NATIONAL UNIVERSITY OF PHARMACY
Department of Educational and Information Technologies

BIOPHYSICS, PHYSICAL METHODS OF ANALYSIS

Lecture 6

Hydrostatics and hydrodynamics. Biophysics of the blood system.

Plan of the Lecture

1. Hydrostatics laws.
2. Hydrodynamics laws.
3. Surface and Interfacial Tensions.
4. Hemodynamics.
5. The coefficient of viscosity.
6. Reynolds' number.
7. Erythrocyte Sedimentation Rate.
8. Pulse wave.

Purpose of the lecture is

- to master the basic provisions of hydrostatics and hydrodynamics and blood circulation laws.

Hydrostatics and hydrodynamics
Biophysics of the blood system

Density

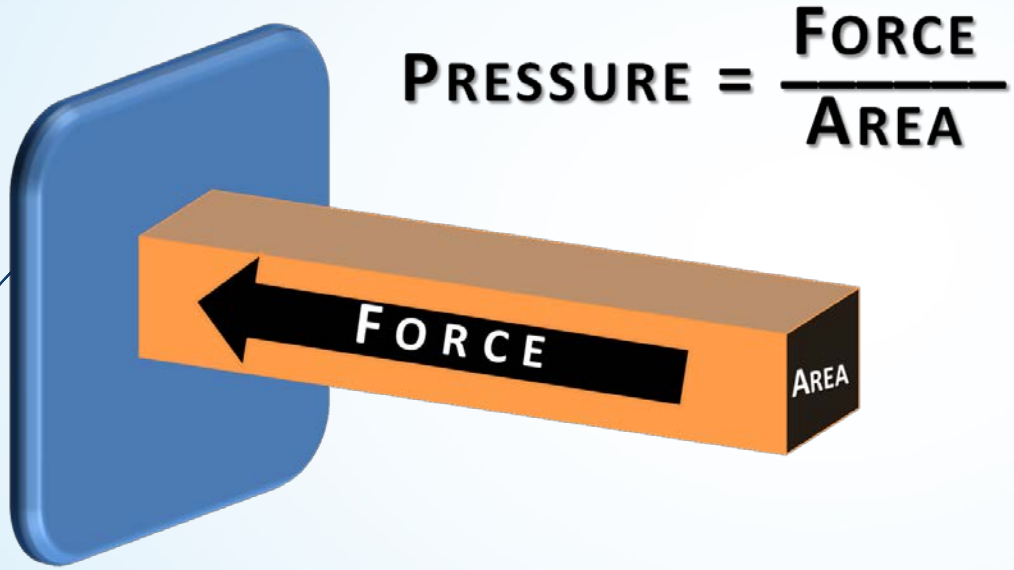
- The **density** of a fluid is the **concentration of mass**

$$density = \frac{mass}{volume} \qquad \rho = \frac{m}{V} \qquad Units\ are\ \frac{kg}{m^3}$$

- Mass = 100 g = 0.1 kg
- Volume = 100 cm^3 = 10^{-4} m^3
- Density = 1 g/cm^3 = 1000 kg m^3

Pressure

- **Pressure** is the concentration of a force – the **force exerted per unit area**

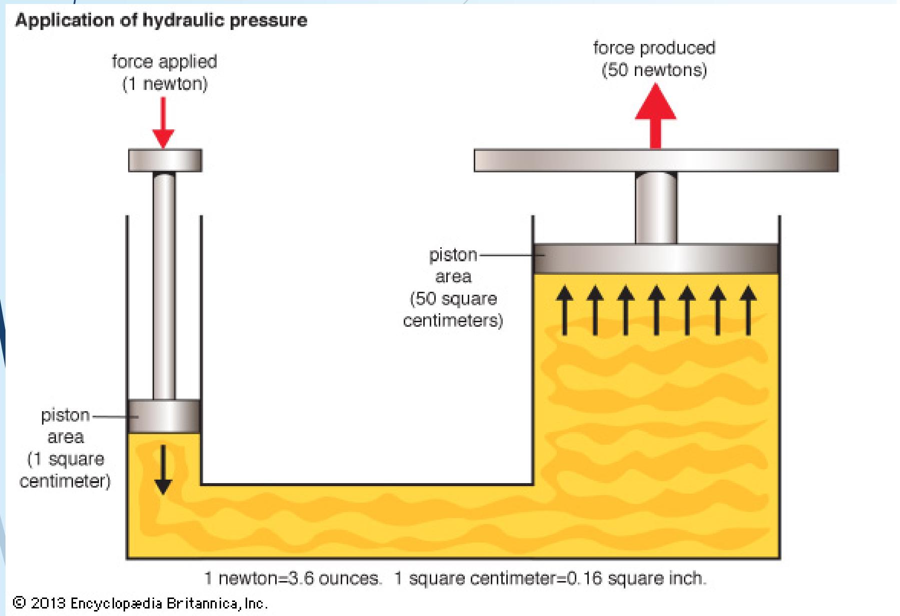


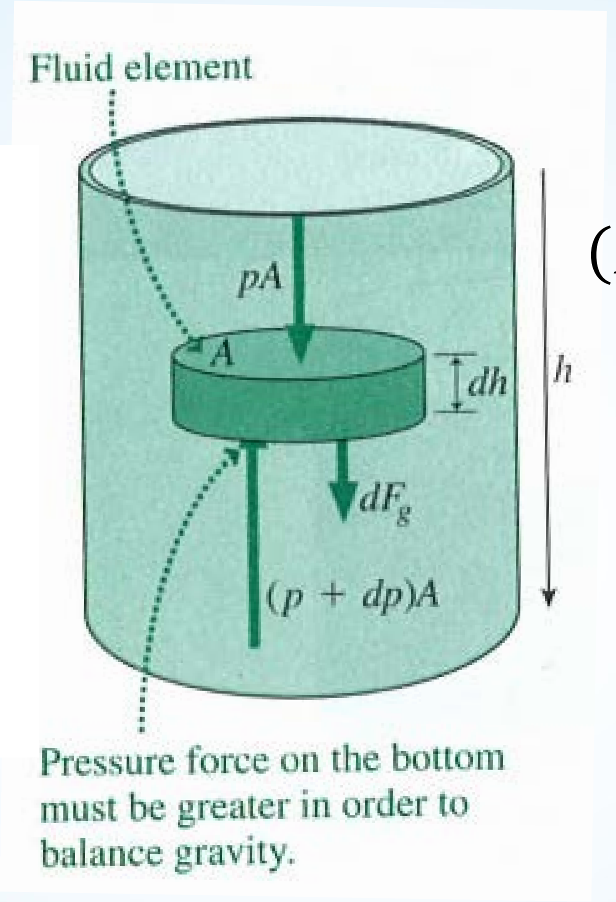
$$p = \frac{F}{A}$$

- **Units of pressure** are N/m^2 or Pascals (Pa) – 1 N/m^2 = 1 Pa

- **Atmospheric pressure** = 1 atm = 101.3 kPa = 1 x 10^5 N/m^2

Hydrostatic Equilibrium with Gravity





Derivation:

$$(P + dP)A - pA = mg$$

$$dP\,A = \rho\,A\,dh\,g$$

$$\frac{dP}{dh} = \rho g$$

$$P = P_0 + \rho g h$$

Pressure in a fluid is equal to the **weight of the fluid per unit area** above it: $P = P_0 + \rho g h$

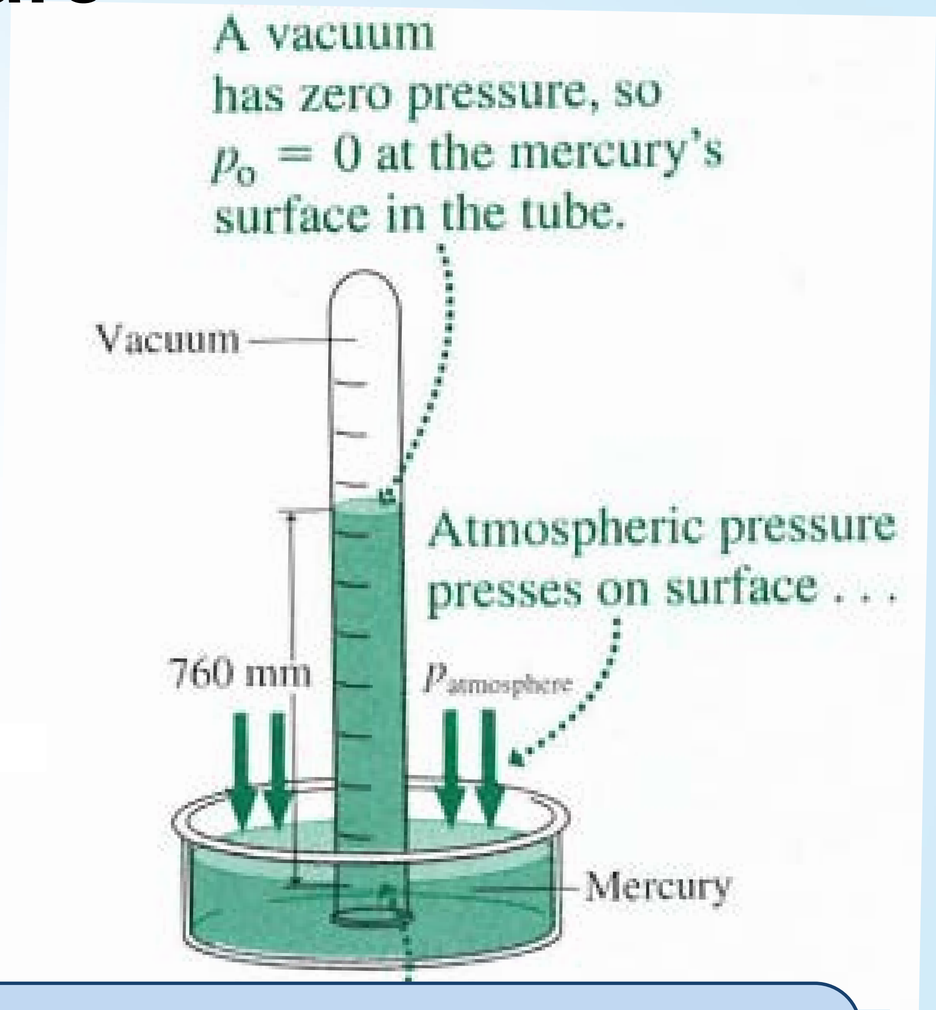
Measuring Pressure

Q. What is height of mercury (Hg) at 1 atm ?

$$\rho_{Hg} = 13.6\ g/cm^3$$

$$P = P_0 + \rho g h \rightarrow h = P/\rho g$$

$$h = \frac{1 \times 105}{1.36 \times 104 \times 9.8} = 0.75\ m$$



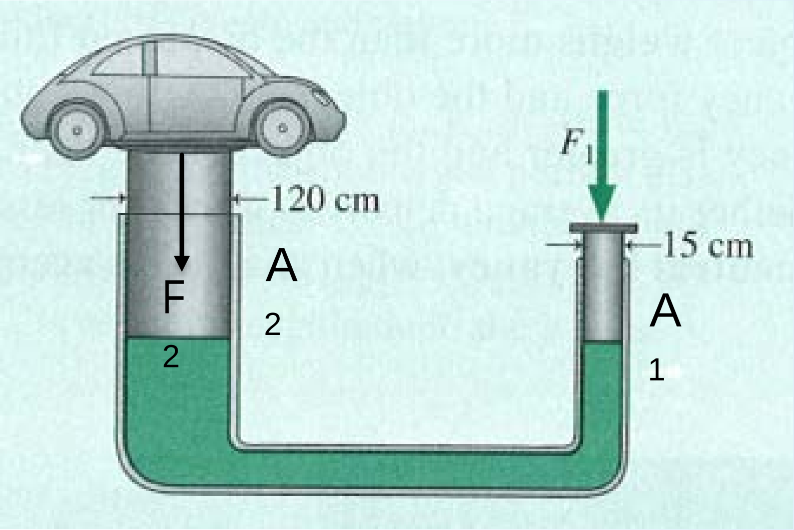
Atmospheric pressure can support a 10 meters high column of water. Moving to **higher density** fluids allows a table top **barometer** to be easily constructed.

$$p = p_0 + \rho g h$$

Pascal's Law

- Pressure force is transmitted through a fluid

Q. A large piston supports a car. The total mass of the piston and car is $3200\ kg$. What force must be applied to the smaller piston ?

