PRESSURE

Def'n: Newton (N): the force that will accelerate a mass of 1 kg at 1.0 $m.s^{-2}$
Gravity (g): 9.81 $m.s^{-2}$
Pascal (Pa): 1 Pa = 1 N acting over 1 m^2
Pa = $N.m^{-2}$

- therefore, the force of gravity on 1 kg will be 9.81 N
- so, 1 newton is equivalent to 1/9.81 kg ≈ 102 gram weight
- 102 g acting over a square metre is small and cumbersome → kPa
- atmospheric pressure at sea level = **101.325 kPa**
  = **760 mmHg**

Def'n: **1 kPa** ≈ 10.2 cmH_2O
≈ 7.5 mmHg → mercury ≈ **13.6x** water density

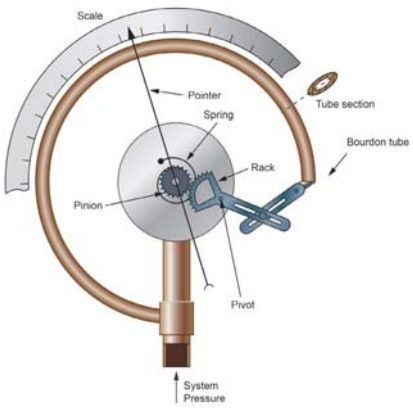
1 bar = 100 kPa
≈ 750 mmHg

■ *Classification*

1. Absolute Pressure: - relative to a perfect vacuum
2. Gauge Pressure: - relative to atmospheric pressure.
3. Differential Pressure: - difference between two pressures
   - useful for flow measurement and level detection

■ *Common Devices*

1. Manometers - cmH_2O, mmHg
2. Bourdon gauges - calibrated → pressure/temperature
3. Digital pressure sensors
   i. strain-guage
   ii. piezo-electric/resistive
   iii. electromagnetic
   iv. capacitance
   v. optical



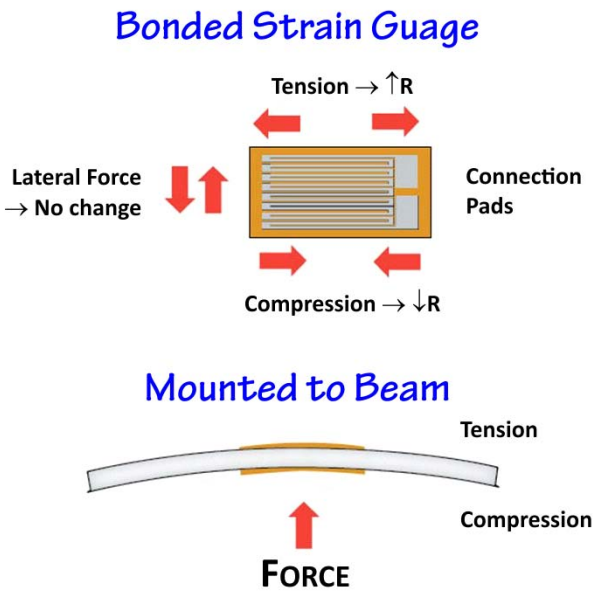
Electronic Pressure Measurement

Def'n: ***Transducer***: a device that converts one type of ***energy*** to another

- electrical ↔ electromechanical ↔ electromagnetic ↔ photonic ↔ photovoltaic, etc.
- commonly implies use as a ***sensor/detector***

■ *Bonded Strain-guage*

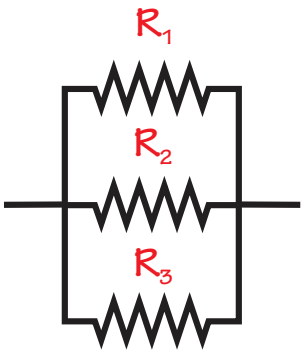
- common form of transducer for pressure measurement → $\delta R_{Electrical} \propto$ Force



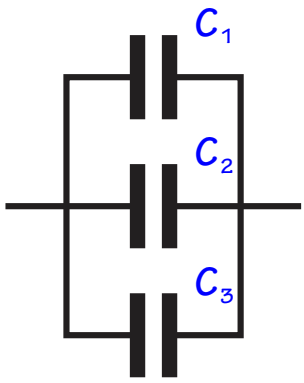
- configurations are employed to measure ***weight***, ***pressure***, ***torque***, etc.
- deformation of calibrated beams, diaphragms, etc to which *mechanical force* is applied
- transducers can be rugged, compact, linear, highly accurate, and temperature compensated
- through proper flexure design and gage placement, a ***linear relationship*** can be achieved between the applied force and the sensed strain
- usually incorporated into a Wheatstone Bridge circuit

