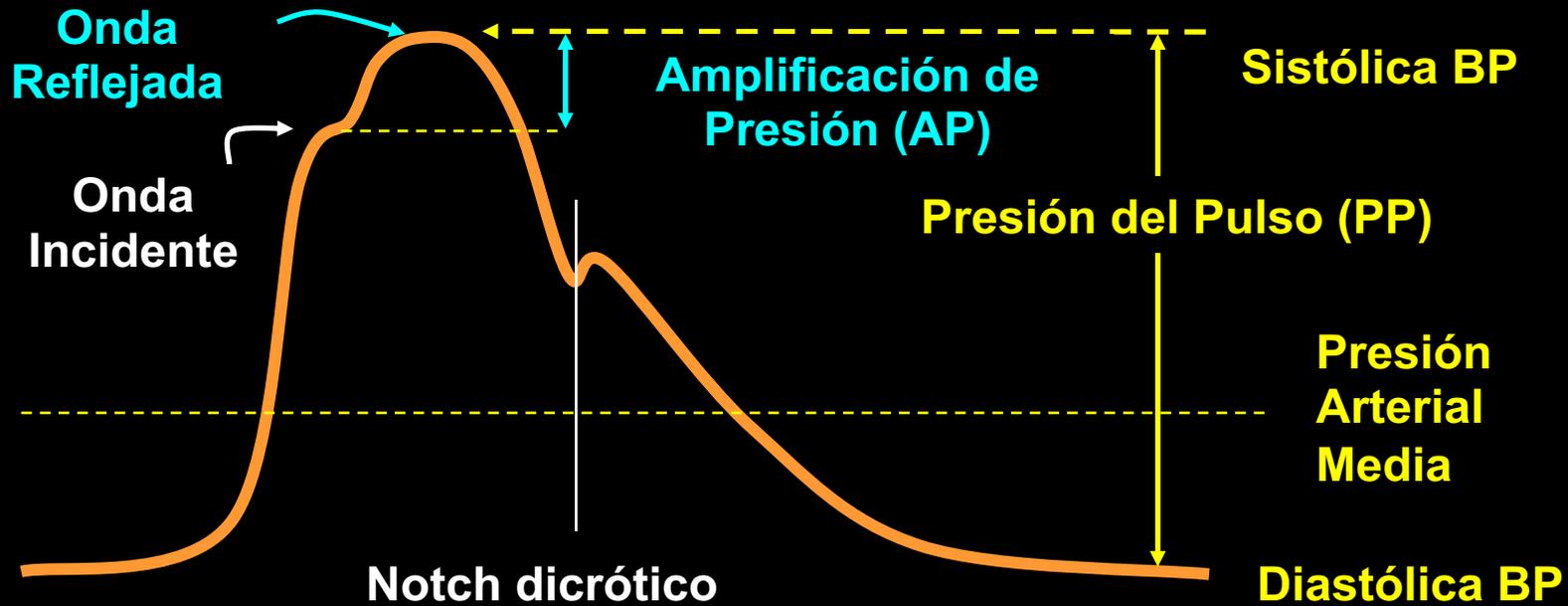


Los efectos de una onda reflejada temprana pueden ser comprendidos comparando las ondas de presión de sujetos jóvenes (tipo C) y sujetos de edad avanzada (tipo A) con la misma presión arterial media, pero con diferente presión media durante la diástole.

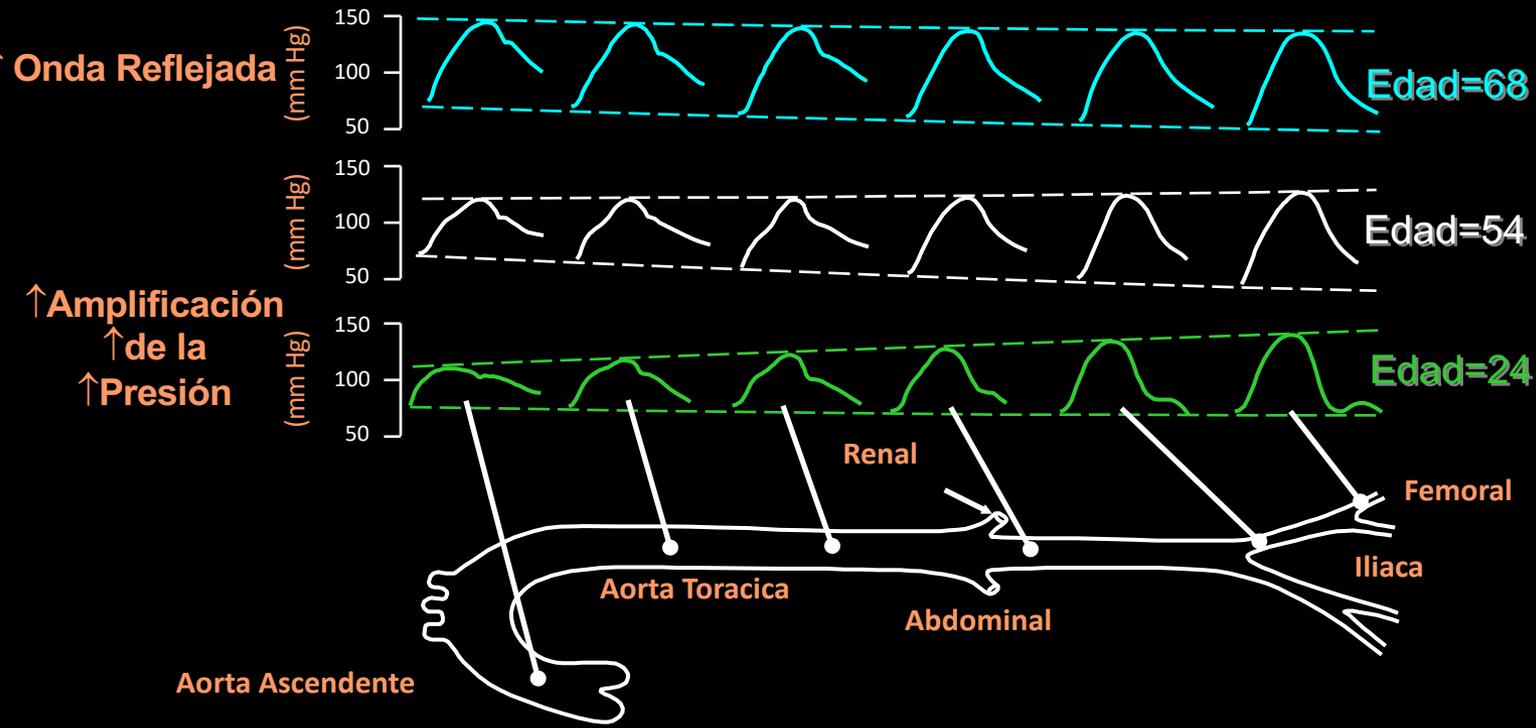


PRESION CENTRAL

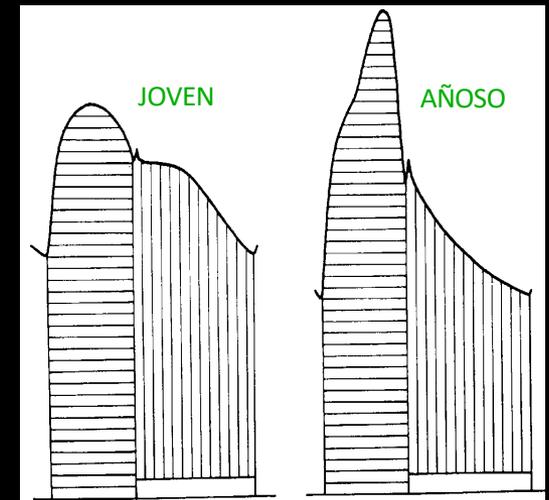
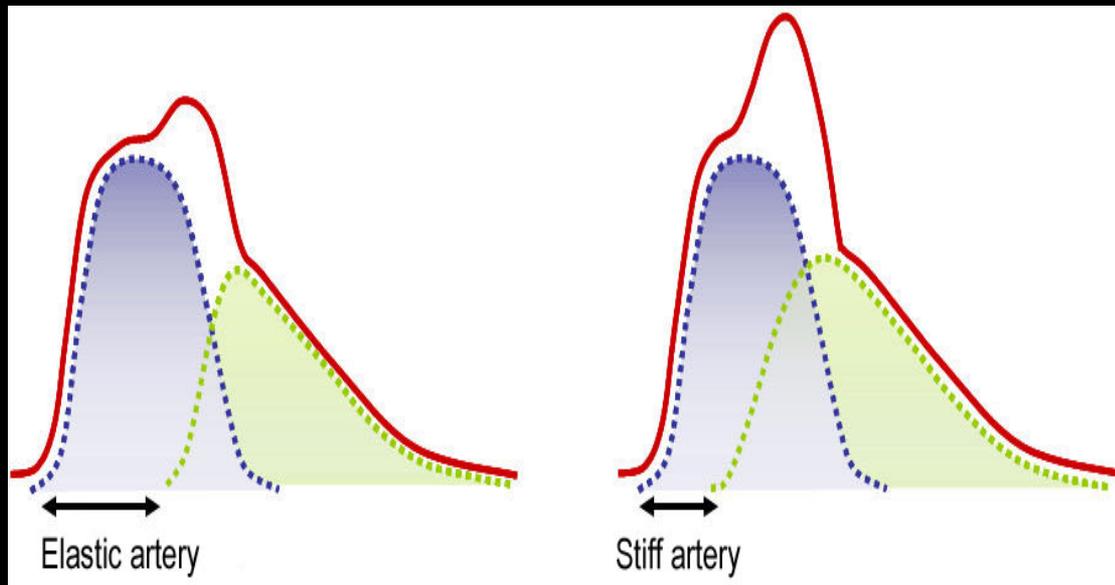
$$\text{Indice de aumentación (AIx)} = \text{AP/PP}$$