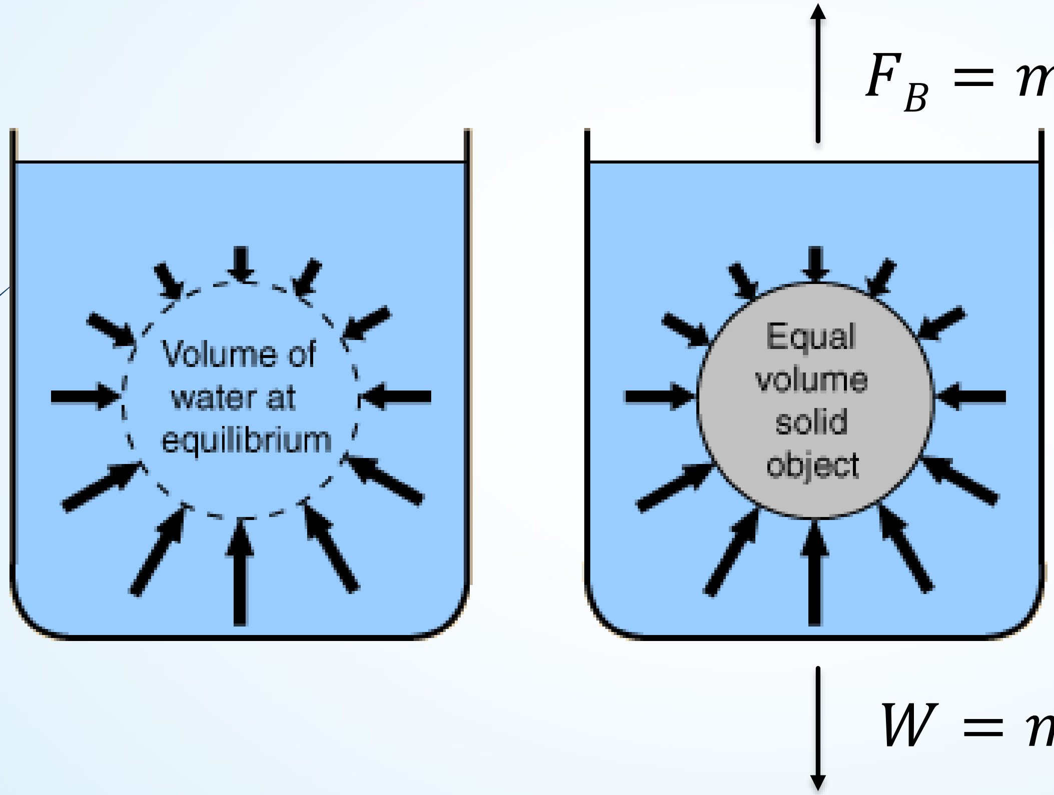


Pressure at the same height is the same! (Pascal's Law)

$$\frac{F_1}{A_1} = \frac{F_2}{A_2} \qquad F_1 = \frac{A_1}{A_2} mg = \frac{\pi \times 0.152}{\pi \times 1.202} \times 3200 \times 9.8 = 490\ N$$

Archimedes' Principle and Buoyancy

Objects immersed in a fluid experience a Buoyant Force!



$$F_B = mwater\ g = \rho_{water} V g$$

$$W = msolid\ g = \rho_{solid} V g$$

The **Buoyant Force** is equal to the weight of the **displaced** fluid !

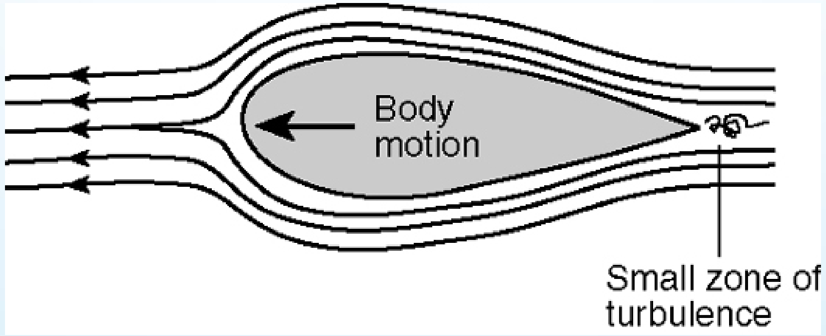
Fluid Dynamics

Let's look at fluids in **motion**

It is important that you understand that an **IDEAL FLUID:**

- **Is non viscous (meaning there is NO internal friction)**
- **Is incompressible (meaning its Density is constant)**
- **Its motion is steady and NON – TURBULENT**

A fluid's motion can be said to be STREAMLINE, or LAMINAR. The path itself is called the streamline. By Laminar, we mean that every particle moves exactly along the smooth path as every particle that follows it. If the fluid DOES NOT have Laminar Flow it has TURBULENT FLOW in which the paths are irregular and called EDDY CURRENTS.



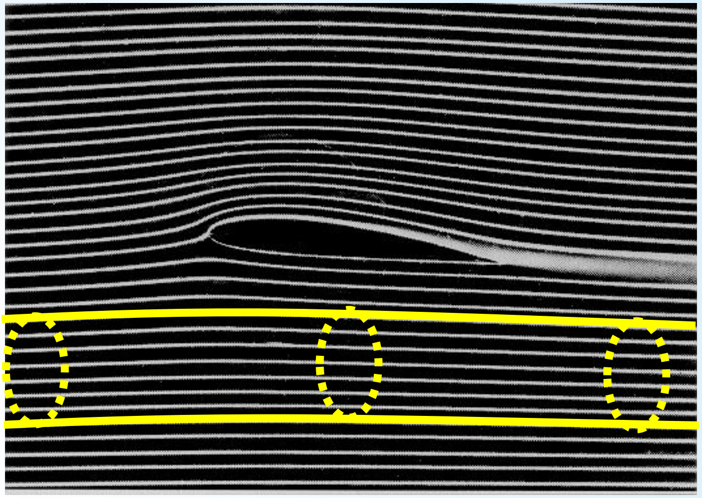
Fluid Dynamics

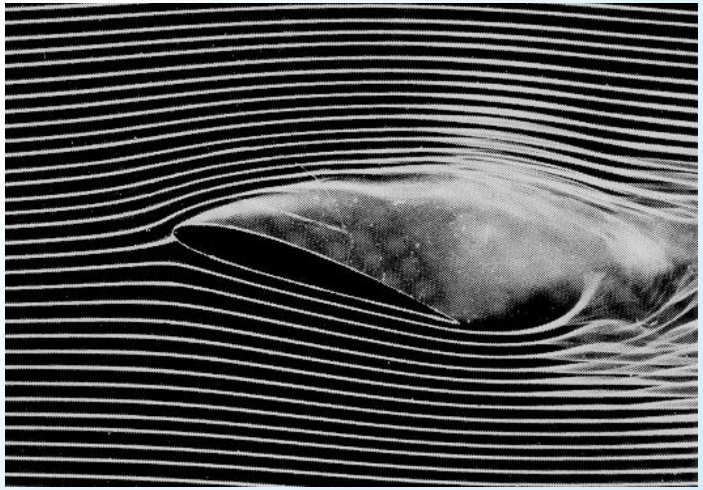
Laminar (steady) flow is where each particle in the fluid moves along a smooth path, and the paths do not cross.

Streamlines spacing measures velocity and the flow is always tangential, for steady flow don't cross. A set of streamlines act as a pipe for an **incompressible** fluid

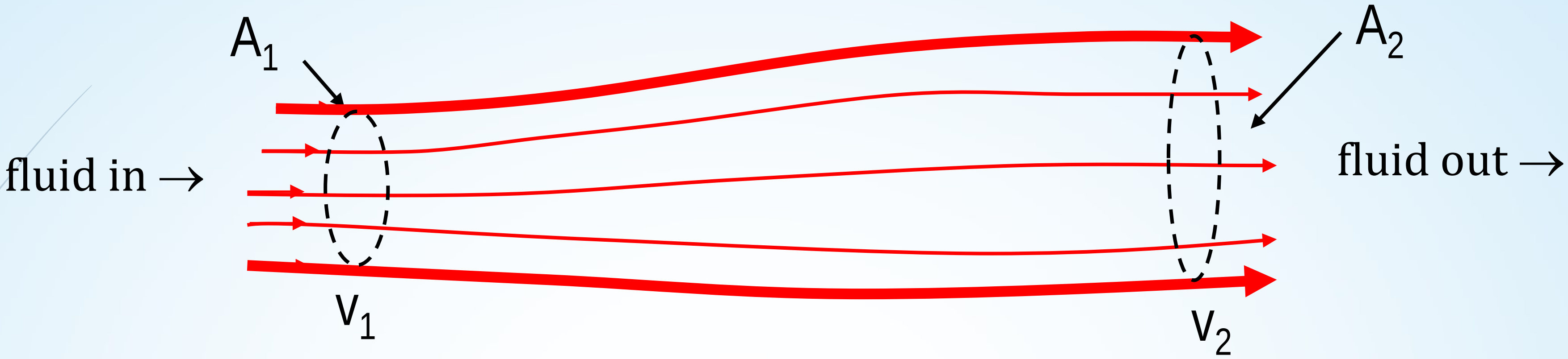
Non-viscous flow – no internal friction (water OK, honey not)

Turbulent flow above a critical speed, the paths become irregular, with whirlpools and paths crossing.

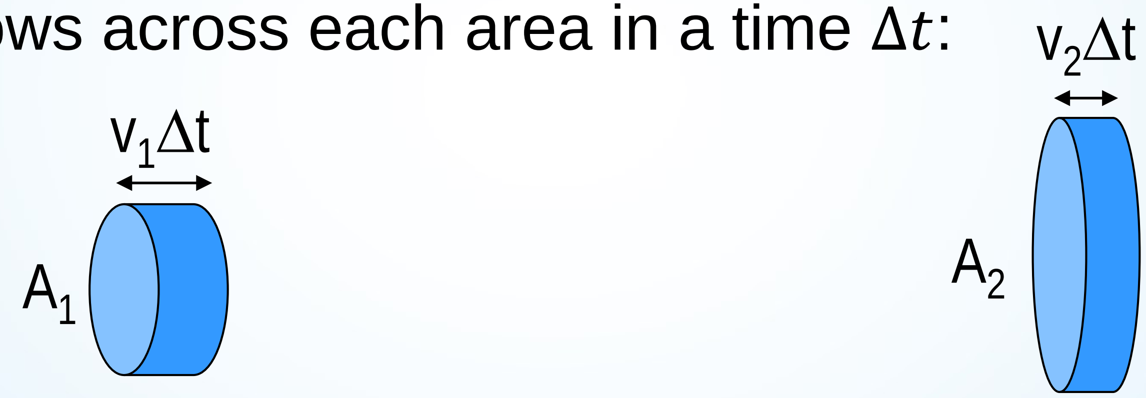




Conservation of Mass: The Continuity Eqn.



Q. How much fluid flows across each area in a time Δt:

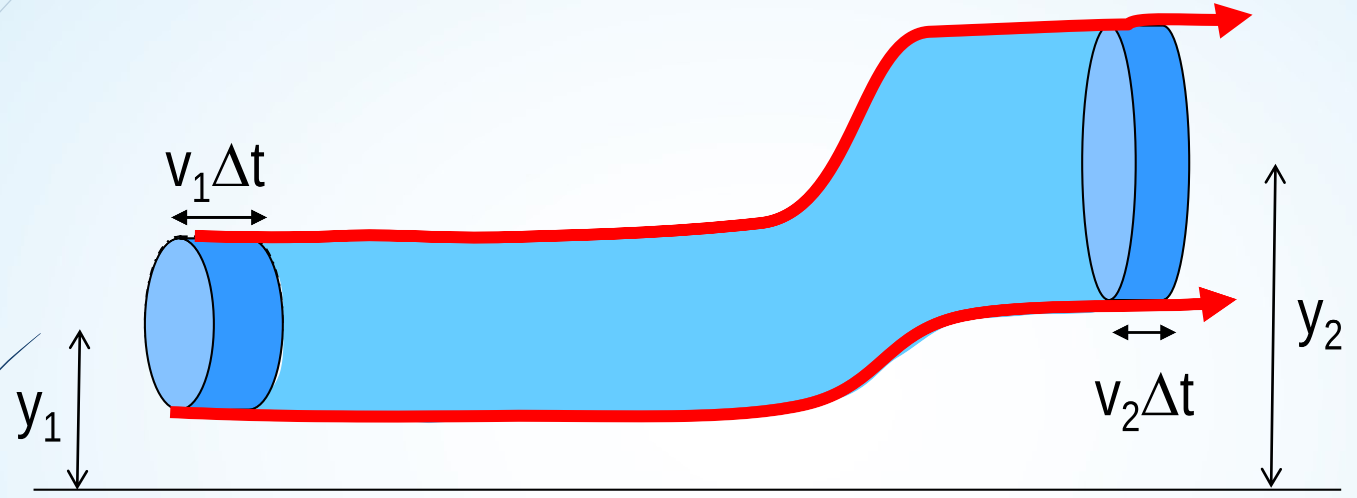


$$\Delta m = \rho V_1 = \rho A_1 v_1 \Delta t \qquad \Delta m = \rho V_2 = \rho A_2 v_2 \Delta t$$

$$flow\ rate: \frac{\Delta m}{\Delta t} = \rho A v \qquad continuity\ eqn: \quad A_1 v_1 = A_2 v_2$$

Conservation of Energy: Bernoulli's Eqn.