■ *Electrical Resistance & Capacitance*

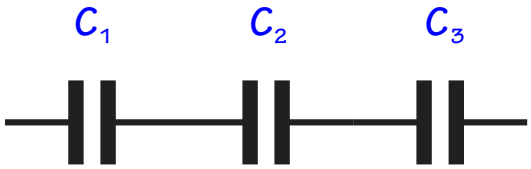
$$R_T = R_1 + R_2 + R_3$$



$$R_T = (1/R_1 + 1/R_2 + 1/R_3)^{-1}$$



$$C_T = C_1 + C_2 + C_3$$



$$C_T = (1/C_1 + 1/C_2 + 1/C_3)^{-1}$$

- in electircal theory, resistance is constant ("Ohmic") and Ohm's Law applies
- in physiology:

a. resistance is almost universally ***flow dependent***, i.e. not constant and Ohm's Law does not apply, except as a conceptual guide

b. flow exhibits both inductive reactance (X_L) and capacitance (X_C),
→ ***impeadance (Z)*** is a more correct analogy:

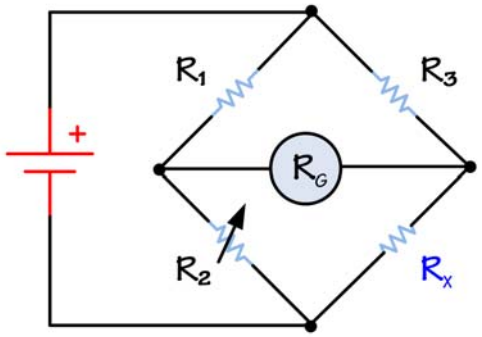
$$Z = \sqrt{R^2 + (X_L + X_C)^2}$$

c. compliance is modelled c.f. electrical capacitance, and like Ohm's Law and resistance, this is an approximation only

■ *Wheatstone Bridge*

- not invented by Charles Wheatstone, actually Hunter Christie (1802-1875)
- Wheatstone popularized the arrangement: battery + 4 resistors (R) + galvanometer (G)

→ termed the circuit a *"Differential **Resistance** Measurer"*

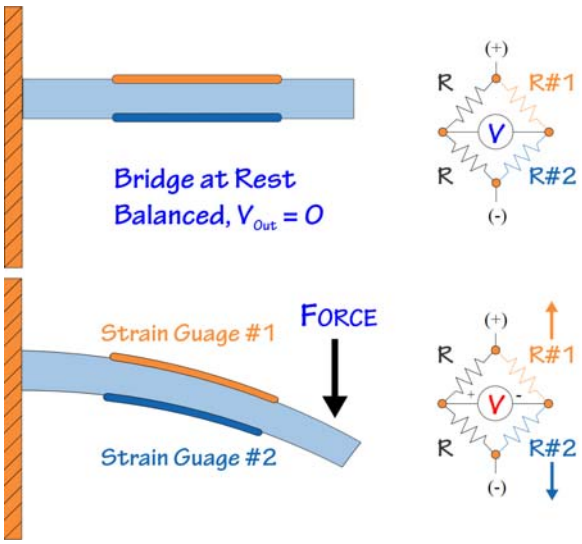


- when balanced: $\mathbf{R_X} = R_2 \,.\, (R_3 / R_1)$
- internal R_G becomes irrelevant as no current flowing through G = "*null balance*"
- the "low" ***internal*** $\mathbf{R_G}$ was a major source of error in early parallel circuits:

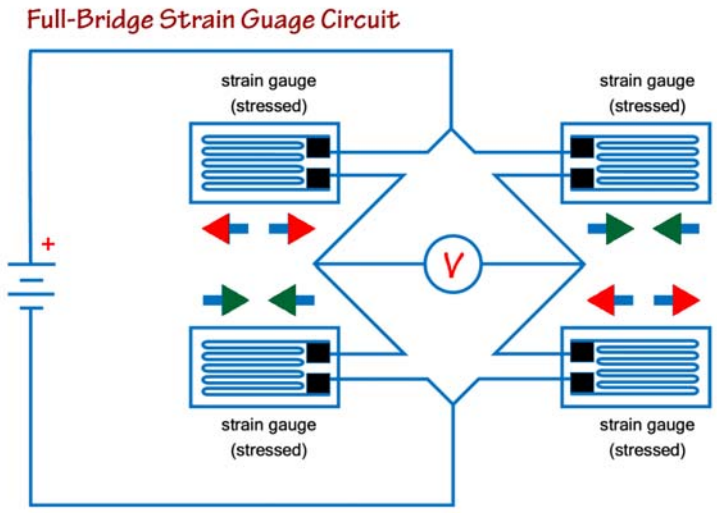
$$\frac{1}{R_T} = \frac{1}{R_1} + \frac{1}{R_2} + \ldots + \frac{1}{R_N}$$