AMPLIFICACIÓN del PULSO



IMPLICANCIAS DEL AUMENTO EN RIGIDEZ



Arribo temprano BP ↗



- * Poscarga VI ↗
- * Consumo de oxígeno ↗
- * Perfusión coronaria ↘



- ISQUEMIA
- HIPERTROFIA
- FALTA CARDIACA
- Stroke/Falla renal

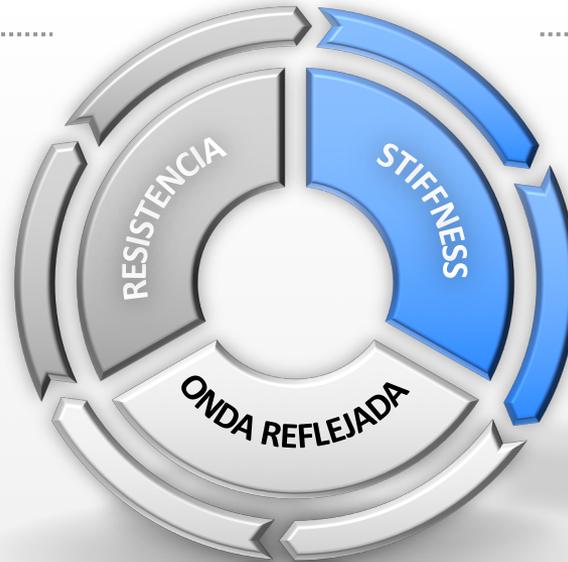


IMPEDANCIA ARTERIAL

CARGA HIDRAULICA AL CORAZON. POSCARGA

RESISTENCIA

CONTENIDO. MICROCIRCULACIÓN
LEY DE POISEUILLE