What happens to the energy density of the fluid if I raise the ends ?



Energy per unit volume

$$\boxed{p_1 + \tfrac{1}{2}\rho v_1^2 + \rho g y_1 = p_2 + \tfrac{1}{2}\rho v_2^2 + \rho g y_2 = const}$$

Total energy per unit volume is constant at **any** point in fluid.

$$p + \tfrac{1}{2}\rho v^2 + \rho g y = const$$

Interfaces

- ***Interface*** is the boundary between two or more phases exist together
- The properties of the molecules forming the interface are different from those in the bulk that these molecules are forming an ***interfacial phase.***
- Several types of interface can exist depending on whether the two adjacent phases are in solid, liquid or gaseous state.
- Important of Interfacial phenomena in pharmacy:
  - Adsorption of drugs onto solid adjuncts in dosage forms
  - Penetration of molecules through biological membranes
  - Emulsion formation and stability
  - The dispersion of insoluble particles in liquid media to form suspensions.

Surface and Interfacial Tensions

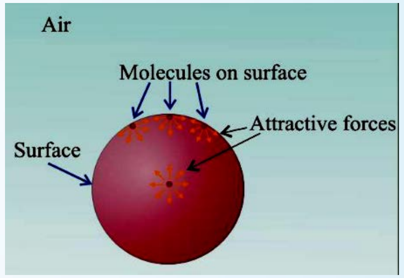
- The net effect is that the molecules at the surface of the liquid experience an inward force toward the bulk.
- Such a force pulls the molecules of the interface together and, as a result, contracts the surface, resulting in a *surface tension*.
- This “tension” in the surface is the force per unit length that must be applied *parallel* to the surface so as to counterbalance the net inward pull.
- *Interfacial tension* is the force per unit length existing at the interface between two immiscible liquid phases and.
- the surface and interfacial tensions, have the units of N/m.
- The surface tensions of most liquids decrease almost linearly with an increase in temperature, that is, with an increase in the kinetic energy of the molecules.

The work W required to create a unit area of surface is known as ***SURFACE FREE ENERGY/UNIT AREA*** **(N/m^2)**

Its equivalent to the surface tension γ

Thus the greater the area **A** of interfacial contact between the phases, the greater the free energy.

$$\boxed{W = \gamma \Delta A}$$



For equilibrium, the surface free energy of a system must be at a minimum.

Thus Liquid droplets tend to assume a spherical shape since a sphere has the smallest surface area per unit volume.

Wetting

- Adsorption at solid surfaces is involved in the phenomena of wetting and detergency.
- When a liquid comes into contact with the solid, the forces of attraction between the liquid and the solid phases begin to play a significant role.
- In this case, the behavior of the liquid will depend on the balance between the forces of attraction of molecules in the liquid and the forces of attraction between the liquid and the solid phases.
- Wettability of tablet surfaces influences disintegration and dissolution and the subsequent release of the active ingredient(s) from the tablet.

According to the nature of the liquid and the solid, a drop of liquid placed on a solid surface will adhere to it or no. which is the wettability between liquids and solids.

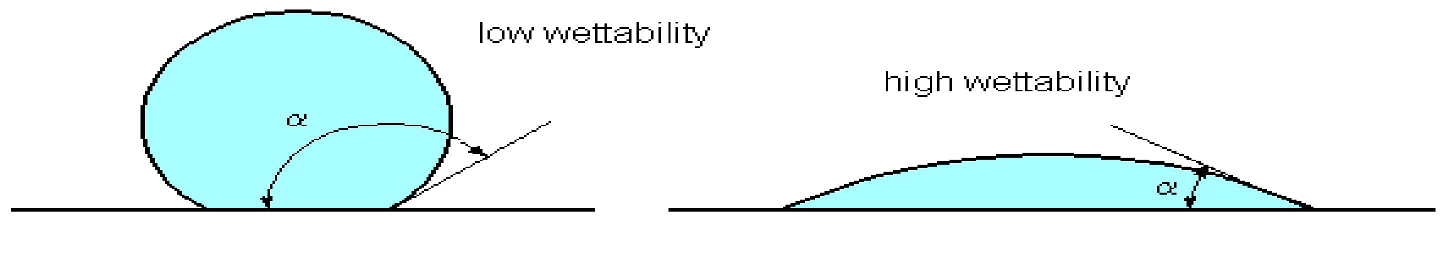
When the forces of adhesion are greater than the forces of cohesion, the liquid tends to wet the surface and vice versa.

Place a drop of a liquid on a smooth surface of a solid. According to the wettability, the drop will make a certain angle of contact with the solid.

A contact angle is lower than 90°, the solid is called wettable

A contact angle is wider than 90°, the solid is named non-wettable.

A contact angle equal to zero indicates complete wettability.



The contact angle of a liquid with a solid is used as wettability index. For $\alpha < 90°$ the liquid wet the wall (eg: water on glass), for $\alpha > 90°$ the liquid does not wet the wall (eg: mercury on glass). If $\alpha = 0°$ the liquid perfectly wet the wall.

➢If a capillary tube of inside radius =r immersed in a liquid that wet its surface, the liquid continues to rise in the tube due to the surface tension, until the upward movement is just balanced by the downward force of gravity due to the weight of the liquid

➢*<u>The upward component of the force</u>* resulting from the surface tension of the liquid at any point on the circumference is given *by:*

$$a = \gamma \cos \Theta$$

Thus the total upward force around the inside circumference of the tube is

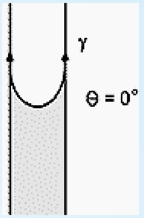
$$a = 2\,\pi\, r\, \gamma \cos \Theta$$

Where

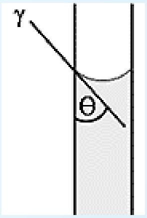
Θ = *the contact angle between the surface of the liquid and the capillary wall*

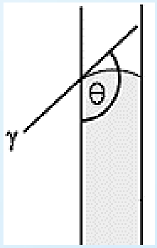
$2\,\pi\, r$ = the inside circumference of the capillary.

For water the angle Θ is insignificant, i.e. the liquid wets the capillary wall so that $\cos \Theta$ = unity



Cont. angle
water and glass





Cont. angle
Mercury and glass

The downward force of gravity

(mass x acceleration) is given by

$$\pi r^{2} h (p - p^{o}) g + w$$

Where:

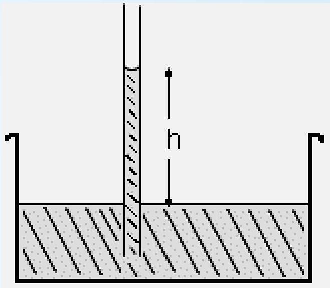
πr^{2} = the cross-sectional area

h = the height of the liquid column to the lowest point of the meniscus

$(p - p\,o)$ = the difference in the density of the liquid p and its vapor po

g = the acceleration of gravity

w = the weight of the upper part of the meniscus.



At Maximum height, the opposing forces are in equilibrium

$$2 \pi r \gamma \cos \Theta = \pi r^{2} h (p - p\,o) g + w$$

$p\,o$, Θ and w can usually be disregarded

Hence the surface tension can be calculated.

$$2 \pi r \gamma = \pi r^{2} h p g$$

$$\gamma = 1/2\, r h p g$$

Hemodynamics

(of Greek haima – blood, dynamis – force)

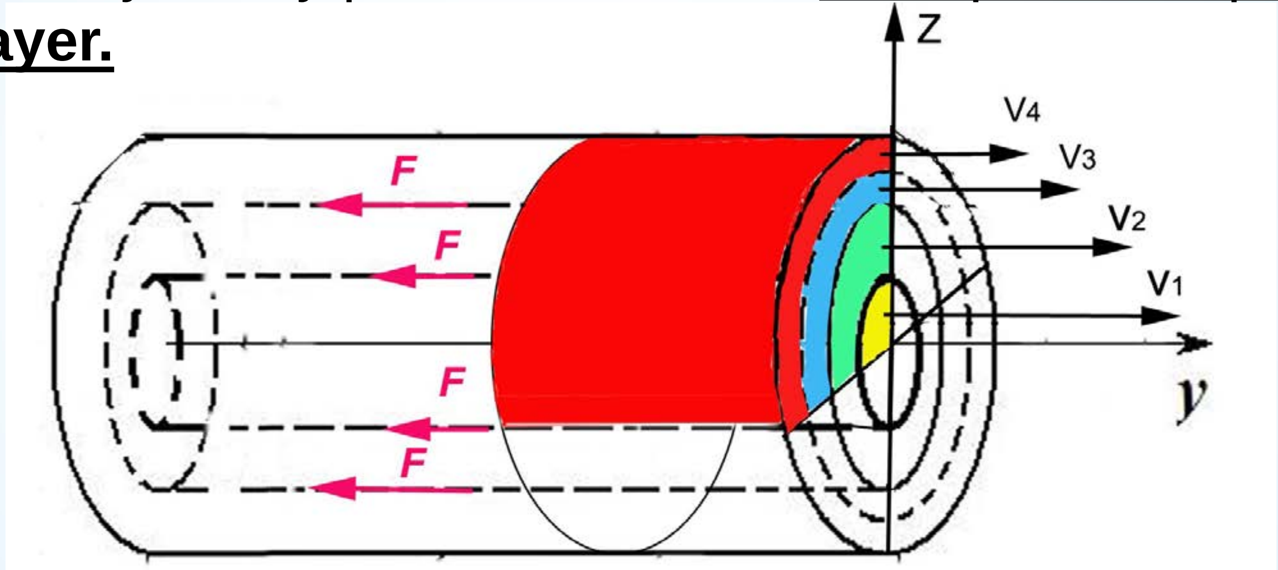
is a science, which appeared and is developing on the crossroads of hydromechanics and biology, it studies blood movement in the closed vascular system of the human organism taking into account morphological structure of blood, its physical-chemical characteristics, specific features of vascular wall, dynamics of the live system with applied adoption of the hydrodynamic postulates.

1. The gradient of the hydrostatic pressure in various segments of the vascular system that is formed due to the pumping function of the myocardium.
2. Rheological properties of blood as the dispersion of the forming elements with properties of the non-Newton liquids.
3. Blood carrying vessels as the viscous-elastic tubes, whose properties (<u>geometrical</u> - size, branching and <u>physical</u> - viscosity, elasticity, penetration) vary in diameter and length.

Real liquids

Real liquids are viscous liquids.

Liquid viscosity is a physical phenomenon of the existence of viscous friction forces in the stream of a real liquid. When real liquid flows, its speed in various points of the stream is different. But there are parts of liquid where velocity of any point is the same. Such parts of liquid are called **liquid layer.**



In a real liquid, forces of internal or viscous friction ***F*** arise from the relative movement of layers.

The internal friction is caused by the molecule momentum (pulse) transfer from one layer to another.