- $R_{1/3}$ and variable R_2 were "precision" laboratory devices → unknown R_X

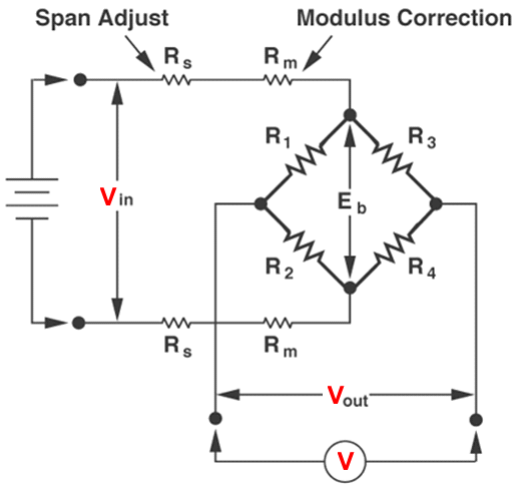
NB: This is ***not*** the purpose / use in modern devices, ***viz:***



- this can be extended to all 4 resistors in the WB, e.g.



- usually arranged in a circuit similar to:



$$E_{Out(mV)} = E_{In(V)} \cdot k \cdot F/100$$

  - where, k = Calibration Factor (mV/V, full scale)
    F = Input variable (% of full scale)

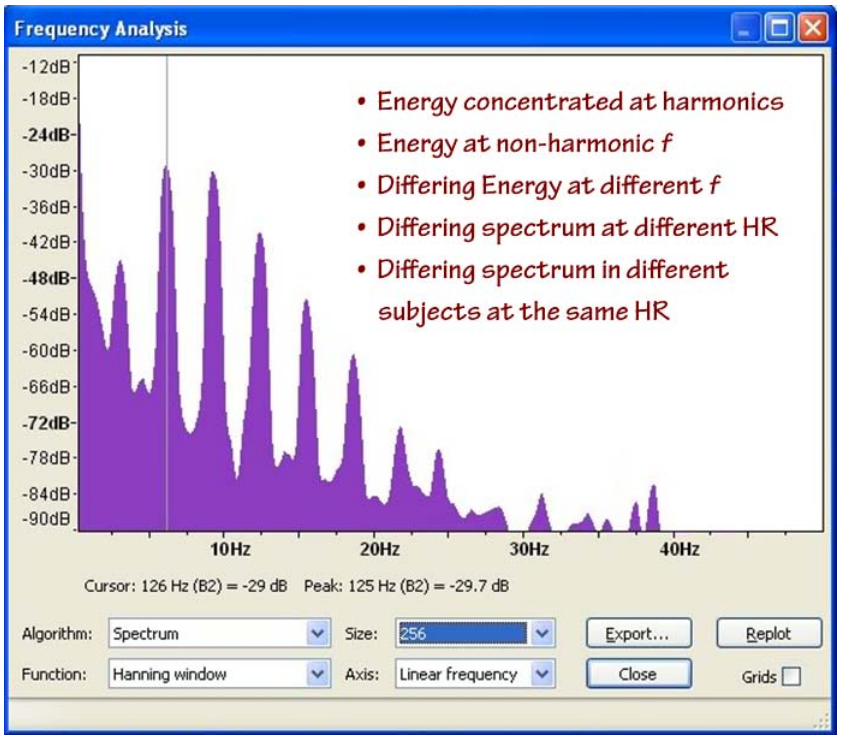
- ***NOT*** being used as a "null deflection" / balanced circuit as Wheatstone described!
  a. inbuilt ***temperature*** compensation
     i. electrical resistance - change in R with δT
     ii. modulus of elasticity - change in bridge elastance
  b. correction for ***lead resistance*** (length/age)
  c. improved ***signal level*** (S:N Ratio)
  d. improved linearity

■ *Calibration*

1. static calibration
   - i. zero *usually only static cal done for IABP
   - ii. gain
     - calibration against reference standard, e.g. mmHg column
     - not routinely performed clinically ∝ manufacturer "batch control"
   - iii. linearity & hysteresis
   - iv. stability with time, temperature variation
2. dynamic calibration
   - i. frequency composition of ***signal*** *-vs-* ***system*** resonant frequency
   - ii. ideally: ***system*** f_0 **>> 10x** fundamental signal frequency
   - iii. damping coefficient, $\zeta \approx 0.68$