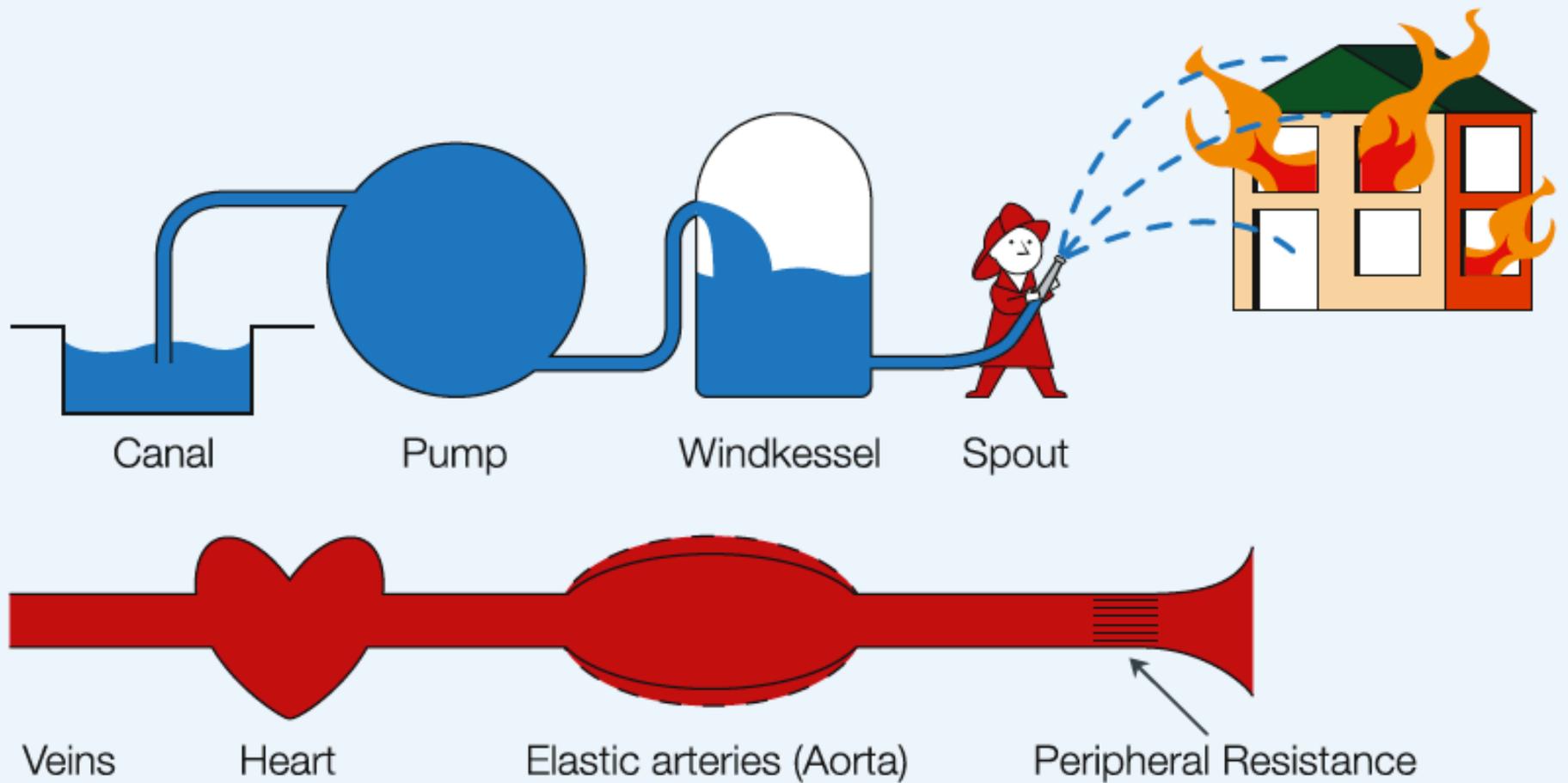


ELASTICIDAD

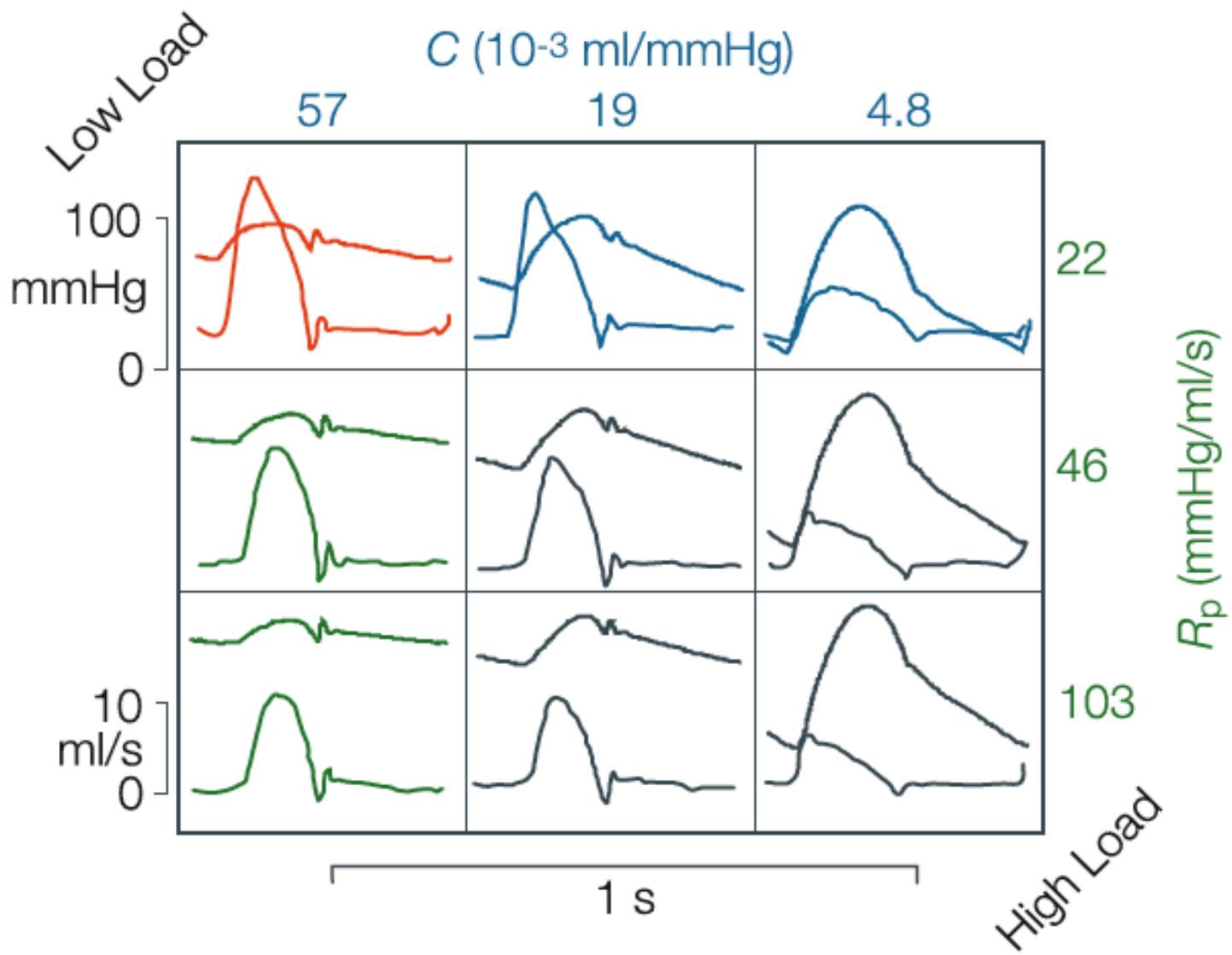
CONTINENTE: ARTERIAL STIFFNESS
LEY DE HOOKE

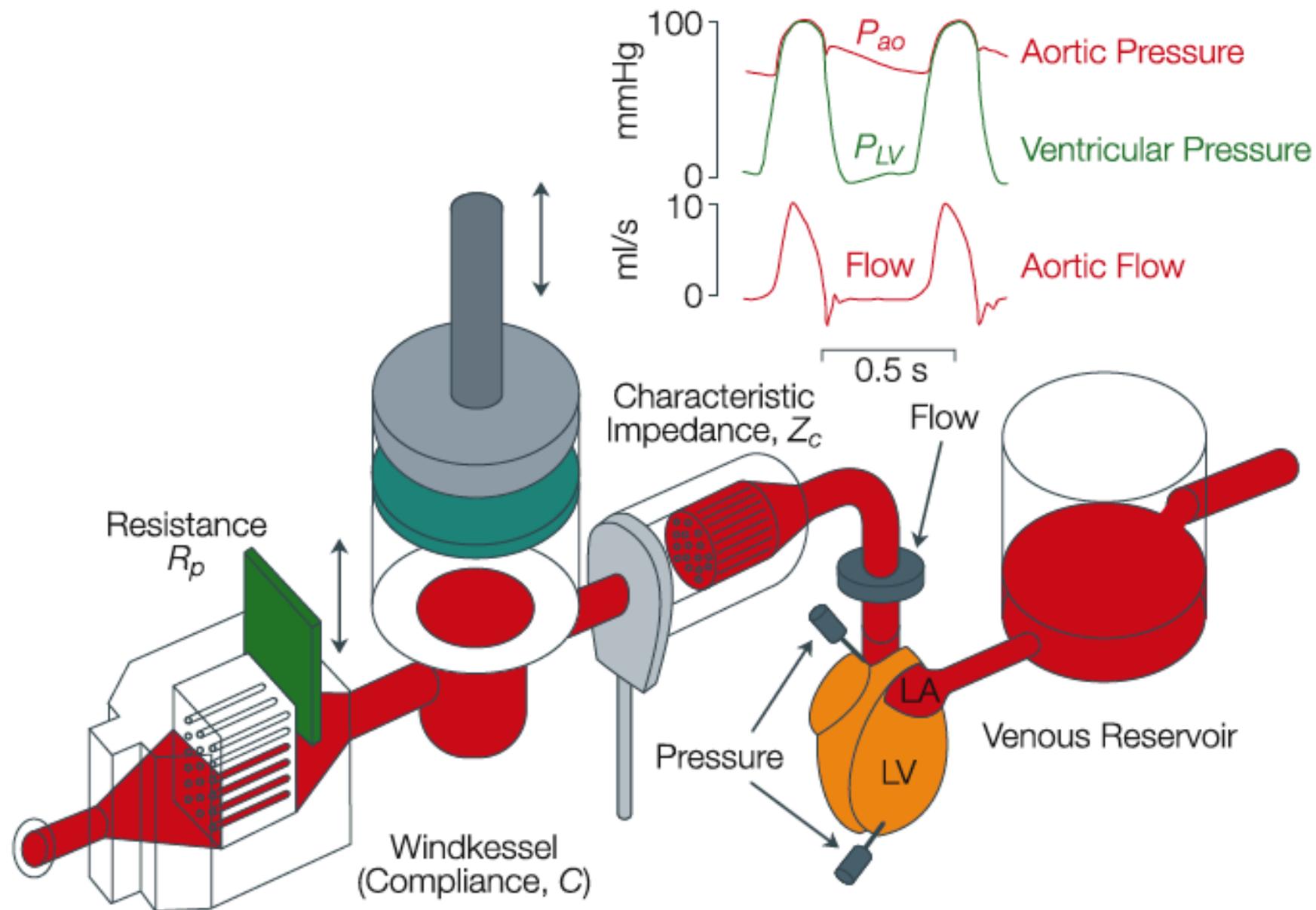
ONDA REFLEJADA

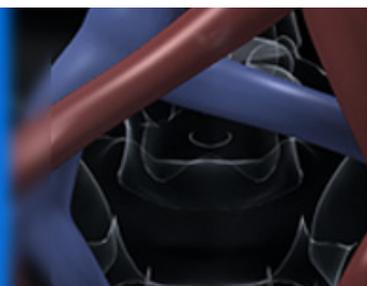
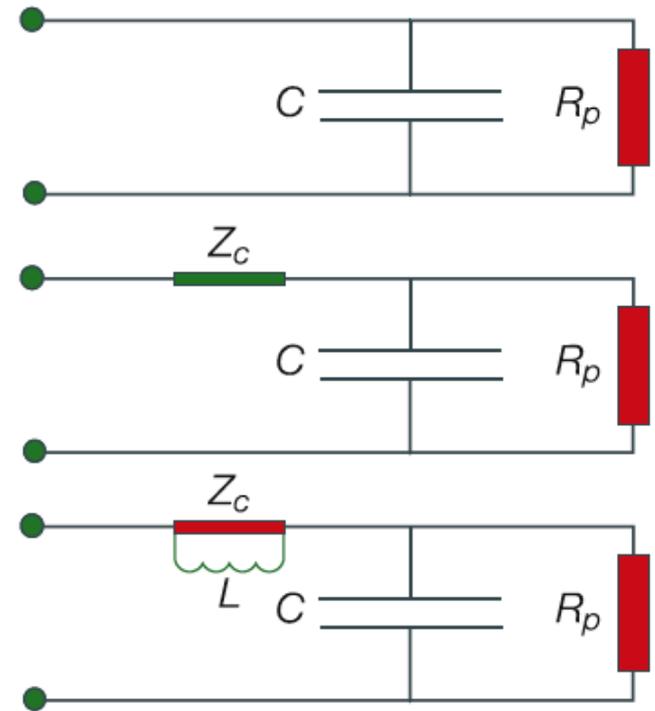
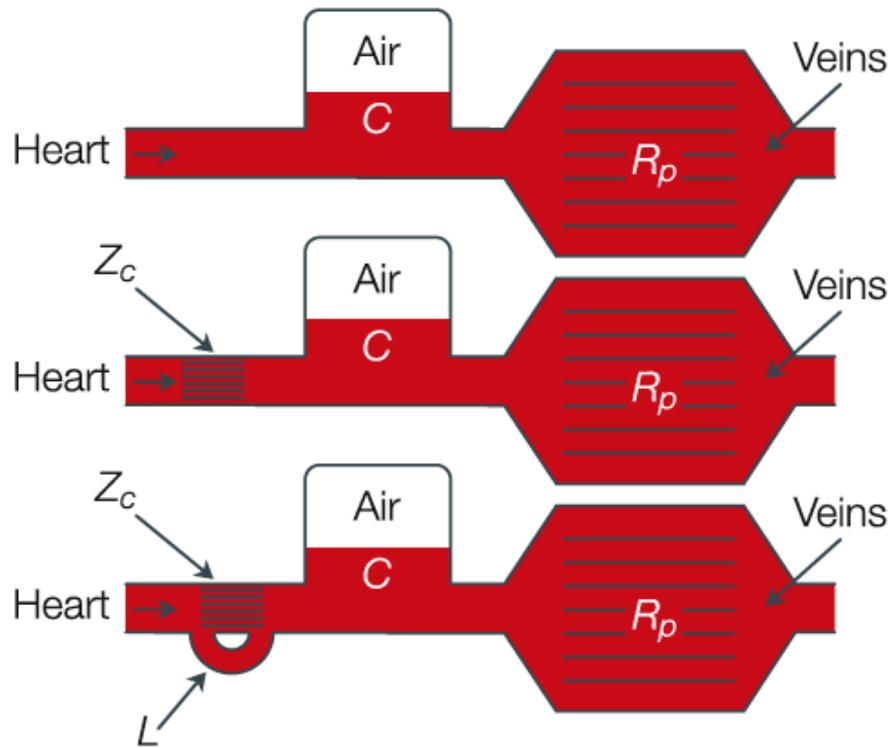
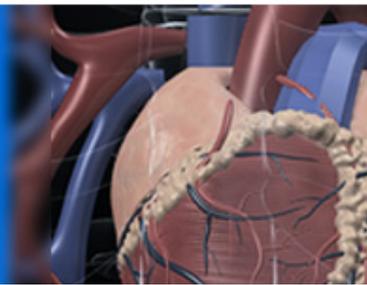
CENTRAL PULSE WAVE ANALYSIS & AUGMENTATION INDEX