Newton's formula (1687) for the viscous friction force

$$F = \eta \frac{dv}{dz} S$$

F - is the force of viscous friction between the liquid layers appearing as a result of their relative shift

η **-** is the proportionality factor, called ***coefficient of viscosity*** (*or* ***dynamic viscosity***) of the liquid

***V* -** is the velocity of the liquid layer movement

z - is the coordinate perpendicular to the layer contact boundary

$\frac{dv}{dz}$ - is a velocity gradient absolute value, which characterizes the degree of change in the velocity of the liquid flow in transfer from one layer to another

s - is the area of the liquid layers contact surface

The coefficient of viscosity

The coefficient of viscosity η characterizes

the <u>layer displacement resistance</u>

<u>The coefficient of viscosity η in the metric SI is measured in Pascal-seconds (Pa·s)</u>

For example, water at temperature ***20℃*** has the viscosity of 1,005·10^{-3} Pa·s; blood viscosity value is (4-5) ·10^{-3} Pa ·s at temperature 37°C.

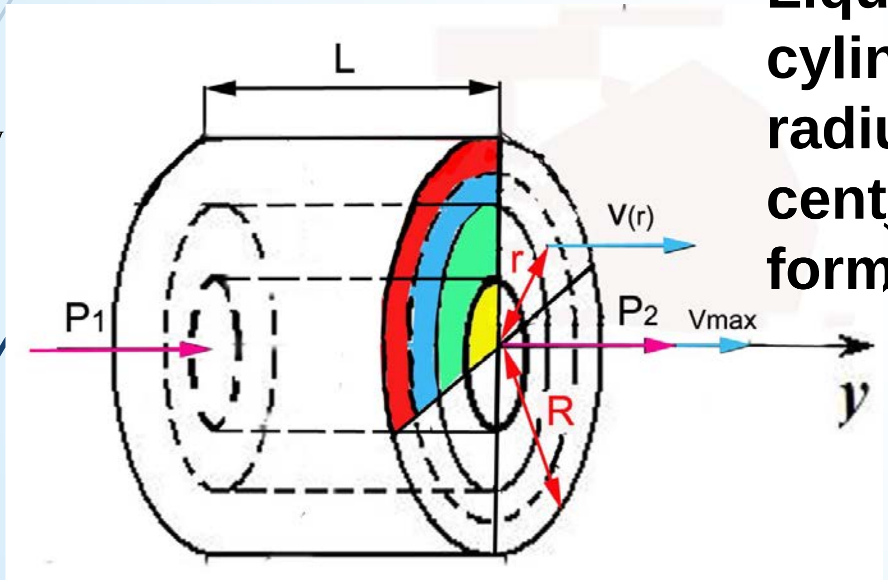
Newtonian and non-Newtonian liquids

A lot of complex liquids, inhomogeneous in the composition of particles forming the liquid do not comply with the Newton's formula. That is why real liquids are subdivided into Newtonian and non-Newtonian liquids with respect to their viscosity properties.

Newtonian	non-Newtonian
Coefficient of viscosity depends only on the kind of liquid and its temperature **As the temperature increases, the coefficient of viscosity decreases**	***Coefficient of viscosity also depends on the values, characterizing the conditions of the liquid flow.*** **For example, on the speed gradient.**

Dependence of liquid layer speed in a portion of a cylindrical pipe on the distance from the center of the pipe (for Newtonian liquids)

The speed of the liquid layers increases as the distance from the wall increases. The layer touching the walls is considered to be a rest.

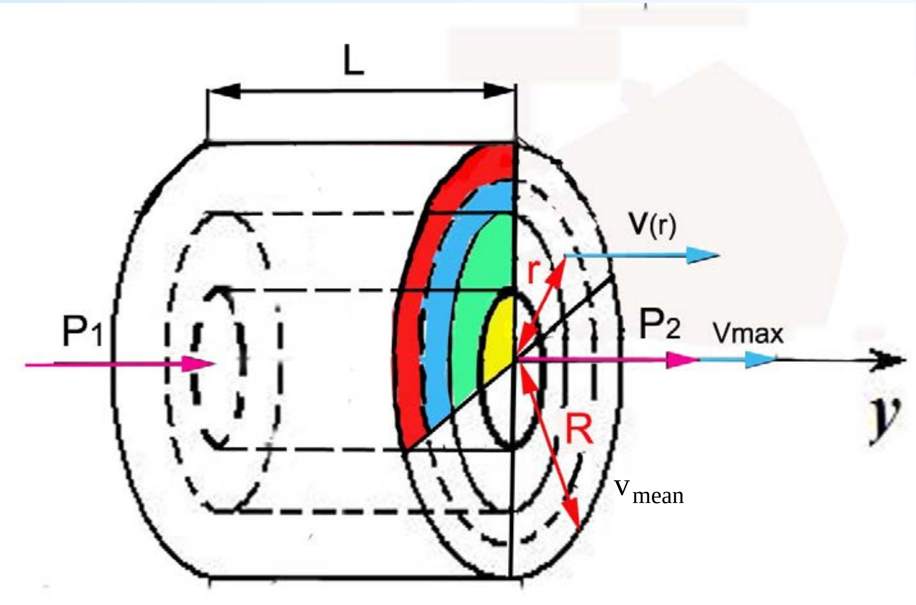


Liquid layer speed in a portion of a cylindrical pipe with the length *L* and radius *R* on the distance (r) from the center of the pipe is described by the formula:

$$v = \frac{P_1 - P_2}{4\eta L}(R^2 - r^2)$$

***P*$_1$ - is the pressure at the beginning of the pipe section; *P*$_2$ - is the pressure at the end of the pipe section.**

The particles moving along the pipe axis (r = 0) have the maximum velocities (v_{max})



$$V_{max} = \frac{P_1 - P_2}{4\eta L} \cdot R^2$$

The mean value of speed (v_{mean}) for a pipe section is calculated by the formula

$$V_{mean} = \frac{V_{max}}{2} = \frac{P_1 - P_2}{8\eta L} \cdot R^2$$

POISEUILLE FORMULA

Poiseuille formula calculates the volume of real liquid passing through cylindrical pipe cross section per unit time for a stationary laminar flow

$$Q = \frac{(P_1 - P_2)\pi R^4}{8\eta L}$$

Q **– is the volume of liquid passing through the cross-section of a pipe per unit time at stationary laminar flow**

P_1 **– is pressure at the beginning of the pipe section**

P_2 **– is pressure at the end of the pipe section**

R **– is radius of cylindrical pipe**

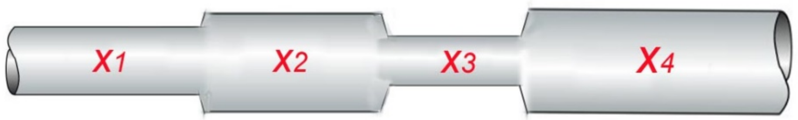
L – is **length of cylindrical pipe**

Analogy between the processes of the liquid flow in the pipe and electric current flow in a conductor

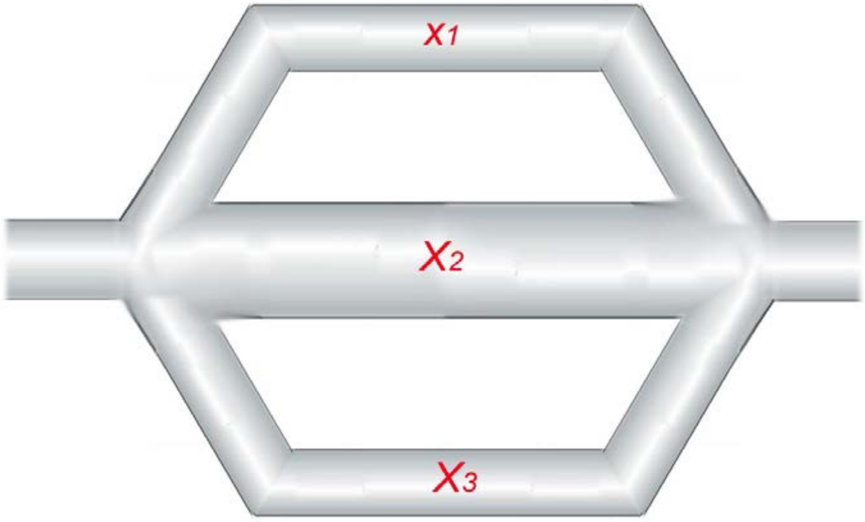
Liquid flow in the pipe	Electric current flow in a conductor
Q - is volume velocity	I – is current strength
P_1 - P_2 - is pressures difference	φ_1 - φ_2 - is potentials difference
X - is hydraulic resistance	R - is electrical resistance
$X = \frac{P_1 - P_2}{Q} = \frac{8\eta L}{\pi R^4}$	$R = \frac{\varphi_1 - \varphi_2}{I}$

The rules of calculation of total hydraulic resistance

If the pipes with hydraulic resistance X_1, X_2, ... , X_n are connected *in series*, or *in parallel* the total hydraulic resistance (X_{total}) for the systems will be the following:



$$X_{total} = X_1 + X_2 + ... + X_n$$



$$\frac{1}{X_{total}} = \frac{1}{X_1} + \frac{1}{X_2} + ... + \frac{1}{X_n}$$

in series ***in parallel***

Reynolds' number (Re)

The ability of liquids to transform from laminar to turbulent flow is characterized by *Reynolds' number* (*Re*).
For non-compressible liquid flowing in a *cylindrical pipe* this number can be calculated as follows

$$Re = \frac{\rho v D}{\eta}$$

The value of the Reynolds' number, at which transition from laminar to turbulent flow takes place, is called *critical Reynolds' number (Re_{cr}).* For smooth cylindrical pipes it approximately equals $Re_{cr} \approx 2300$.

***If Re < Re_{cr}* – laminar flow**

***If Re > Re_{cr}* – turbulent flow**

Kinematic viscosity

Ability of the liquid to change the flow pattern depends on its liquid dynamic viscosity η and liquid density, or, to be more exact, on their ratio. That is why a notion of *kinematic viscosity* should be introduced to provide a more exact characteristic of liquids from this point of view.

Kinematic viscosity (ν) is equal to

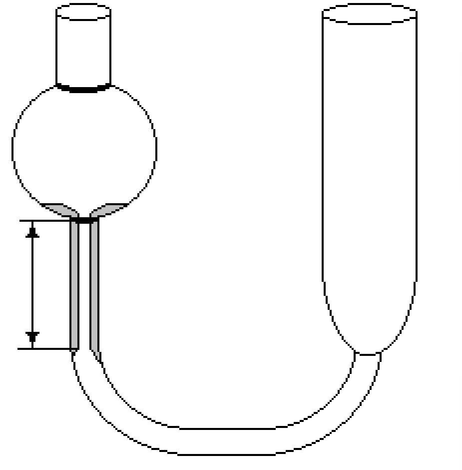
$$\nu = \frac{\eta}{\rho}$$

Kinematical viscosity in SI is measured in [$m^2 \cdot s^{-1}$]
In this case the *formula for the Reynolds'* number will be

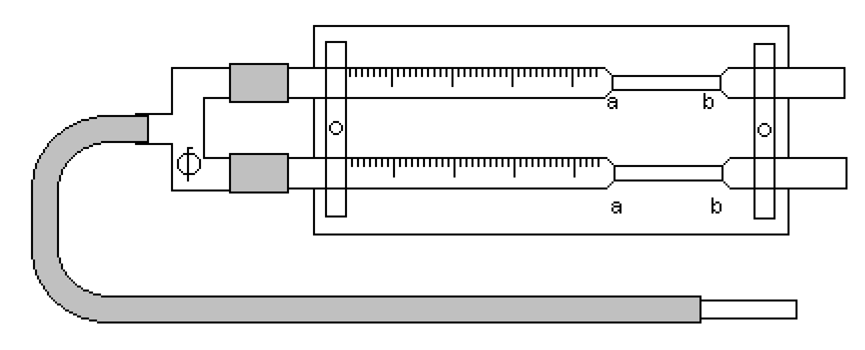
$$Re = \frac{VD}{\nu}$$

Viscosimetry

Viscosity measurement is called viscosimetry, and the instruments for measuring of viscosity are called ***viscometers***.



Ostvald's viscometer



Hess' viscometer

used in medicine for measurement of blood viscosity

Foundations of Biorheology

The formulae of hydrodynamics are mainly used to describe Newtonian liquids only. That is why to describe to the flow of blood (a non-Newtonian liquid) through blood vessels, notions and formulae of rheology are used.