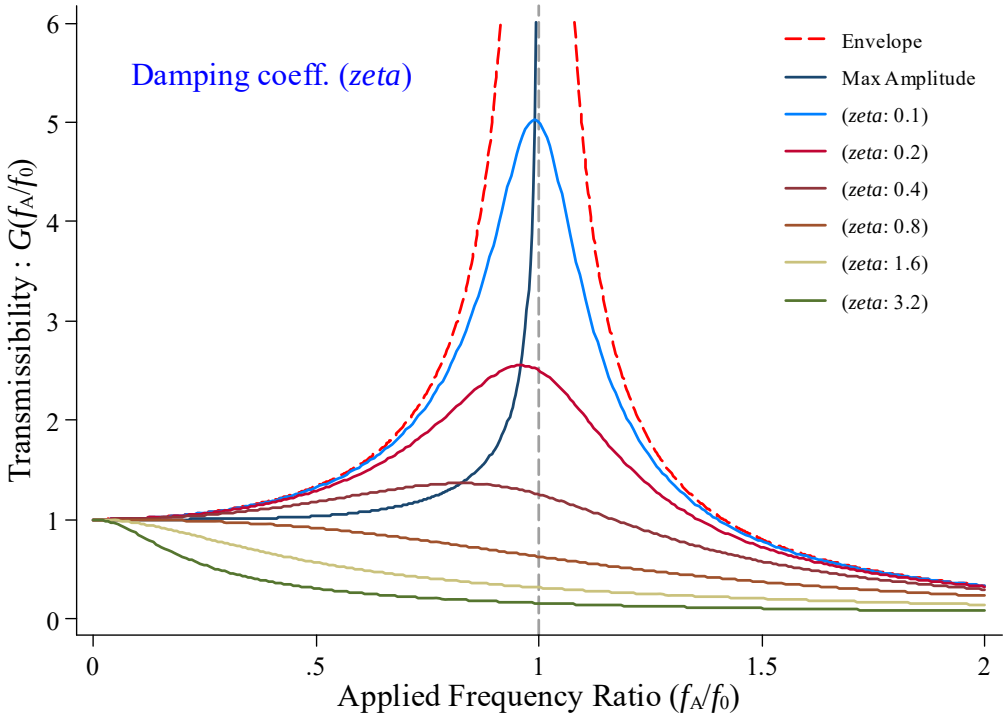
■ *Dynamic Calibration*

1. frequency composition of ***signal energy***
   - ***Fourier analysis*** → any biological waveform ≅ summation of sine waves
   - empirically, for biological waveforms
     → "virually all energy" between ***base/fundamental*** and ***10th harmonic***



- IABP → signal E ≈ 0.5 - 30Hz
- ECG → signal E ≈ 0.5 - 40Hz

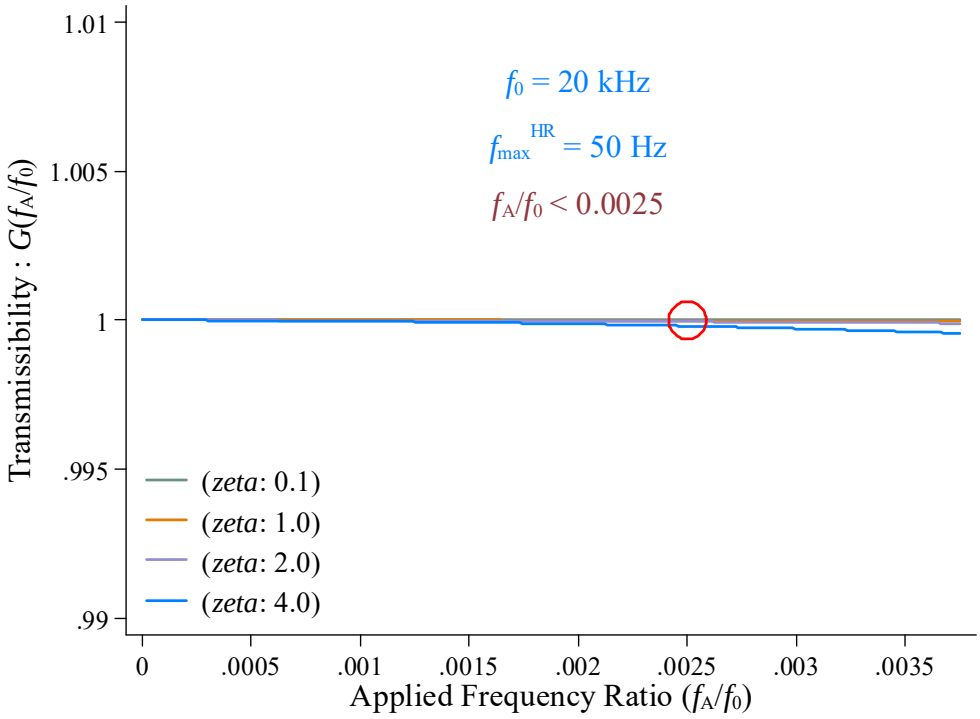
2. measurement system ***resonant frequency & damping***
   - ***resonant frequency*** (***f_0***) is the natural frequency at which an object or system vibrates most easily and with the greatest energy
   - ***damping*** (zeta, ζ) in an oscillating system, is the loss of ***energy*** by dissipation → viscous damping/drag, surface friction, heat/radiation
     - in the absence of energy input, $\zeta > 0$, system oscillation → zero
     - in the absence of damping, *$\zeta = 0$, with ongoing energy input, system oscillation will → infinity **theoretical only*
3. system ***energy transmissibility*** (**G**), therefore depends upon:
   i. applied energy frequency (f_A), in relation to system f_0 = **(f_A/f_0)**
   ii. system damping, *viz*.