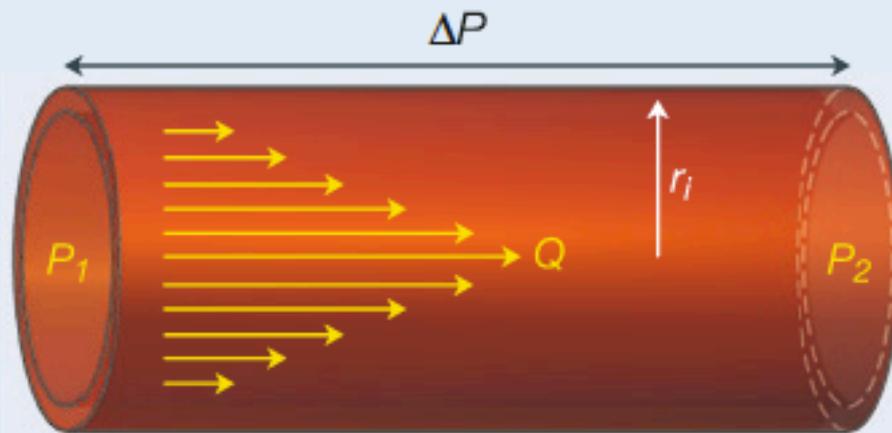
The analogy between the fire engine with the Windkessel and the arterial system. The peripheral resistance is the summed resistance of all small arteries, arterioles and capillaries. Total arterial compliance is the sum of the compliances of all arteries, mainly the conduit arteries. The Windkessel model can help us to understand how the arterial system functions, can be used as a realistic load in isolated heart studies, can be used in modeling, and can form the basis for estimating arterial system parameters. Adapted from [1], used by permission.







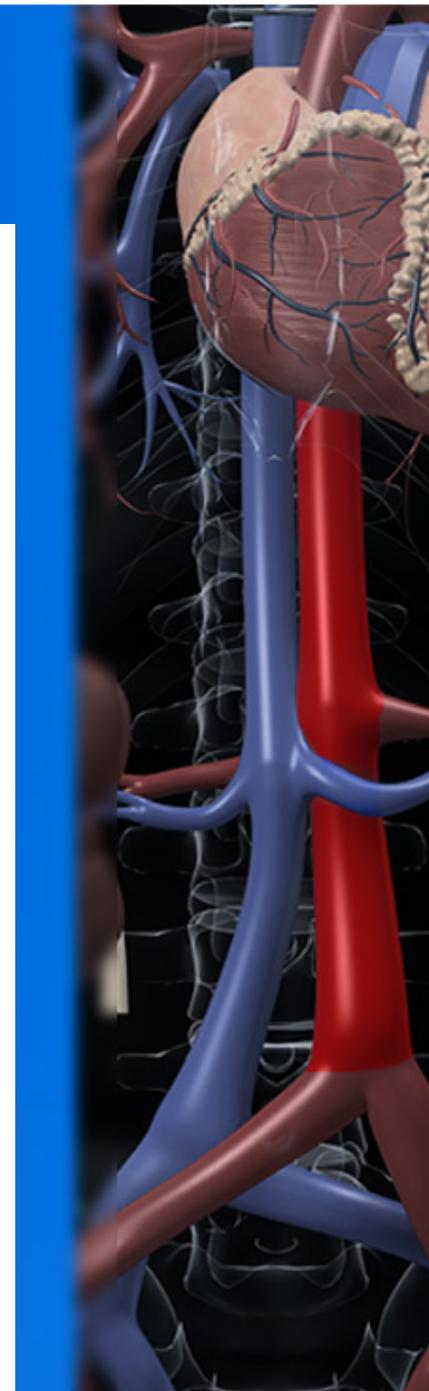
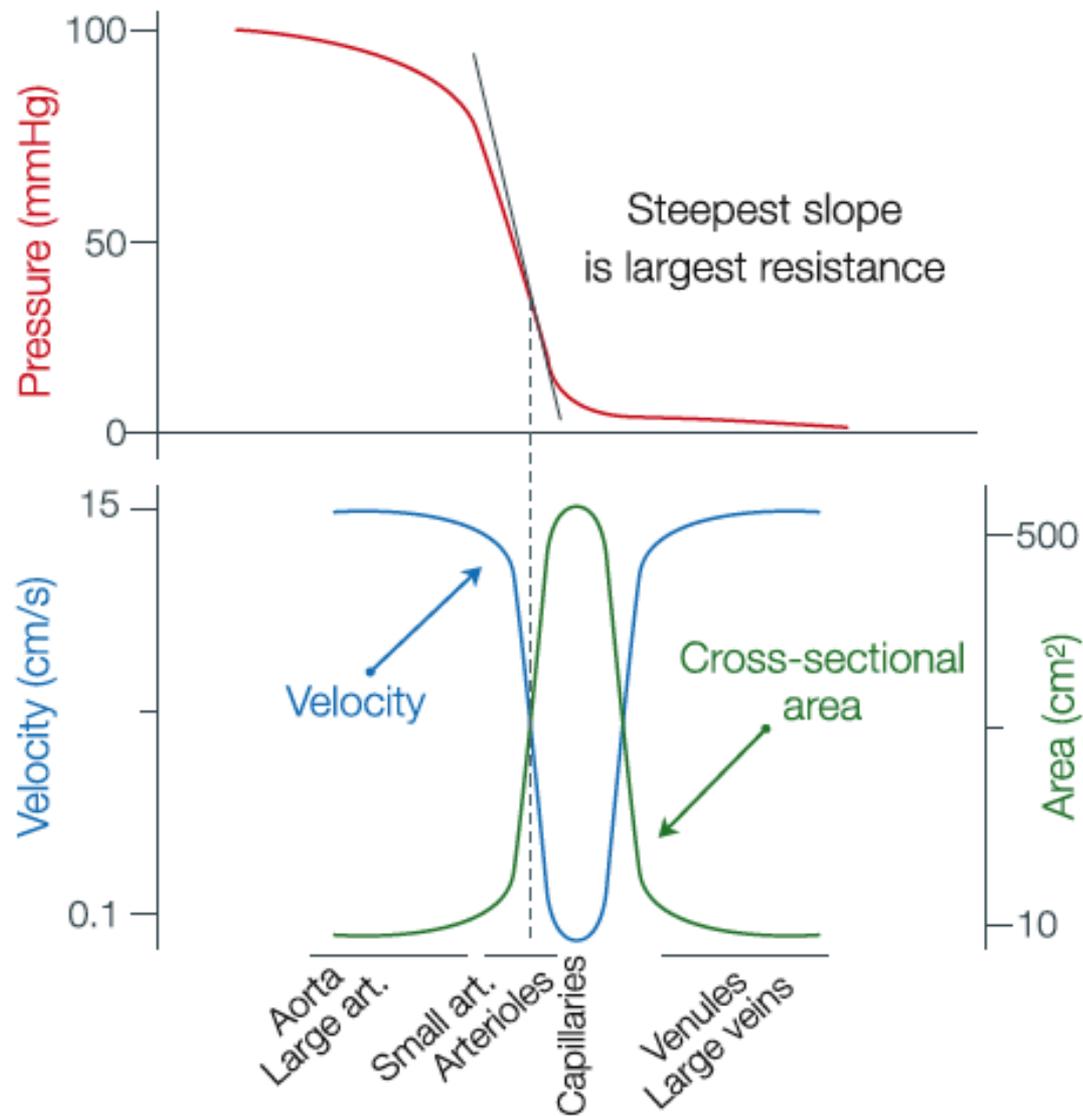
RESISTENCIA