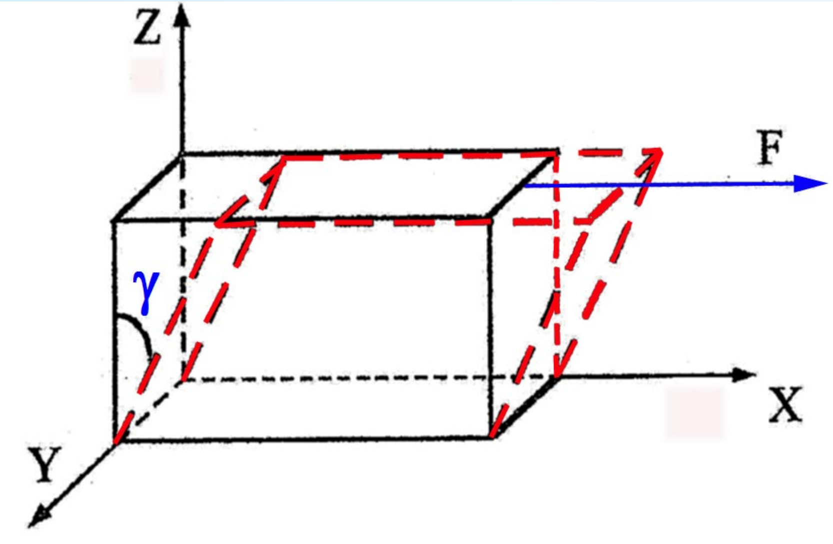
Rheology **is a field of physics, studying deformations and fluidity of real liquids and media.**

In ***biorheology*** **the blood-flow is regarded as a shear deformation.**

The basic quantitative parameters of biorheological processes are

- **shear stress** (τ)
- **shear angle** (γ)
- **shear velocity** $(\dot{\gamma})$

Rheological model of liquids flow



- Applied force (F) causes elastic shear deformation and the parallelepiped undergoes deformation
- Amount of deformation is characterized by shear angle γ

- The horizontal layers of parallelepiped are shifted relatively to each other in the direction of the OX axis
- Under such conditions an elastic force F is created in the layers contact surfaces
- This elastic force F causes the shear stress between the layers contact surfaces, it is calculated by formula $\tau = \frac{F}{S}$

Rheological model of liquids flow

- ***In the process of layer shifting the shear angle γ increases***
- **The time derivative of γ is called *shear velocity* and is designated as $\dot{\gamma}$**

Shear velocity in hydrodynamics called as velocity gradient absolute value.

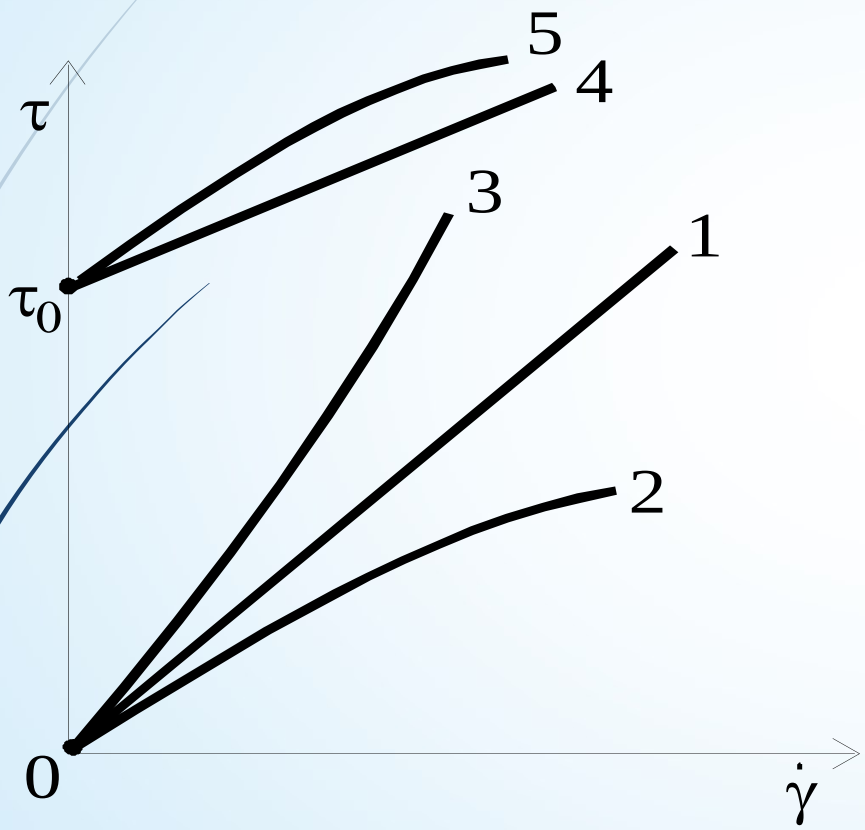
$$\dot{\gamma} = \frac{dv}{dz}$$

The Newton's formula for the force of viscous friction in terms of rheology is:

$$\tau = \eta\dot{\gamma}$$

Flow curves

The graph showing the dependence of τ on $\dot{\gamma}$ is called flow curve



1 – Newtonian liquid

2 – Pseudo-plastic liquid

3 – Dilatant liquid

4 and **5 – Viscous-plastic liquids**

τ_0 - **is called limit shear stress**

Model describing the behavior of viscous-plastic liquids at different shear rates

Shvedov-Bingham's model	Caisson's model
at high values of shear rate	*at low values of shear rate*
$\tau = \tau_0 + k\,\dot{\gamma}$	$\sqrt{\tau} = \sqrt{\tau_0} + k\sqrt{\dot{\gamma}}$
where k is a coefficient equidimensional with viscosity	***where k is a coefficient called as Caisson's coefficient***

Parameters	In hydrodynamics	In hemodynamics
Tube's wall	unchangeable hard	elastic, pulsating
Porosity of the tube's wall	absent	there are effects of penetration of the liquid through the vacular wall in small vessels
The tube's content	constant density of the real liquid	variable density on non-Newton liquids
a pump	constant parameters of the pump functioning	pumping function of the myocardium is the variable quantity
Segments of a water supply system	intake and outlet main lines	third link apppears - capillary bed

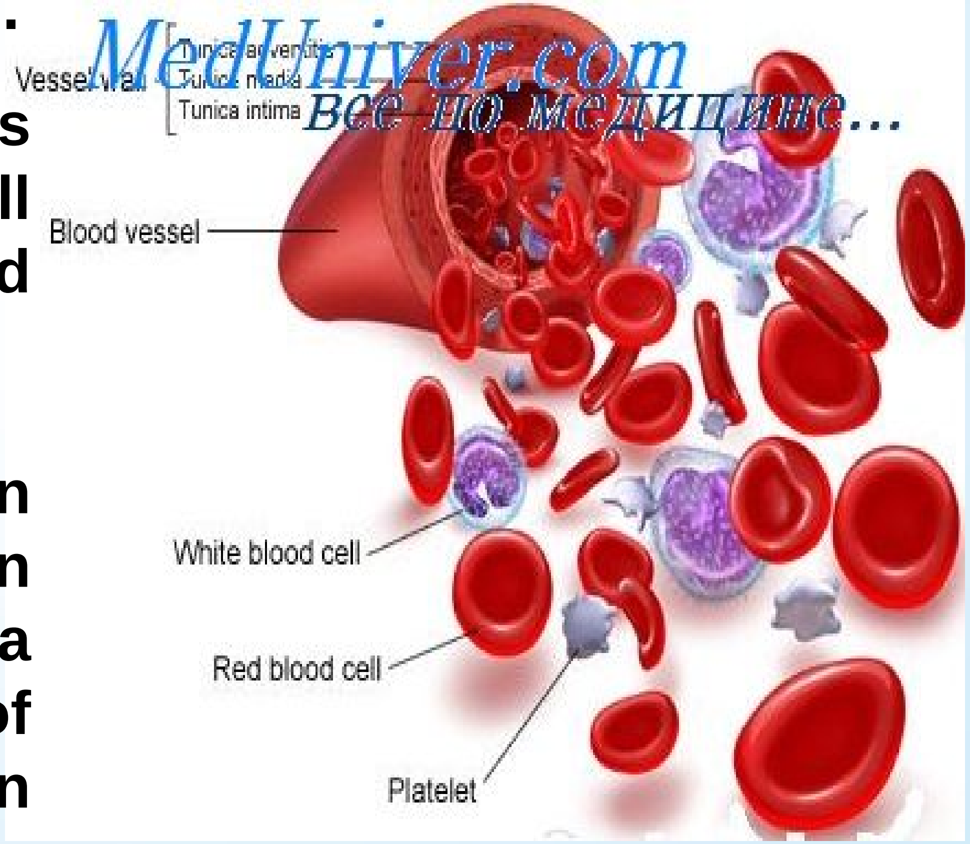
Parameters	In hydrodynamics	In hemodynamics
Autoregulating mechanisms	absent or are controlled outside	present
character of movement	pressure	is not a pressure one in the classical underestanding, it differs by the possibility of formation of a blood depot due to stretching of vascular walls
Dempfers	almost absent	present
Dynamic system	absent	present
coresponding to the gravitation forces	constant	constantly variable due to variations of body positions

Rheological Properties of Blood

Blood is a liquid with salient non-Newtonian properties. Blood is one of the viscous-plastic liquids.

At the same time, blood plasma, though possessing non-Newtonian properties, exhibits these properties to a much less extent compared to whole blood.

Thus, non-Newtonian properties of blood are linked first of all with the blood cells in blood plasma.

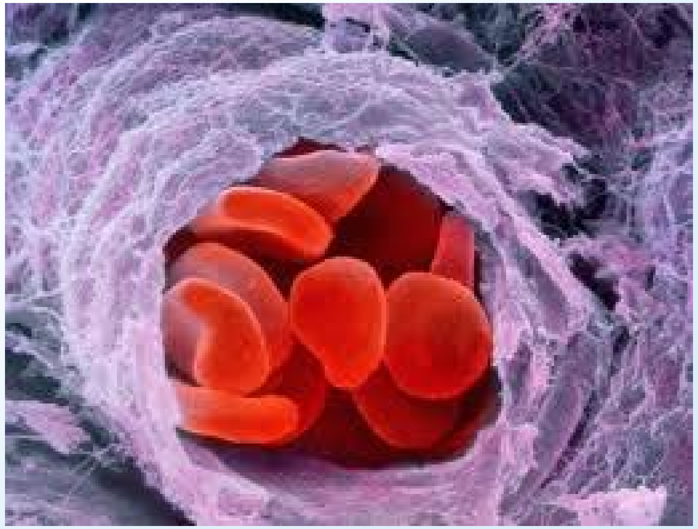


Blood serum is a Newtonian liquid, therefore, non-Newtonian properties of the blood plasma being due to macromolecules of organic compounds dissolved in it, first of all, albumens.