NB: for any given system,

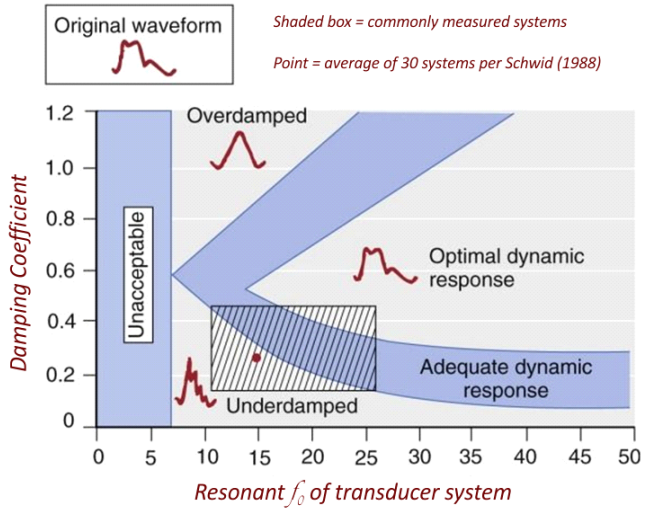
- as (f_A/f_0) → ≈ 1.0 ζ becomes increasingly important
  ζ has some ***optimal value*** where G ≈ 1.0
- as (f_A/f_0) → ≈ 0 ζ becomes irrelevant, and G → 1.0

- e.g. if we used a piezo-electric transducer (f_0 ≈ 20 kHz) → f_A/f_0 ≈ (50/20,000) = 0.0025
  → damping becomes essentially irrelevant



NB: common piezo-electric crystals: f_0 ≈ **20-30 MHz** i.e. $f_A/f_0 < 0.00001$!

- this would eliminate ***any*** concern re resonace/damping for IABP monitoring, or any other biological waveform analysis
- however, we use fluid-catheter systems for:
  a. repeated blood sampling
  b. continuous flush reduces protein deposition & extends functional life
  c. ability to re-zero over time, c.f. Codman ICP, where ***drift*** becomes problematic



NB: the above graph is reproduced in multiple textbooks & papers

- interaction between damping coefficient and natural frequency
- the point within the shaded box shows the mean values of 30 systems

*as recorded by **Schwid (1988).**

NB: this is historical and ***largely irrelevant*** data !
modern transducer systems are markedly better, typically with f_0 **> 80-100 Hz**

■ *Facotrs Affecting the Fidelity of IABP Monitoring*

1. Equipment factors ↑ f_0 **System**
   - i. high frequency transducer response
   - ii. short, stiff, non-compliant tubing
   - iii. elimination of air bubbles
   - iv. slow continuous flush device
   - v. correct static calibration of the transducer
2. Patient factors which ***reduce*** fidelity ↑ f_Λ **Signal**
   - ι. ↑ HR
   - ii. reflection of high frequencies within circulation
     - elderly, atherosclerosis, high SVR
   - iii. site of catheter insertion - radial | femoral | aortic
   - iv. profound vasodilatation - post-CPB

FLUID FLOW

Laminar Flow

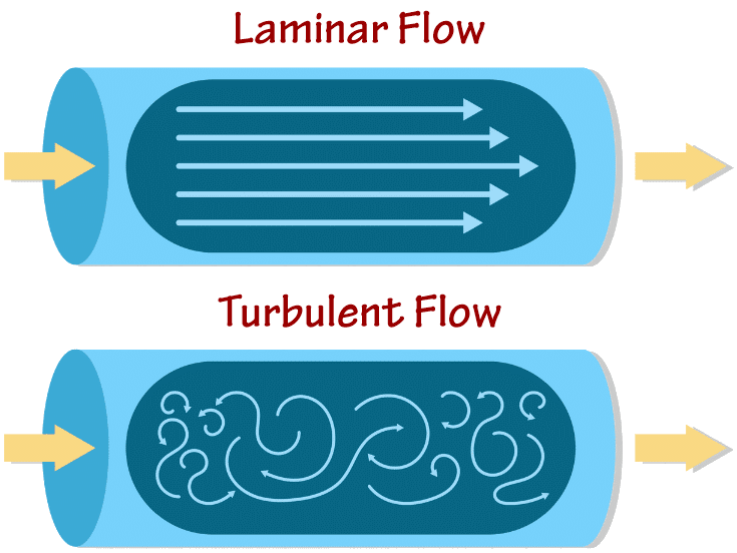
$$\dot{Q} = \frac{\pi r^4 . \delta P}{8 \eta l}$$