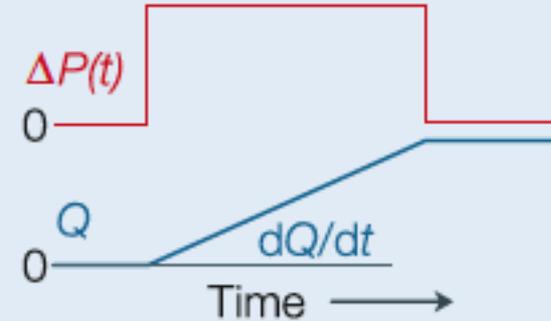
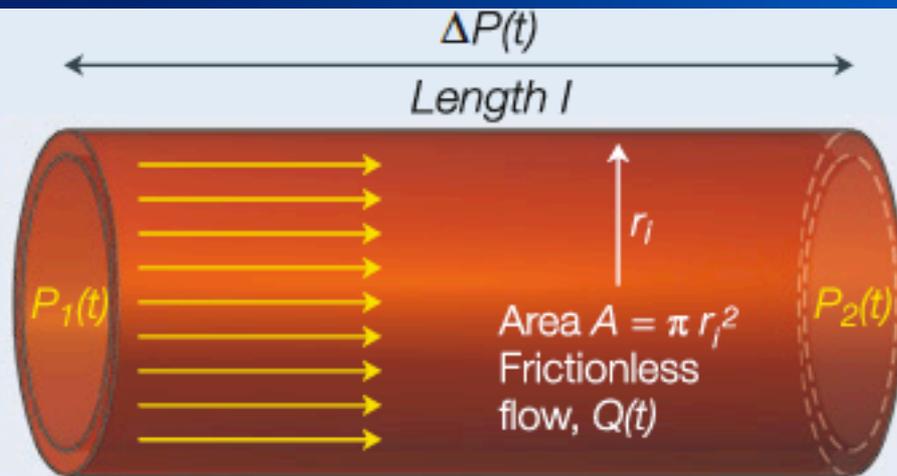
$$R = \Delta P / Q$$
$$= 8\eta / \pi r_i^4$$

Resistance is a practical quantitative description of the relation between pressure difference and flow through a blood vessel. For a single uniform vessel Poiseuille's law gives its resistance, but in practice resistance can be obtained using Ohm's law. In other words, resistance although depending on the vascular geometry and blood viscosity, can be calculated directly from measurements of mean pressure difference and mean blood flow. Detailed knowledge of the vascular geometry is not required. Ohm's law not only pertains to single blood vessels but may also be applied to combinations of vessels, whole organ beds, and the whole systemic or pulmonary circulation. Rules for addition of resistances are discussed below. Resistance should always be calculated from a pressure difference, $P_1 - P_2$, indicated by ΔP . However, in the systemic circulation venous pressure is usually much lower than aortic pressure and can be disregarded. This is not the case in the pulmonary circulation. The systemic, and pulmonary vascular resistances are mainly determined by the resistance of small arteries and arterioles. This means that the mean pressure in all conduit arteries is almost the same. The arterioles act as resistances to regulate flow to the local tissue.

RESISTENCIA



INERTANCIA

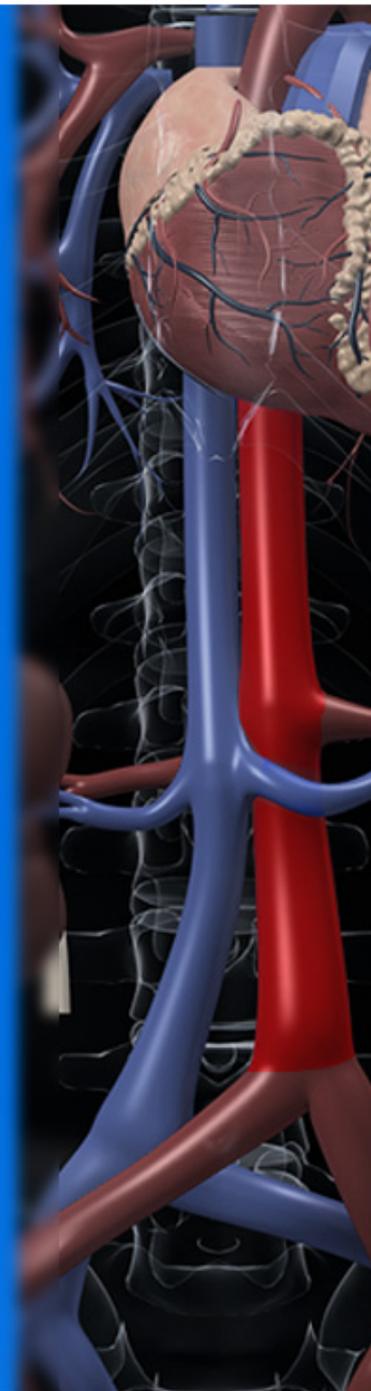
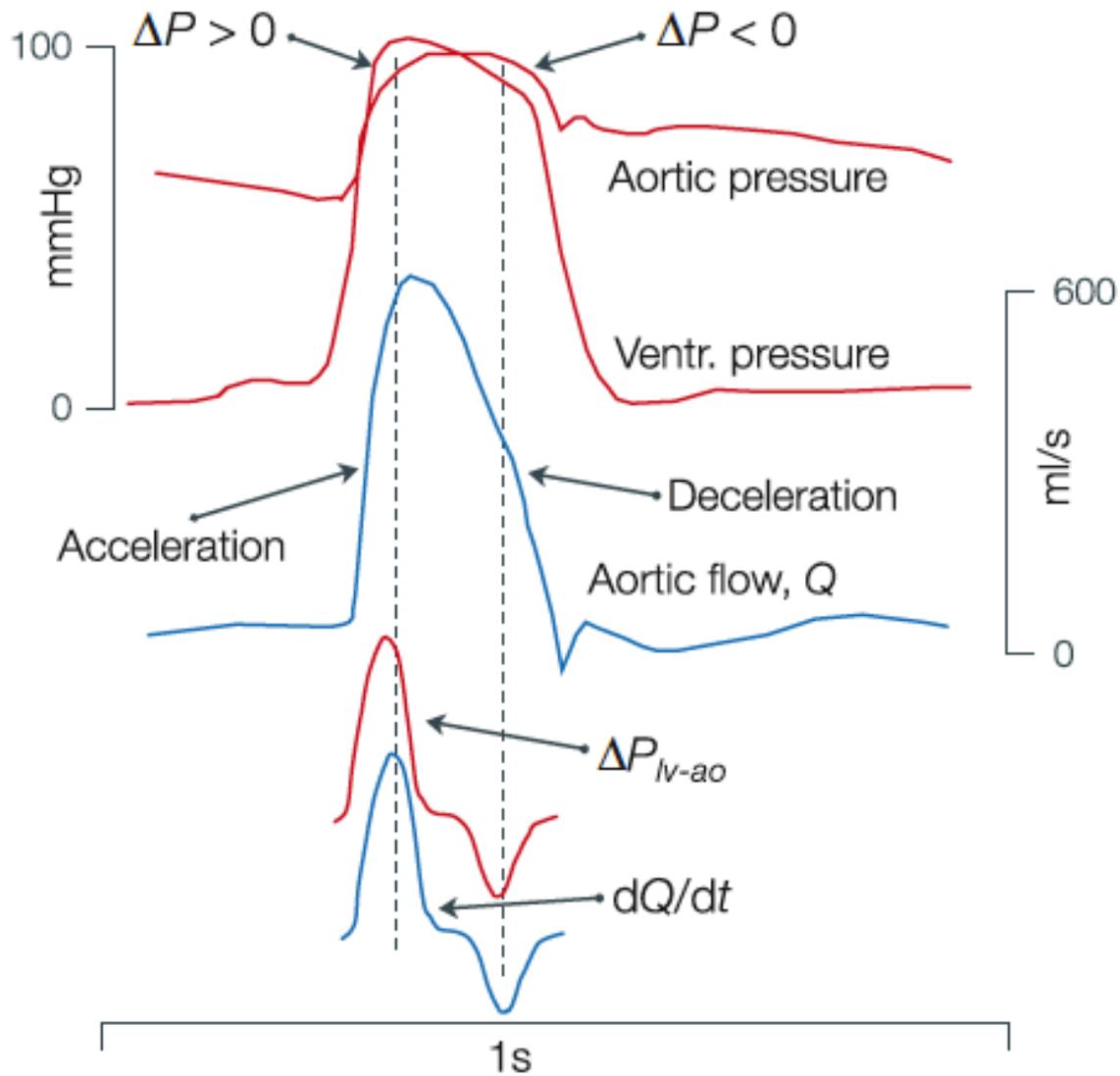