Rheological Properties of Blood

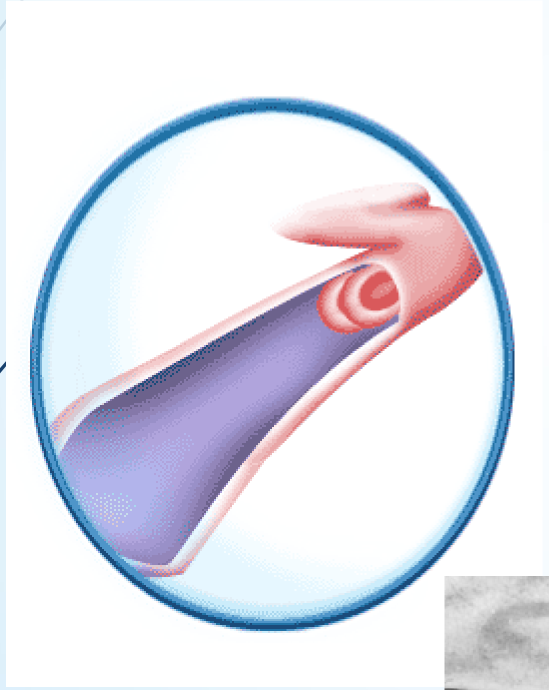
Erythrocytes as a blood cells are most of all responsible for the non-Newtonian properties of blood. This can be accounted for by a number of factors:

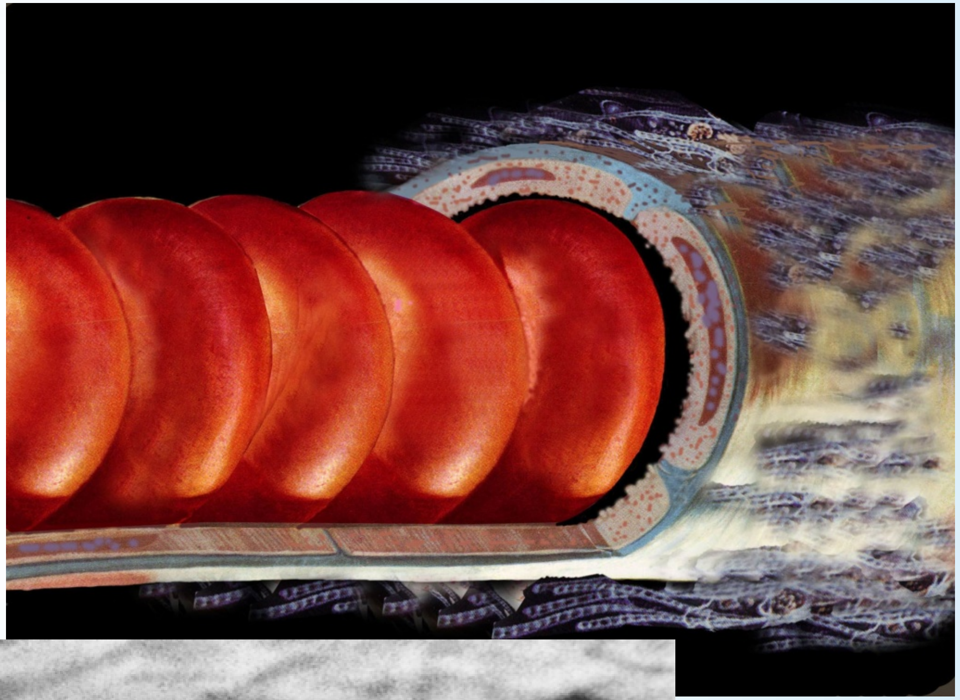
1) a large amount of erythrocytes in blood (of the order of 5 mln per 1 mm^3);

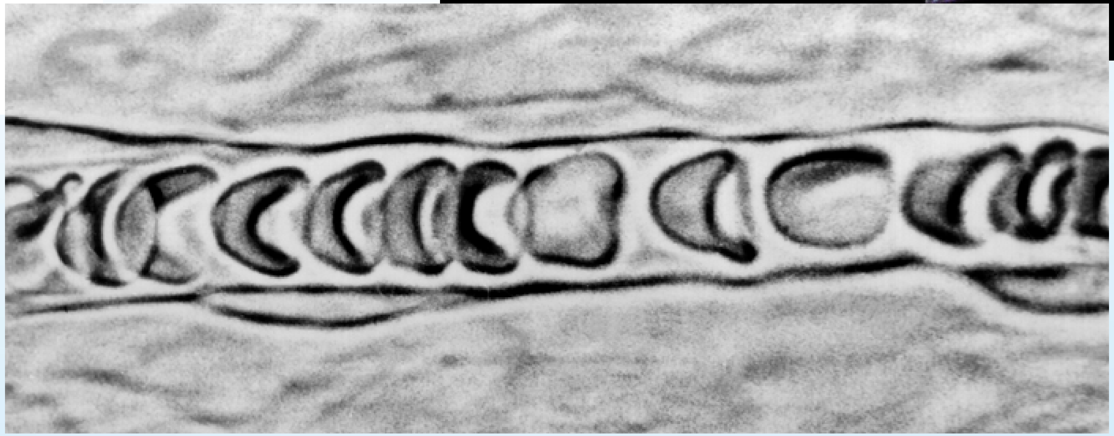


2) the capacity of undergoing deformation when passing through small-diameter vessels;

3) the capacity of erythrocytes to form aggregate arrangements, the size of which depends on the conditions of the blood flow.

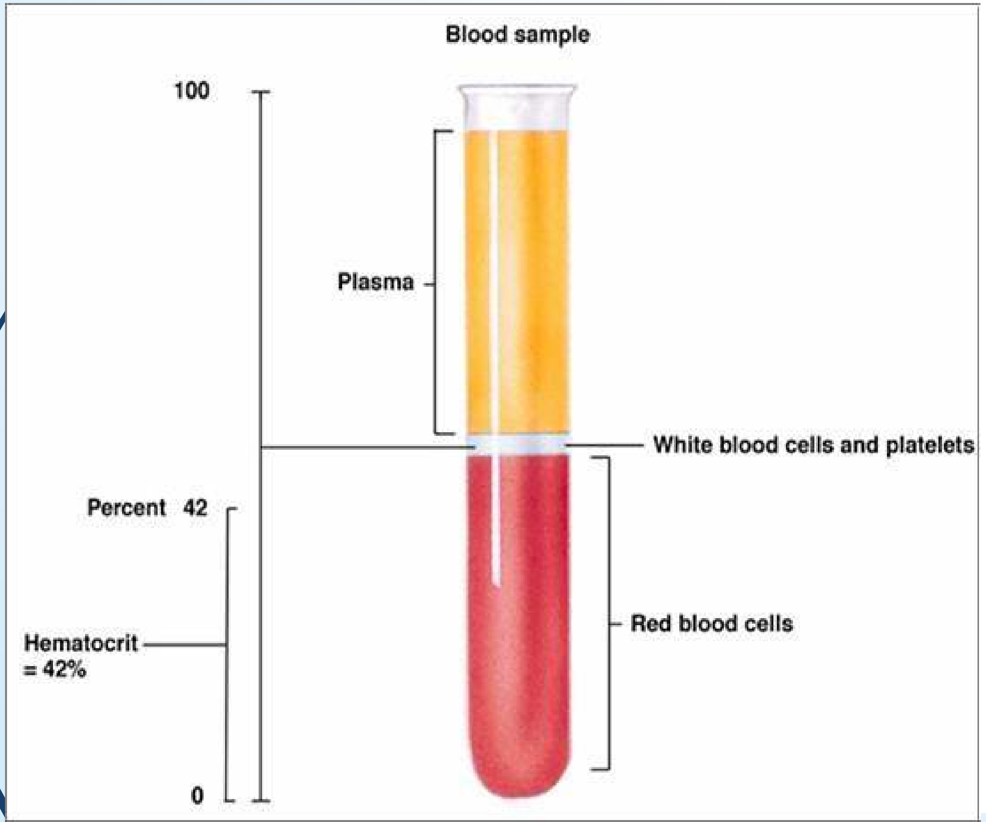


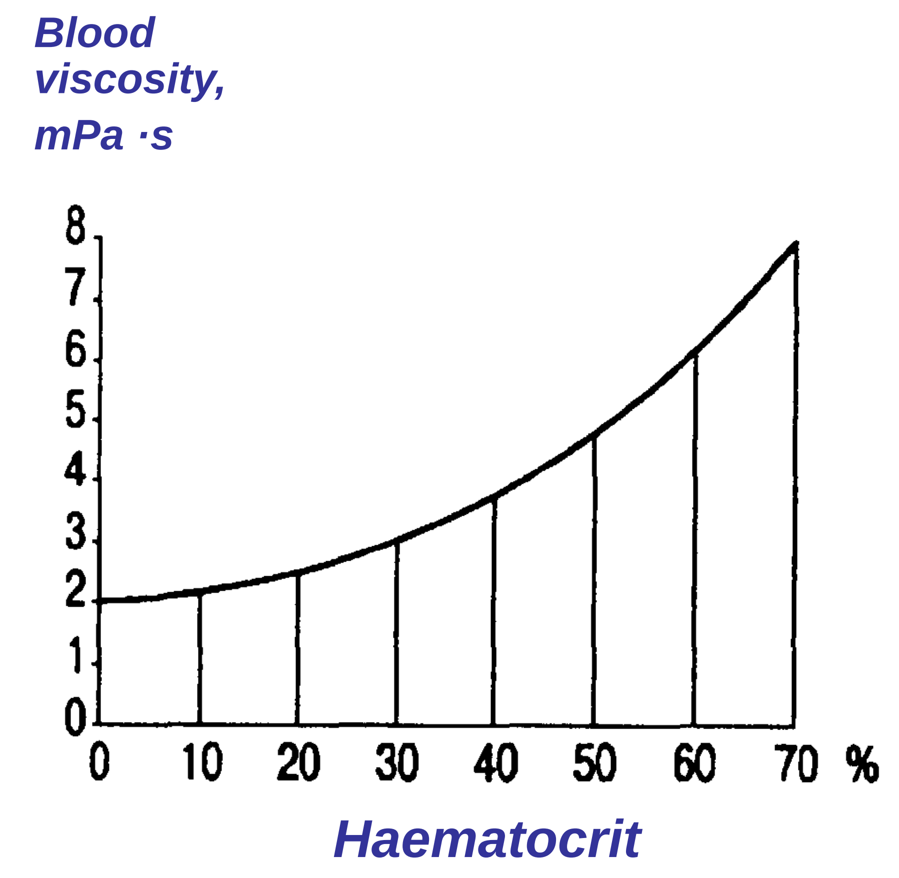




The quantitative indicator characterizing the content of blood cells in blood is the *haematocrit value (H)*, which is equal to the relative volume of blood cells in blood, in percentage.

Normally, this index in humans would be about 46%. The haematocrit value increasing, the blood viscosity increases, this relation being quite complicated.





What is Hematocrit?

1. The ratio of the volume of red blood cells to the total volume of blood.

2. Blood is composed mainly of red blood cells and white blood cells suspended in an almost clear fluid called serum.

3. In most labs, the hematocrit is measured by a machine that automatically determines a variety of blood tests referred to as the blood count CBC (complete blood count).

4. The spun hematocrit or "spun crit“ is centrifuged blood that divides the blood components by density.

5. The red cells collect at the bottom and form a red column and are separated from the straw-colored serum column by a small area composed of white blood cells.

6. The height of the red cell column divided by the height of the total fluid in the capillary tube equals the hematocrit (percentage of RBC's in the total blood volume).

Erythrocyte Sedimentation Rate

It is the rate of downward descent of RBCs in a vertical column of blood.

<u>Principle</u>:

If anti-coagulated blood is allowed to stand vertically in a narrow tube , the red cells will settle progressively to the bottom leaving clear plasma above.

-The cells settle due to :

1-Density of RBCs is greater than that of plasma.

2-RBCs tend to aggregate to form Rouleaux.

(Rouleaux differs from agglutinatin that agglutinated cells are irreversibly bound together and can not be separated)

Westergren method for estimation of ESR :

Equipments:

-Westergren tube (straight glass tube 30 cm in length , 2.5 mm in dimeter and graduated from 0 – 200 mm)

-Special stand.

-3.8% Sodium Citrate.

-5 ml disposable syringe.