Hagen-Poiseuille Equation

but as R = δP/Q, so

$$R = \frac{8 \eta l}{\pi r^4}$$

- where,
  1. ***eta* (η)** = the ***viscosity*** of the fluid in pascal seconds
  2. there are no eddies or turbulence
  3. flow
     - i. is greatest at the centre, being ≈ 2x mean
     - ii. near the wall → 0
     - iii. is directly proportional to the driving pressure



Turbulent Flow

- the velocity profile across the lumen is lost
- flow becomes directly proportional to the ***square root*** of the driving pressure

  NB: therefore, as pressure flow is not linear, ***resistance*** is not constant, and the flow at which the resistance is measured must be specified

- other factors in turbulent flow may be summarized,

$$\dot{Q} = \frac{k.r^2.\sqrt{\delta P}}{\rho l}$$

where, k = a constant
ρ = ***rho***, the density of the fluid in $kg.m^{-3}$

- thus, radius has less of an effect on turbulent flow
- the likelihood of the onset of turbulent flow is predicted by,

$$\text{Reynold's number (Re)} = \frac{\rho v d}{\eta}$$

- where, d = the diameter of the tube
  v = the velocity of flow
  ρ = ***rho***, the density of the fluid in $kg.m^{-3}$
  η = ***eta***, the viscosity of the fluid in pascal seconds

- empirical studies for cylindrical tubes, if Re > 2000 turbulent flow becomes more likely
- for a given set of conditions there is a ***critical velocity*** at which Re = 2000

■ *Clinical Aspects*

- the transition from laminar to turbulent flow depends on the mixture of gases present
- in the patient's airway gases are humidified, contain CO_2 and are warmed
  → ↑ critical velocity
  ∝ ↓ density due to warming of the gases
- for a typical anaesthetic mixture, critical flow (L/min) ≈ airway diameter (mm)
- as breathing is cyclical, with peak flows > 50 L/min:
  - turbulent flow usually predominates during peak flow
  - laminar flow is present during other times in the respiratory cycle
- flow through bronchi and smaller airways → laminar ∝ marked reduction in velocity
- in general, during quiet breathing flow tends to be laminar, while during speaking, coughing, or deep breathing flow becomes turbulent in the larger airways

Viscosity

- for a given set of conditions, flow is inversely proportional to viscosity
- blood viscosity increases with,
    a. low temperatures
    b. increasing age
    c. increasing haematocrit
    d. abnormal elevations of plasma proteins
    e. cigarette smoking

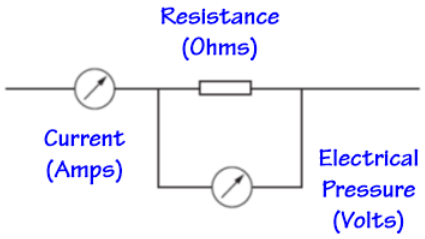
- this may be reduced with low MW dextran
- the viscosity of blood is anomalous due to the presence of cells

    → behavior is ***non-newtonian***

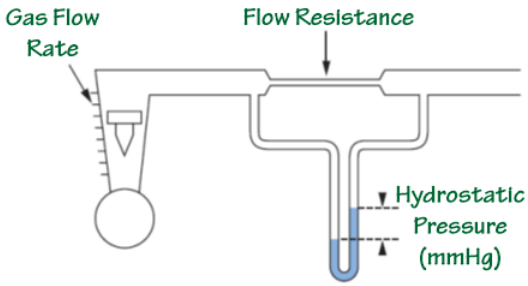
Flow Measurement

■ *Fleisch Pneumotachograph*

- based upon Ohm's Law:



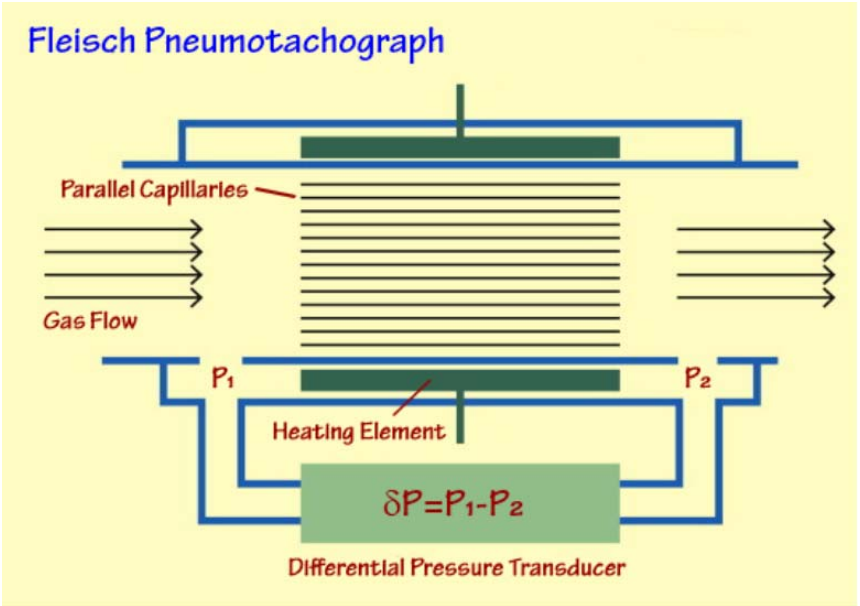
$R = V/I$



$R = \delta P/Q$

$Q = \delta P/R$

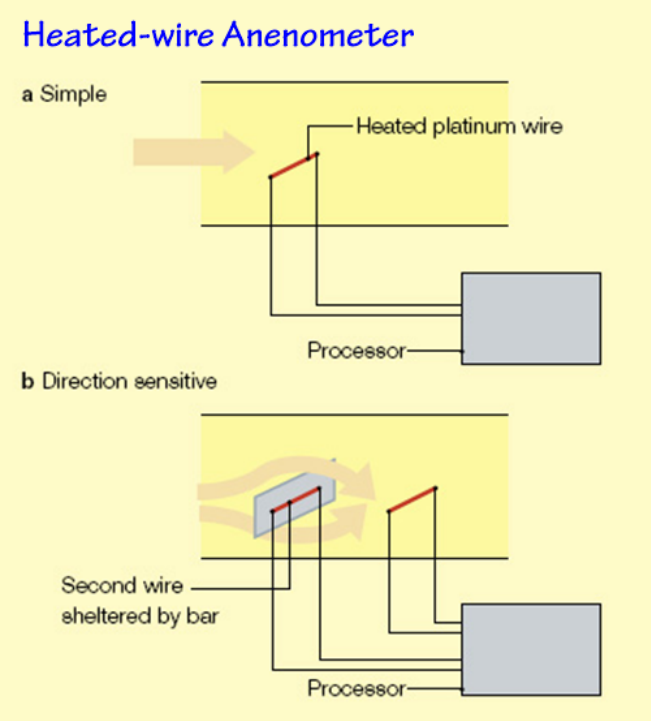
- multiple parallel capillaries — laminar flow; low total R_{Flow}
- differential pressure transducer — $R_{Trans} \rightarrow \infty$
- heated element to prevent condensation



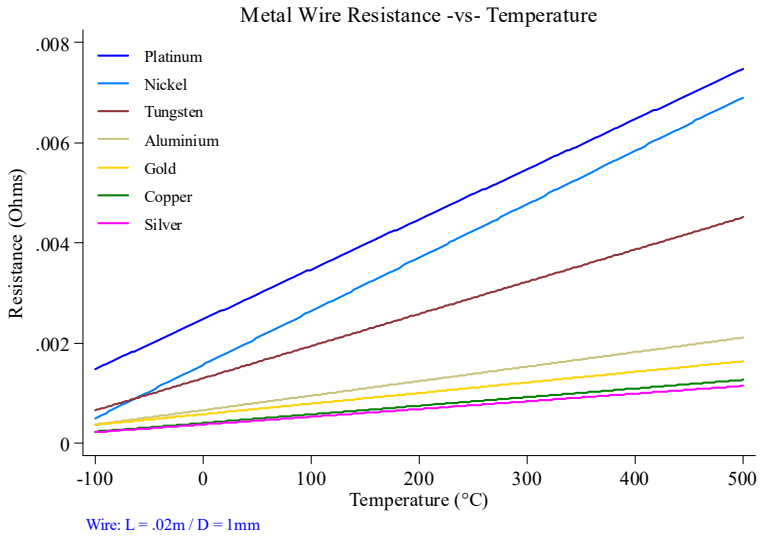
■ *Heated-wire Anenometer*

- current passed through ***platinum*** wire → heated to known T

$\uparrow T \rightarrow \uparrow R_{wire}$ $\uparrow Flow \rightarrow \downarrow T \rightarrow \downarrow R_{wire}$



- ***platinum*** chosen
  - higher resistivity / unit length
  - higher δR/δT (SNR)
  - linearity & range
  - stability over time (noble metal)



Tension

▪ *Laplace's Law*

$$P = T.h.(1/r_1 + 1/r_2)$$

thus, for straight tubes,

$$P = T.h./r$$

and, for spheres,

$$P = \frac{2T.h}{r}$$

where, **T** = the tangential force in N/m, acting along a length of wall
h = the thickness of the wall (usually small)