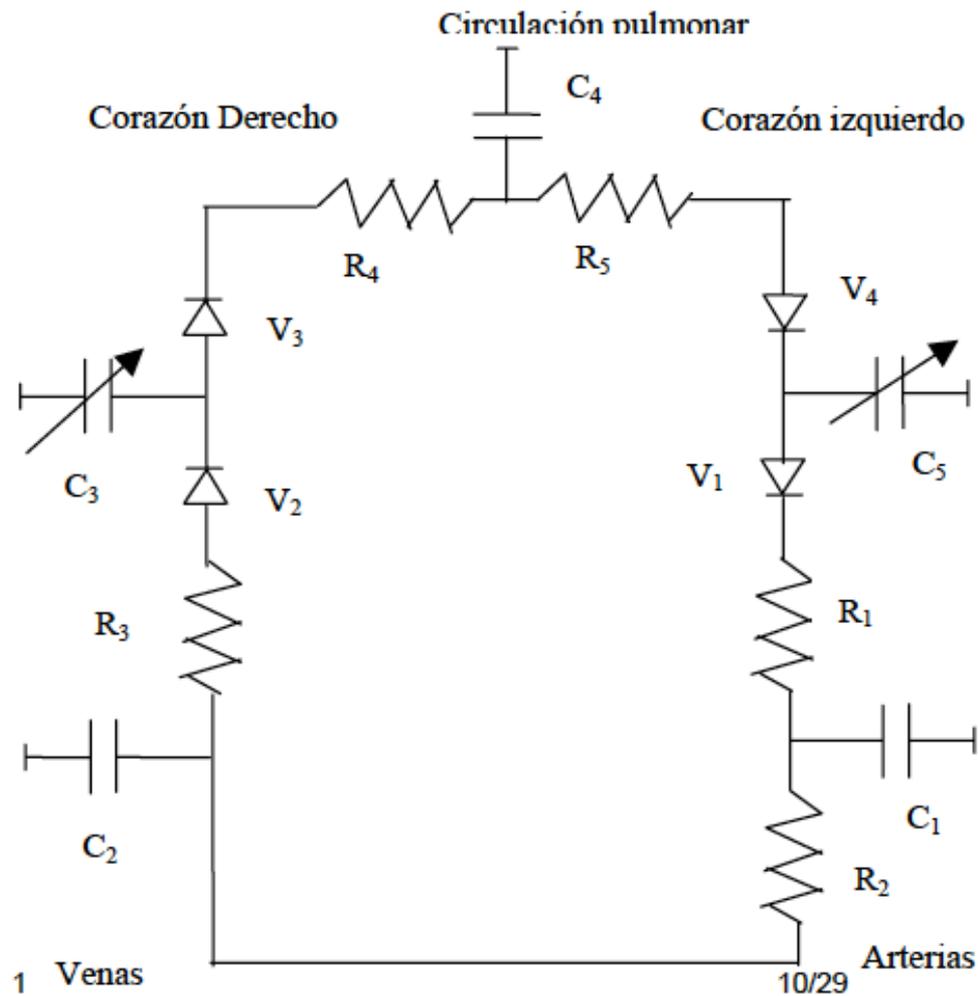
$$\Delta P(t) = L dQ/dt$$

with $L = \rho l/A$

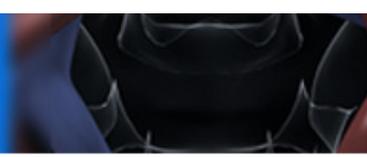
Inertance relates the pressure drop with the acceleration of flow. This figure shows that when blood is subjected to an increase in pressure difference the velocity will change. Assuming no friction, the relation between the pressure drop, $\Delta P(t) = P_1(t) - P_2(t)$, and the rate of change of blood flow dQ/dt , depends on the density of the blood, ρ , the cross-sectional area, A , and the length, l , of the blood vessel. We call this combined effect the inertance, $L = \rho l/A$. Inertance plays a dominating role in large blood vessels, where the viscous resistance is small and pulsatility is considerable. Inertance and Poiseuille resistance form the basis of the Oscillatory Flow Theory. The inertance and compliance in combination determine the wave speed and the characteristic impedance of a blood vessel (Appendix 4).

INERTANCIA





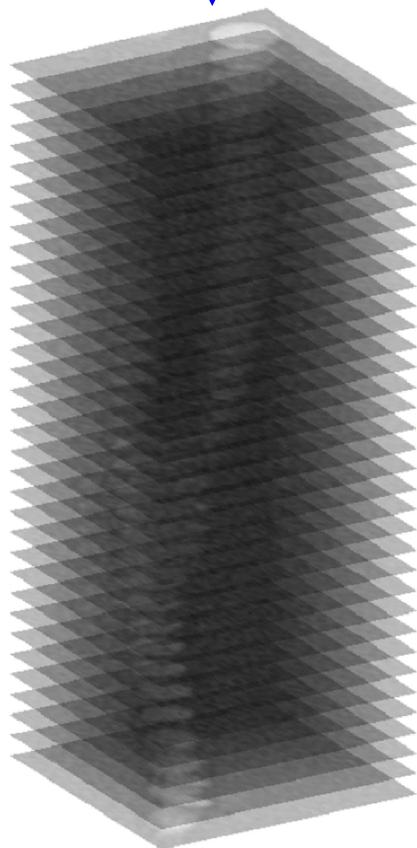
<i>Valores del circuito</i>	
$R1$ [dyn.s/cm ⁵]	250
$R2$ [dyn.s/cm ⁵]	3800
$R3$ [dyn.s/cm ⁵]	350
$R4$ [dyn.s/cm ⁵]	625
$R5$ [dyn.s/cm ⁵]	200
$C1$ [cm ³ /dyn]	0.00019
$C2$ [cm ³ /dyn]	0.0326
$C4$ [cm ³ /dyn]	0.0082



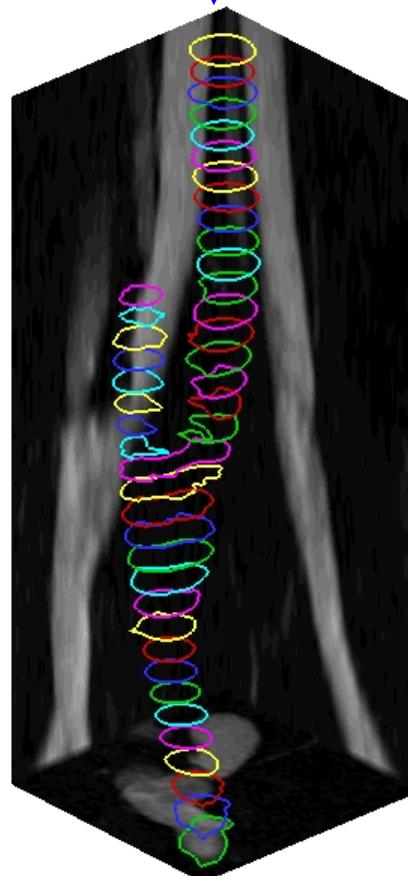
Extracción de la geometría de imágenes medicas