Procedure :

1-Withdraw blood sample in a syringe and mix it with 3.8% Na Citrate at a ratio 4 : 1 (e.g. 2ml blood + 0.5 ml Na Citrate)

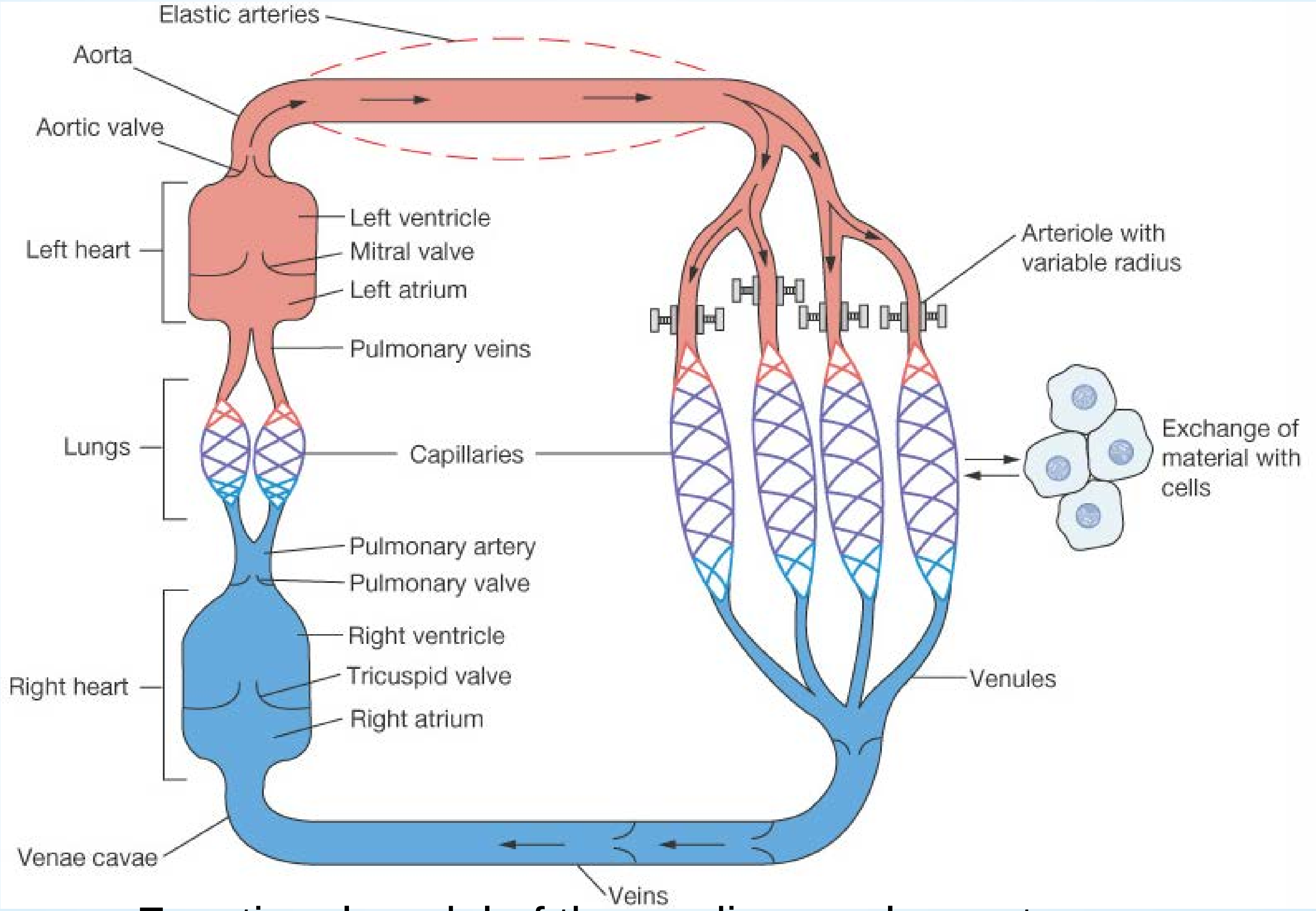
2-suck the citrated blood to Westergren tube up to zero mark exactly and place your finger over its openning.

3-Place the tube in a special stand that fix the upper lip with a clip.

4-The height of clear plasma on the top of the tube is measured after one and two hours.

HAEMODYNAMICS

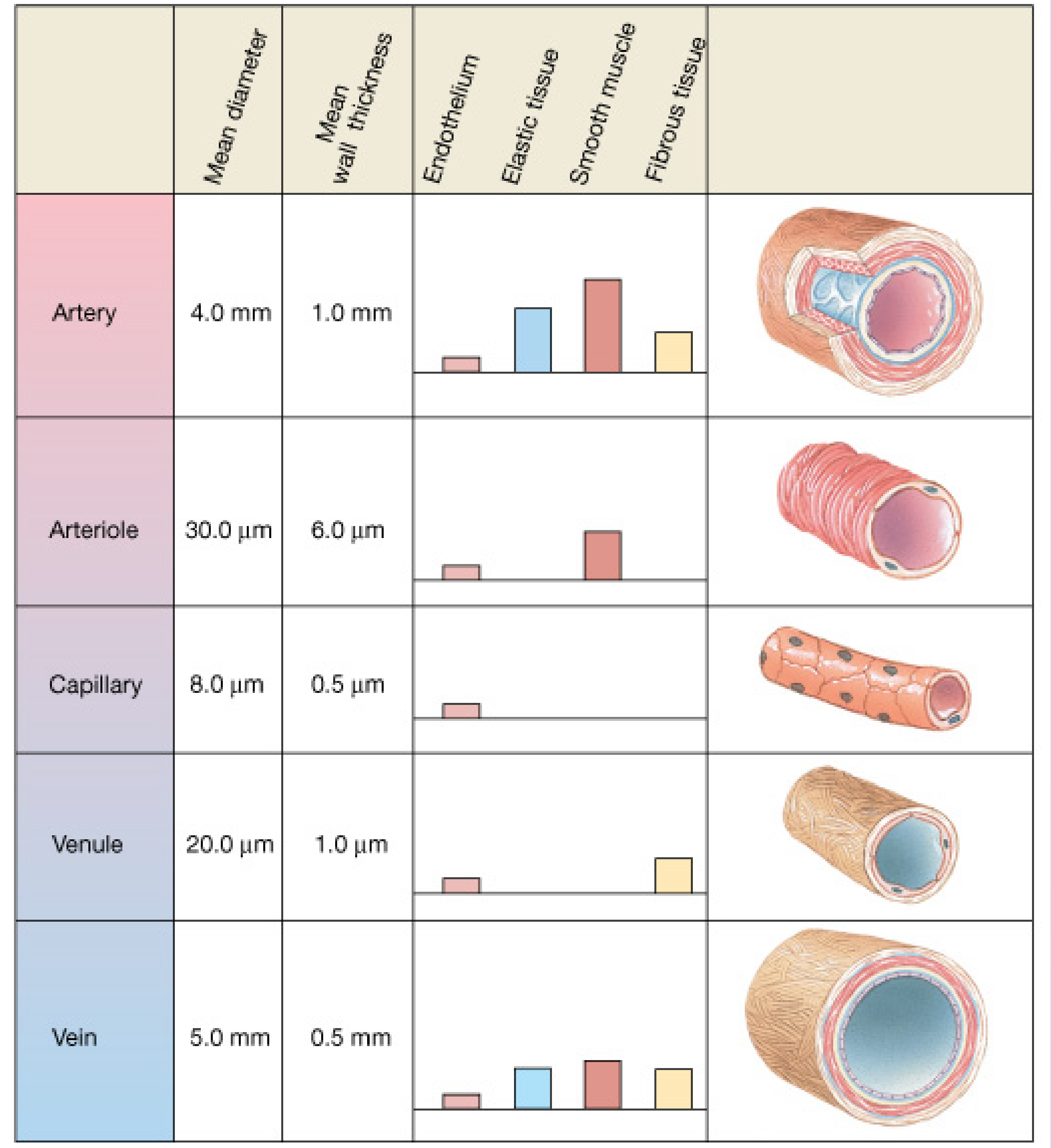
The Blood Vessels and the Cardiovascular System



Functional model of the cardiovascular system

Make Up of Blood Vessels: Arteries and Arterioles

- Endothelium
- Elastic tissues
    - Rebounds
    - Evens flow
- Smooth muscles
- Fibrous tissue
    - Tough
    - Resists stretch



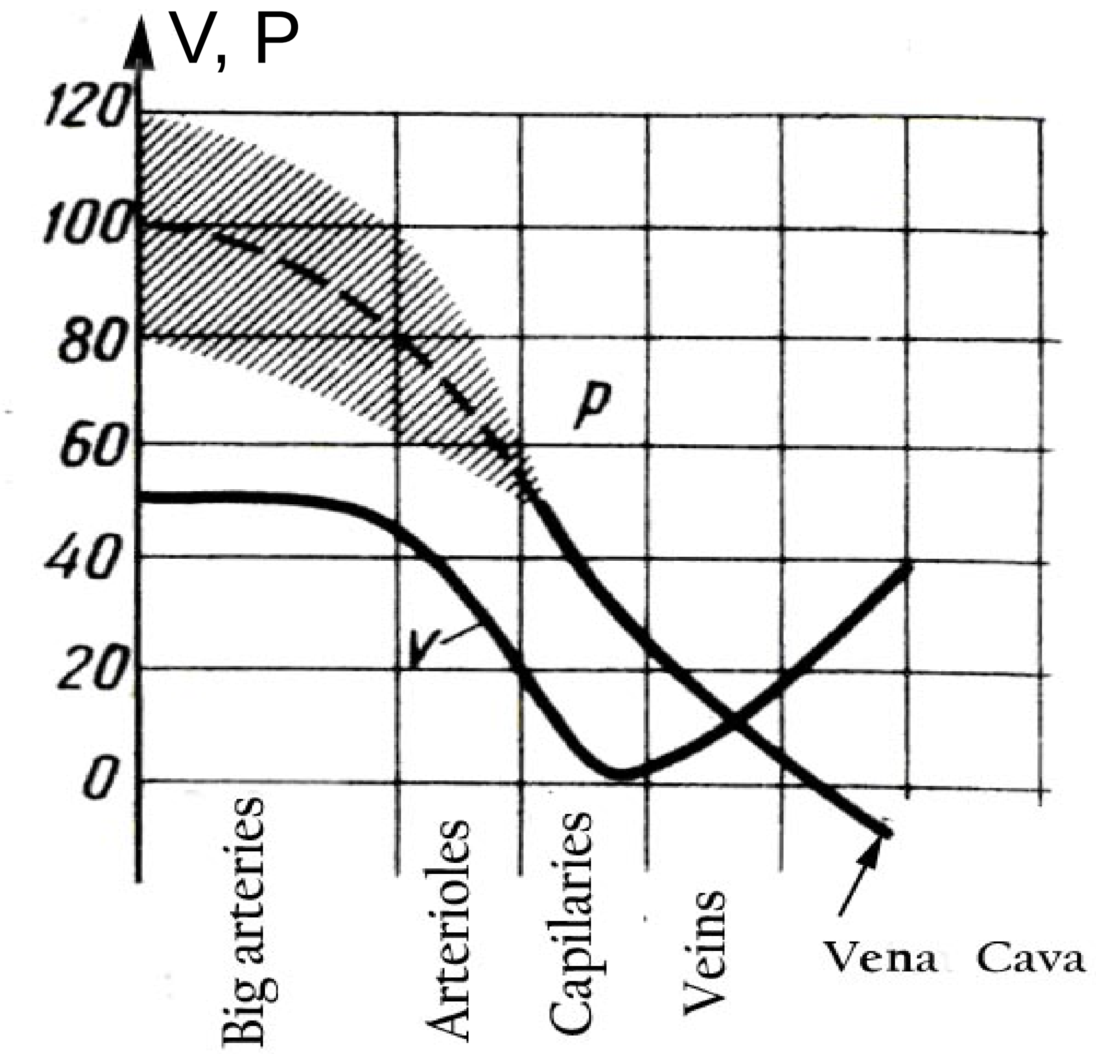
	Mean diameter	Mean wall thickness	Endothelium / Elastic tissue / Smooth muscle / Fibrous tissue	
Artery	4.0 mm	1.0 mm		
Arteriole	30.0 µm	6.0 µm		
Capillary	8.0 µm	0.5 µm		
Venule	20.0 µm	1.0 µm		
Vein	5.0 mm	0.5 mm		

Blood vessels

Basic Characteristics of Blood Dynamics in the Blood Circulation System

Vessel	*Cross-section area, cm²*	*Total number*	*Length m*	V_{mean} $m \cdot s^{-1}$	$\dot{\gamma}$ s^{-1}	Re_{mean}
Aorta	2.0–8.0	*1*	*80*	0.3-0.6	100	1200–5800
Big arteries	0.01–0.3	*~10³*	*40–20*	0.2	400	100–1000
Aarterioles	$3{\cdot}10^{-4}$–$8{\cdot}10^{-3}$	*~10⁸*	*5–0.2*	0.002 - 0.1	$>10^2$	0.01-10
Capillaries	$2{\cdot}10^{-7}$– $8{\cdot}10^{-7}$	*≥10⁹*	*0.1*	$(5–7){\cdot}10^{-4}$	400	0.001–0.003
Veinules,	$3{\cdot}10^{4}$ – 0.03	*~10⁹*	*0.2–1*	10^{-5} - 10^{-4}	$\sim10^2$	0.01–1
Big veins	0.2–0.8	*~10³*	*10–30*	0.1– 0.2	100	100–600
Vena cava	~3	*2*	*50*	0.1– 0.2	50	600–1000

The distribution curve of the mean flow velocity of blood and mean blood pressure in different parts of blood circulation system



Indirect bloodless way of taking blood pressure (according to Korotkov)

The physical idea of this method is based on an approximate equality of pressures in the cuff and in the soft tissues, and on the possibility of registering the audio signals accompanying the vibration of the vessel walls in turbulent flow of blood through the vessels.