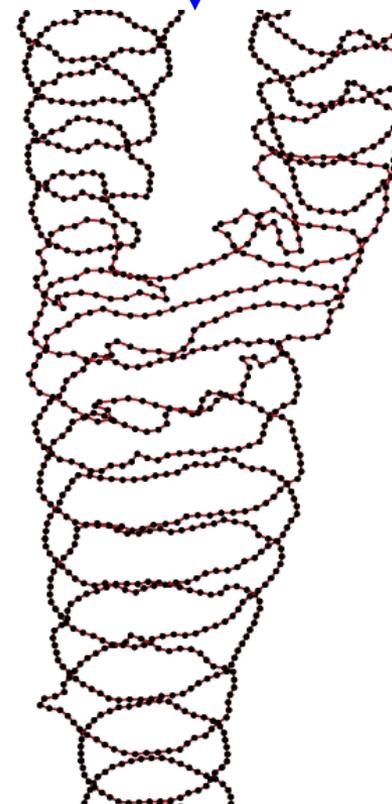
**MRI (Resonancia
MAgnética)**



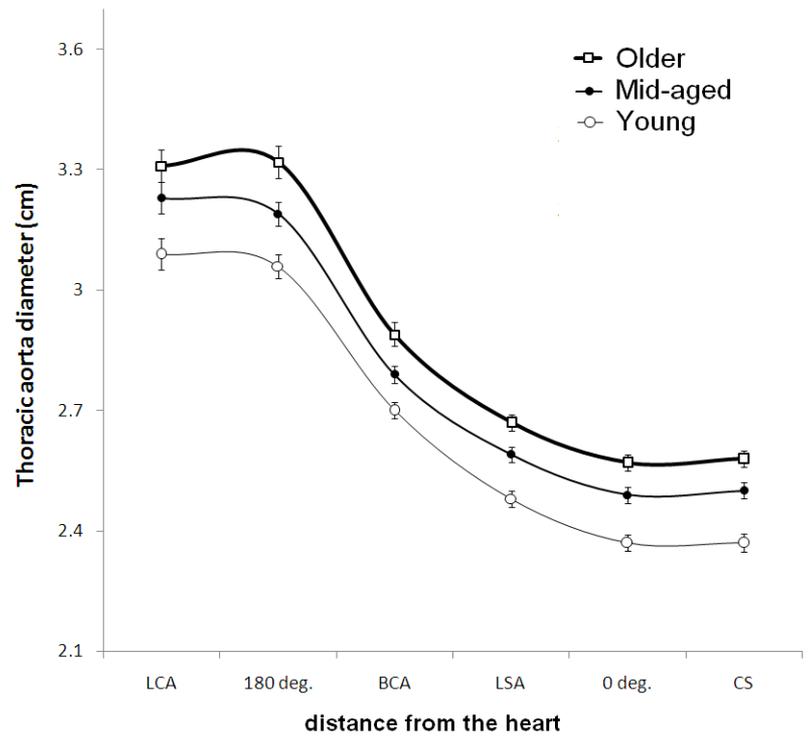
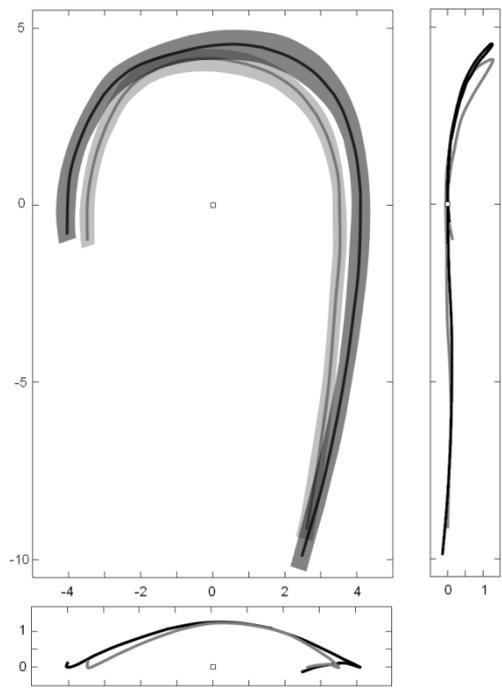
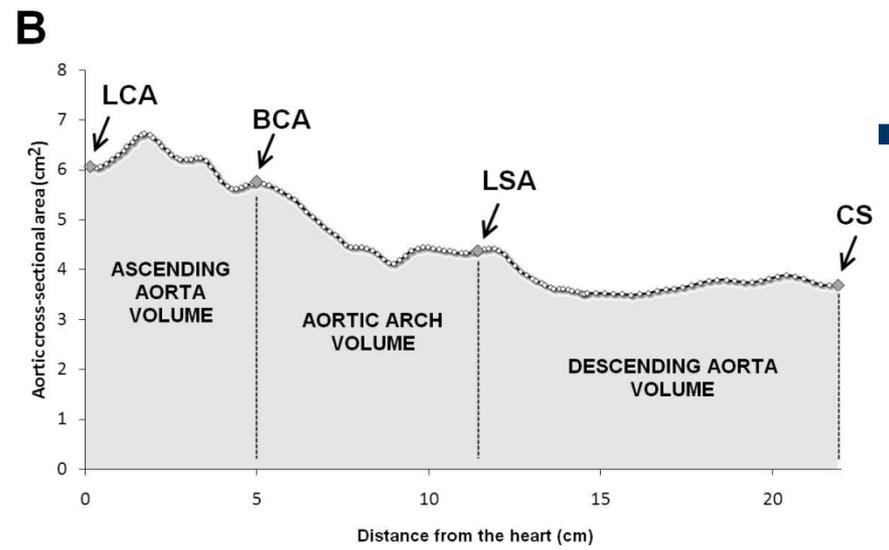
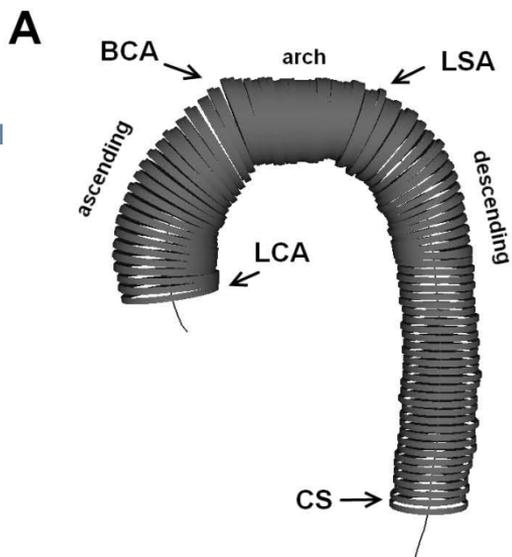
**Stack de imagenes
de MRI (1mm)**



**Extracción de
contorno (B-Splines)**



**Muestreo de puntos
de la geometría
extraida**



Abordaje local-global

Local (level 1):

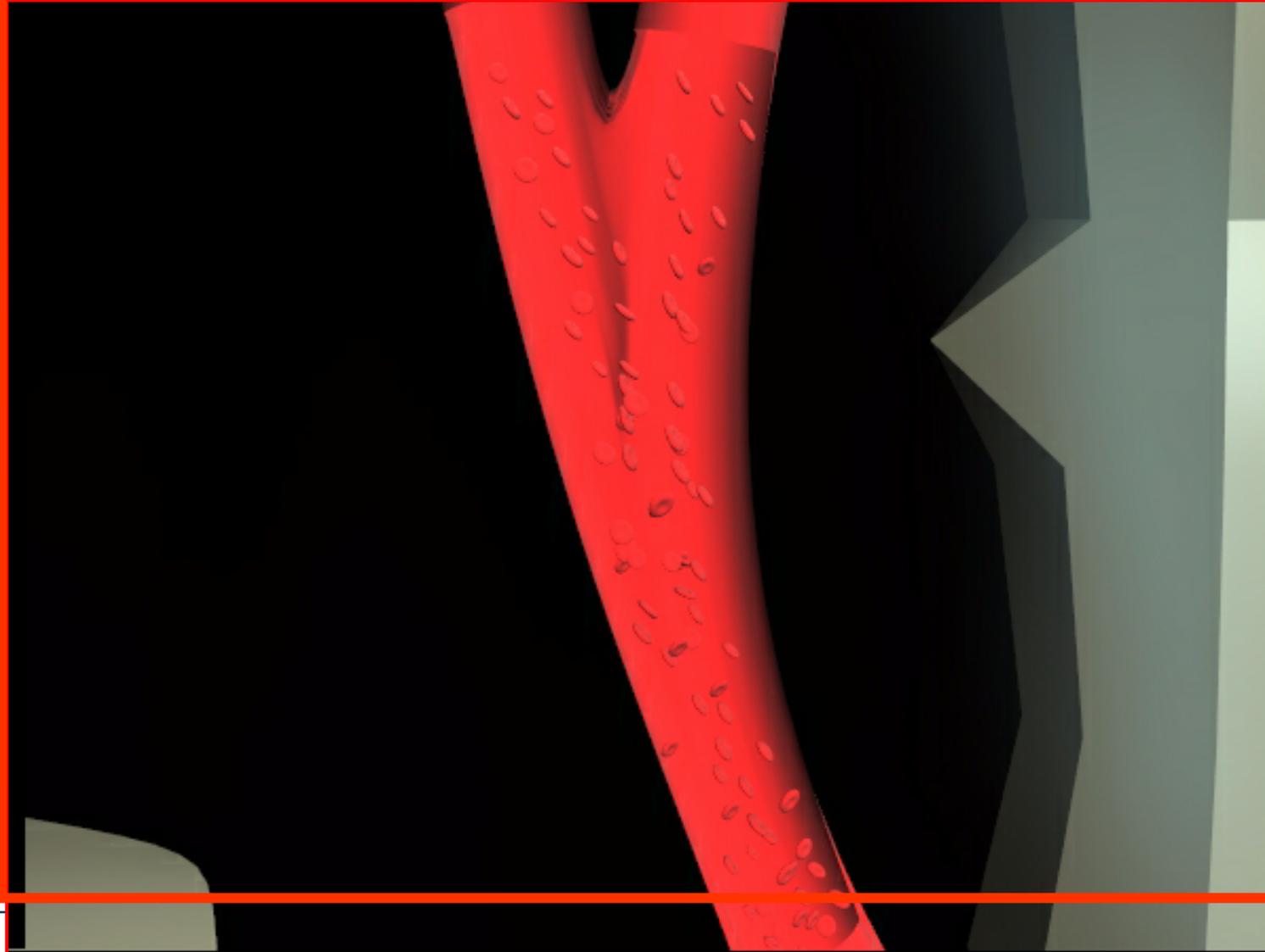
3D modelo de flujo

Global (level 2):