The manometer readings at the first pulse beat show the ***maximum*****, or *systolic pressure*** (normally it is about 16 kPa or 120 mmHg).

The manometer readings at the moment of a sharp decrease of tones correspond to the ***minimum*****,** or ***diastolic pressure*** (*11 kPa, or 80* mmHg*)*

Work and Power of Heart

When contracting, the heart does work against the force of the blood pressure, thus pushing the blood into the artery, and gives it kinetic energy. Work which done by the ventricle to push the blood per one contraction can be calculated using the formula $A = P \cdot \Delta V$, where P is the mean pressure in the ventricle at contraction, ΔV is a variation in the volume of the ventricle.

The variation in the volume of the heart chamber per one contraction is referred to as *heartbeat volume* (or *stroke volume*), that is, $\Delta V = V_s$. The kinetic energy of the blood pushed onto the artery is

where m is the mass of the pushed blood, ρ is the blood density, v is the mean velocity of the blood in the artery.

Work and Power of Heart

Thus, work done by the left ventricle (A_{lv}):

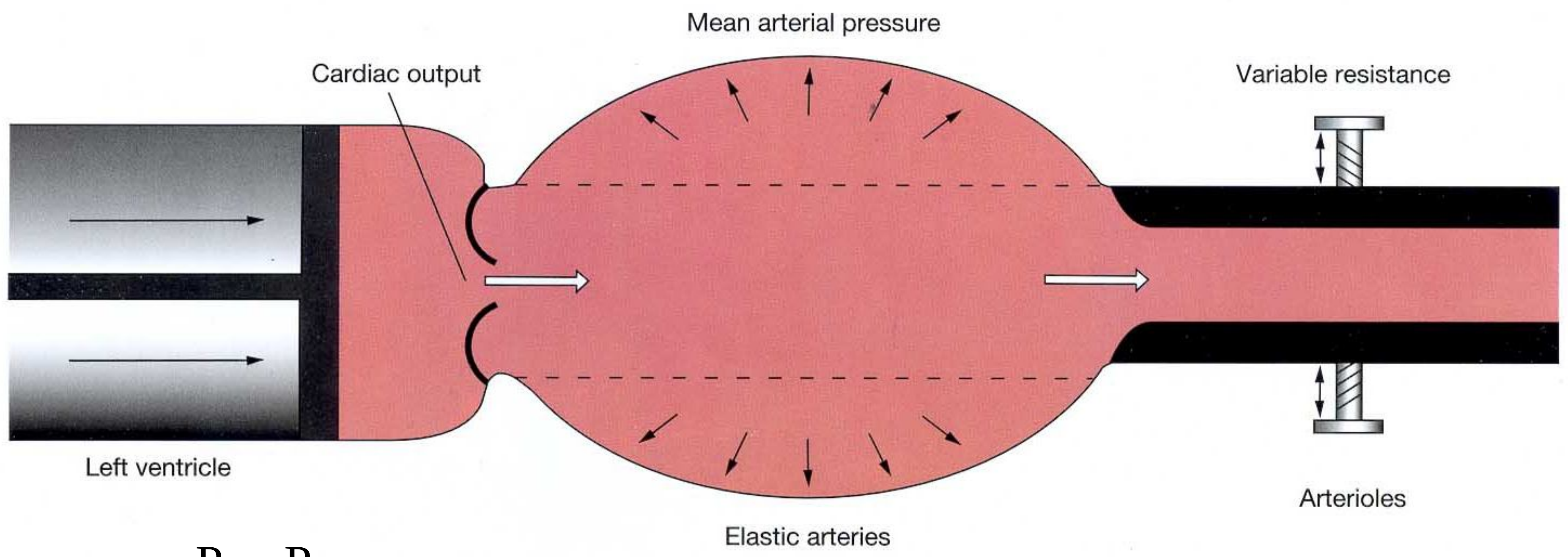
$$A_{lv} = V_s(P+\rho v^2/2)$$

Work done by atrium being neglected, and assuming that work done by the right ventricle approximately equals 20% of that of the left one, we get

$$A = 1{,}2 \cdot V_s(P+\rho v^2/2)$$

If we represent this in numerical expression, we will see that work done by the heart per one heart contraction approximately equals to 1 J. The time of one heart contraction being approximately 0.3 s, the power of heart approximately equals to 3.3 W.

Mean Arterial Pressure

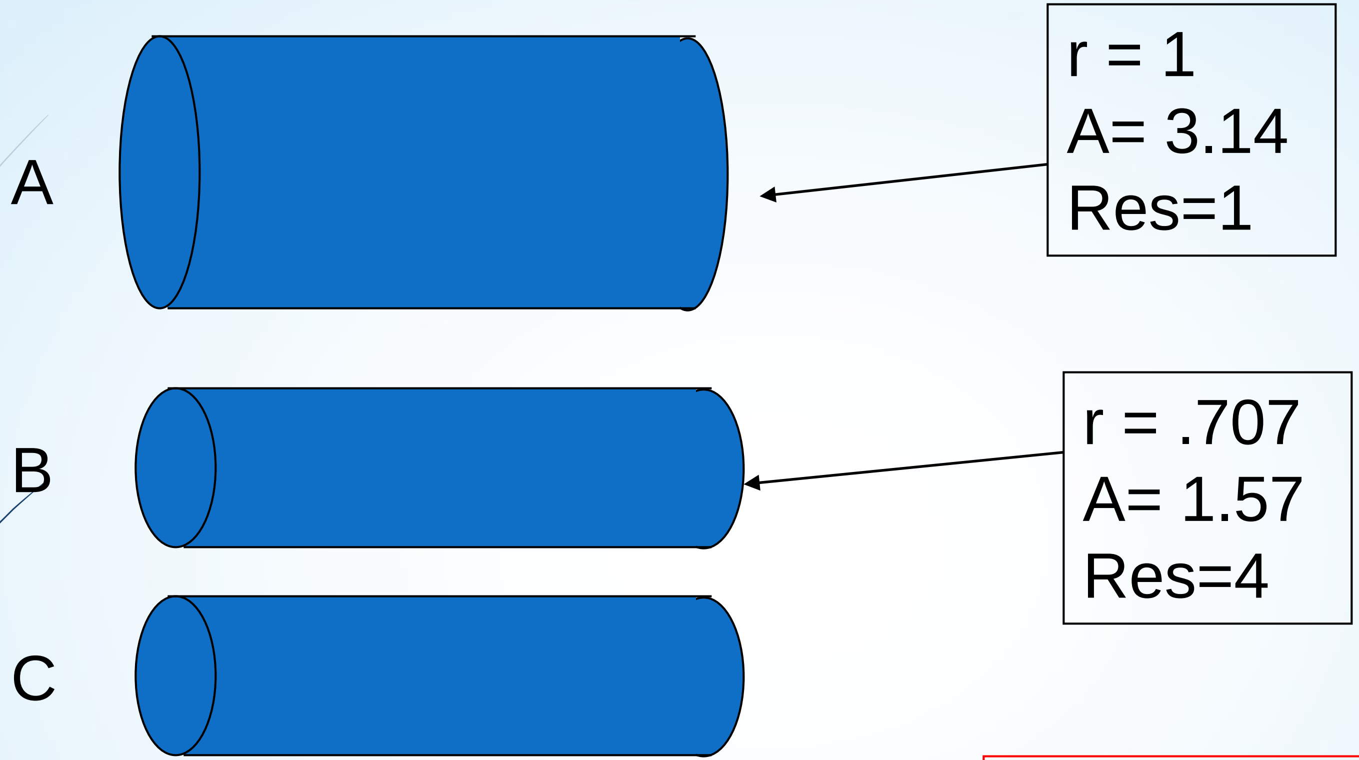


$$\dot{Q} = \frac{P_a - P_v}{R}$$

MAP = cardiac output x total resistance

$$\overline{P}_a - \overline{P}_v = \dot{Q} \bullet R$$

Branching to smaller radius vessels increases resistance



B and C in parallel have a combined resistance of 2 and a cross sectional area of 3.14

Resistance ~ (r^4)
Area ~ (r^2)

Pulse wave

- *Pulse wave* is a high-pressure wave that spreads through aorta and arteries. It is caused by blood being pushed from the left ventricle during the systole. Pulse wave speed is 5-10 $m \cdot s^{-1}$. The pulse wave corresponds to the blood stream pulsation speed in big arteries. Blood speed (0,5 $m \cdot s^{-1}$) is considerably lower than the speed of the pulse wave.

- Contracting during the systole, the heart pushes blood to aorta and branching arteries. If the walls of these vessels were rigid, the pressure developed at the heart output would be transmitted to the peripherals with the speed of the sound.

- The walls of the vessels being elastic, the blood pushed by the heart during the systole expands the aorta, arteries and arterioles. The blood inflow to the big vessels during the systole exceeds its outflow to the peripherals

Speed of the wave is related to the stiffness of the artery it is traveling in

The stiffer the artery; the higher the wave speed

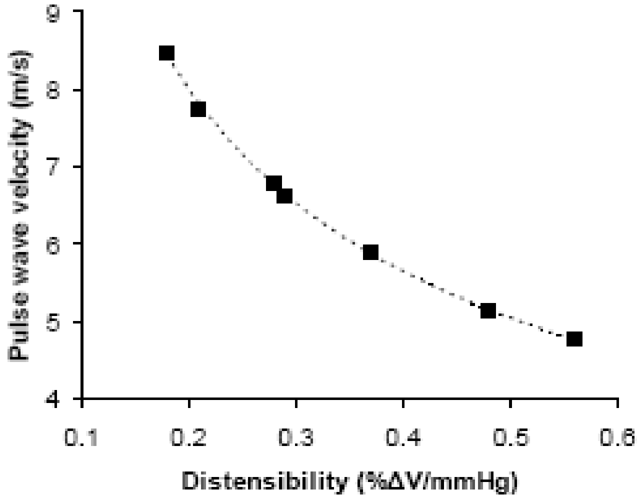
Wave speed is proportional to the square root of arterial stiffness

- Stress (σ, sigma)
  - Force per unit area = (F/A)
- Strain (ε, epsilon)
  - Change in length per unit length = ($\Delta L/L_0$)
- Elastic (Young's) modulus (E) $= \dfrac{F\,L_0}{A\,\Delta L}$
  - stress/strain = σ/ε

Pulse wave velocity

- PWV is inversely proportional to square root of arterial compliance

$$pwv \propto \sqrt{\frac{\Delta p \cdot v_0}{\Delta v}}$$



Bramwell & Hill, 1922

Hemingway *et al*, 1928

Control Questions

1. The rules of calculation of total hydraulic resistance.
2. Rheological Properties of Blood.
3. Hematocrit.
4. Erythrocyte Sedimentation Rate.
5. Mean Arterial Pressure.
6. Pulse wave velocity.

Recommended literature:

Basic:

1. Vladimir Timanyuk, Elena Zhivotova, Igor Storozhenko. Biophysics: Textbook for students of higher schools / Kh.: NUPh, Golden Pages, 2011.- 576p.
2. Vladimir Timaniuk, Marina Kaydash, Ella Romodanova. Physical methods of analysis / Manual for students of higher schools/– Kharkiv: NUPh: Golden Pages, 2012. – 192 p.
3. Philip Nelson. Biological Physics. – W. H. Freeman, 1st Edition, 2007. – 600 p.
4. Biophysics, physical methods of analysis. Workbook: Study guide for the students of higher pharmaceutical educational institutions / Pogorelov S. V., Krasovskyi I. V., Kaydash M. V., Sheykina N. V., Frolova N. O., Timaniuk V. O., Romodanova E.O., Kokodii M.H. – Kharkiv., – 2018. – 130 p.
5. Center for distance learning technologies of NPhaU. Access mode: http://nuph.edu.ua/centr-distancijjnih-tehnologijj-navcha/

Support:

1. Eduard Lychkovsky. Physical methods of analysis and metrology: tutorial / Eduard Lychkovsky, Zoryana Fedorovych. – Lviv, 2012. – 107 p.
2. Daniel Goldfarb. Biophysics DeMYSTiFied. – McGraw-Hill Professional, 1st Edition, 2010. – 400 p.


Thanks for
your attention