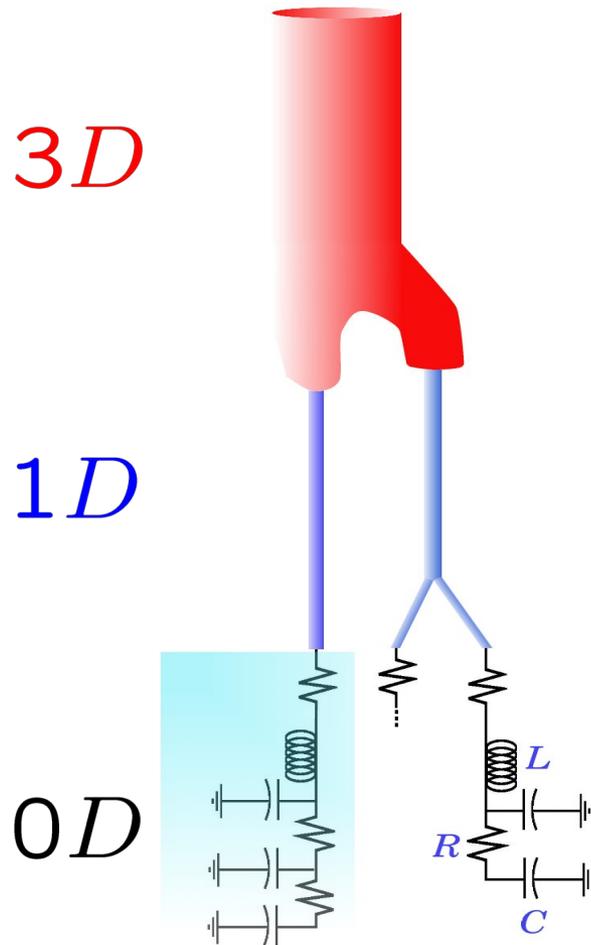
red 1D de arterias y venas

Global (level 3):

0D red capilar



Reducción dimensional por geometría multiescala



3D

3D Navier-Stokes (F) +
3D ElastoDynamics (V-W)

1D

1D Euler (F) +
Algebraic pressure law

0D

0D lumped parameters
(system of linear ODEs)

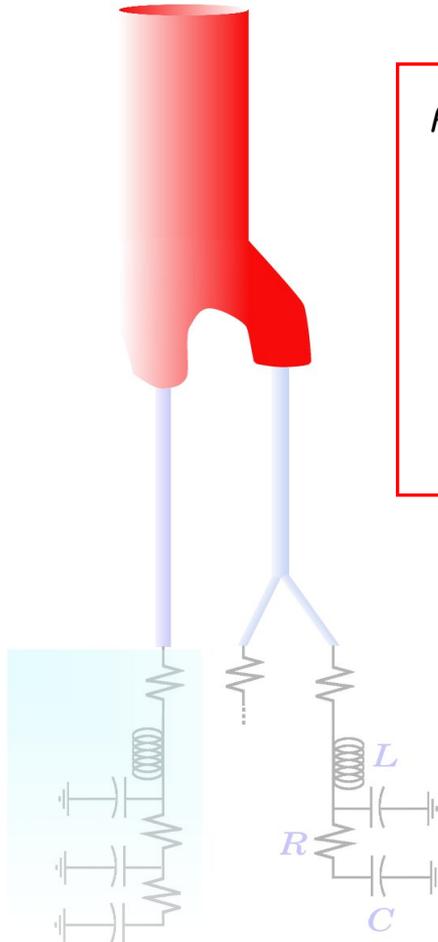
Modelos geometría multiescala

3D Navier-Stokes (F) + 3D ElastoDynamics (V-W)

$$\begin{aligned} \rho_f [\partial_t \mathbf{u} + (\mathbf{u} - \mathbf{w}) \cdot \nabla \mathbf{u}] - \mu \Delta \mathbf{u} + \nabla p &= 0 & \text{in } \Omega_f \\ \operatorname{div} \mathbf{u} &= 0 & \text{in } \Omega_f \end{aligned}$$

$$\partial_{tt} \boldsymbol{\eta} - \operatorname{div} \boldsymbol{\sigma}(\boldsymbol{\eta}) = f(\boldsymbol{\eta}) \quad \text{in } \Omega_w$$

$$\begin{aligned} \boldsymbol{\sigma}(\boldsymbol{\eta}) \cdot \mathbf{n} &= \mathbf{T}(\mathbf{u}, p) \cdot \mathbf{n} & \text{on } \Gamma \\ \mathbf{u} &= \partial_t \boldsymbol{\eta} & \text{on } \Gamma \end{aligned}$$

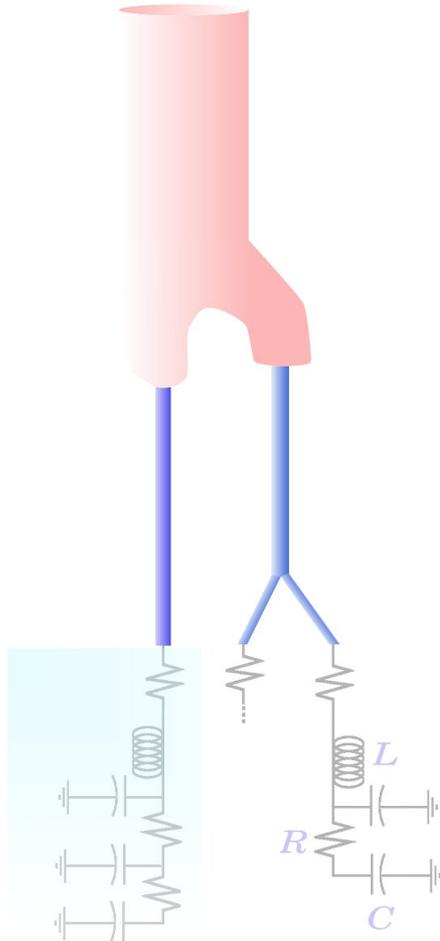


Assume that:

- $u_z \gg u_x, u_y$
- u_z has a prescribed steady profile
- average over axial sections
- static equilibrium for the vessel

Then we obtain a 1D problem.

Modelos geometría multiescala



1D Euler(F) + Algebraic pressure law

$$\begin{aligned}\partial_t A + \partial_x Q &= 0, \\ \partial_t Q + \partial_x \left(\frac{\alpha Q}{A} \right) + \frac{A}{\rho} \partial_x P &= -K_r \frac{Q}{A}, \\ P(A) &= \beta \frac{\sqrt{(A)} - \sqrt{(A_0)}}{A_0}\end{aligned}$$

Assume to

- linearize 1D equations
- consider average internal variables
- relate interface values to averaged ones

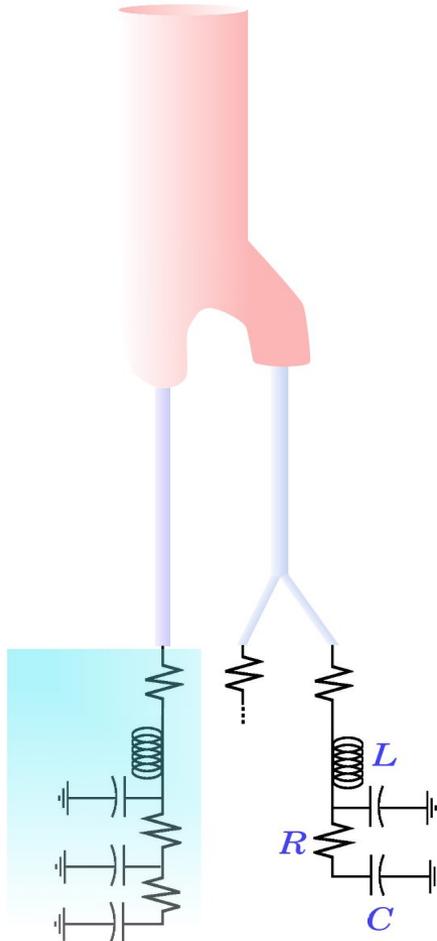
Then we obtain a 0D problem (ODE).

Modelos geometría multiescala

0D Lumped parameters (system of linear ODE's)

$$C \frac{dP_i}{dt} = -(Q_{i+1} - Q_i),$$

$$L \frac{dQ_i}{dt} = -(P_i - P_{i-1}) - RQ_i$$



The analogy

Fluid dynamics	Electrical circuits
Pressure	Voltage
Flow rate	Current
Blood viscosity	Resistance R
Blood inertia	Inductance L
Wall compliance	Capacitance C

- RLC circuits model “large” arteries
- RC circuits account for capillary bed
- Can describe compartments (such as peripheral circulation)

A full geometric multiscale model: 0D-1D-2D (or 3D) coupling