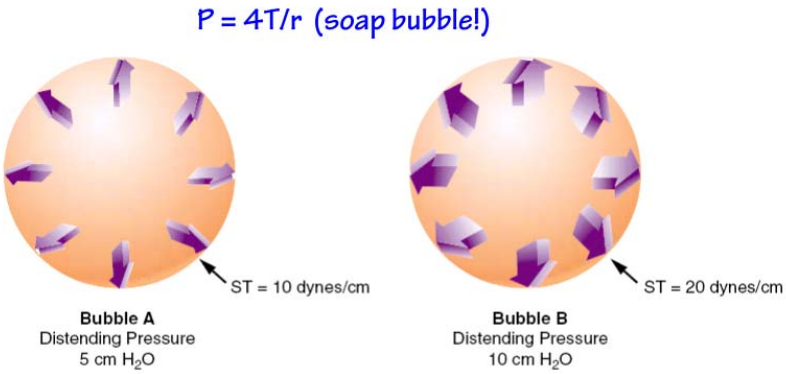


Figure 2–13. *Bubbles A and B are the same size. The surface tension (ST) of bubble A is 10 dynes/cm and requires a distending pressure (P) of 5 cm H_2O to maintain its size. The surface tension of bubble B is 20 dynes/cm H_2O (twice that of bubble A) and requires a distending pressure of 10 cm H_2O (twice that of bubble A) to maintain its size (r = radius).*

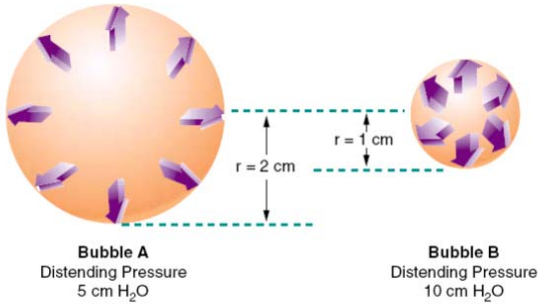


Figure 2–14. *The surface tension (ST) of bubbles A and B is identical. The radius (r) of bubble A is 2 cm, and it requires a distending pressure (P) of 5 cm H_2O to maintain its size. The radius of bubble B is 1 cm (one-half that of bubble A), and it requires a distending pressure of 10 cm H_2O (twice that of bubble A) to maintain its size.*

- thus, as the diameter of a vessel becomes smaller, the collapsing force becomes greater
- this can lead to vessel closure at low pressures, the ***critical closing pressure***
- also seen in alveoli, leading to instability with small alveoli tending to fill larger ones
- however, due to the action of ***surfactant*** alveolar stability is maintained

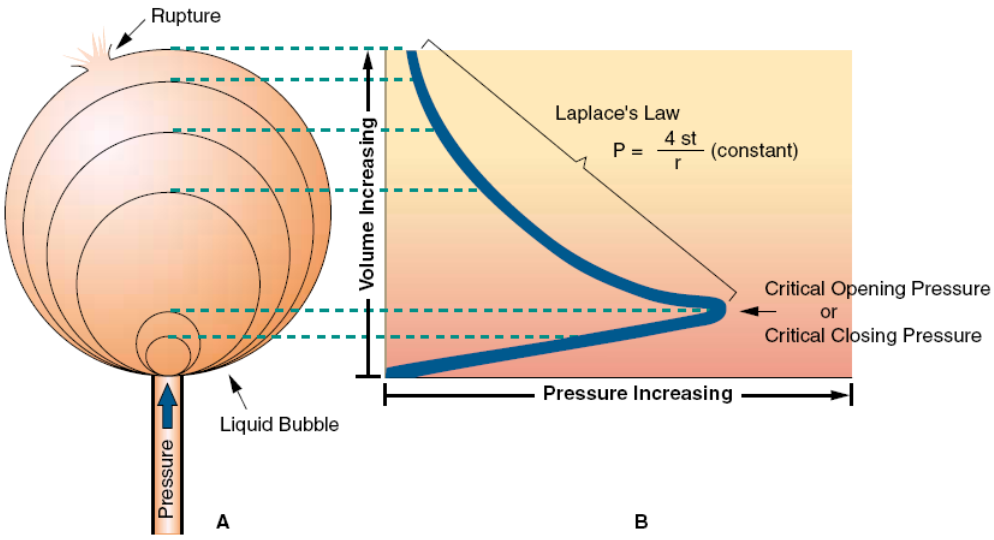


Figure 2–16. *(A) Model showing the formation of a new liquid bubble at the end of a tube. (B) Graph showing the distending pressure required to maintain the bubble's size (volume) at various stages. Initially, a very high pressure, providing little volume change, is required to inflate the bubble. Once the critical opening pressure (same as critical closing pressure) is reached, however, the distending pressure progressively decreases as the size of the bubble increases. Thus, between the critical opening pressure and the point at which the bubble ruptures, the bubble behaves according to Laplace's law. Laplace's law applies to the normal functional size range of the bubble.*

The Bernoulli Principal

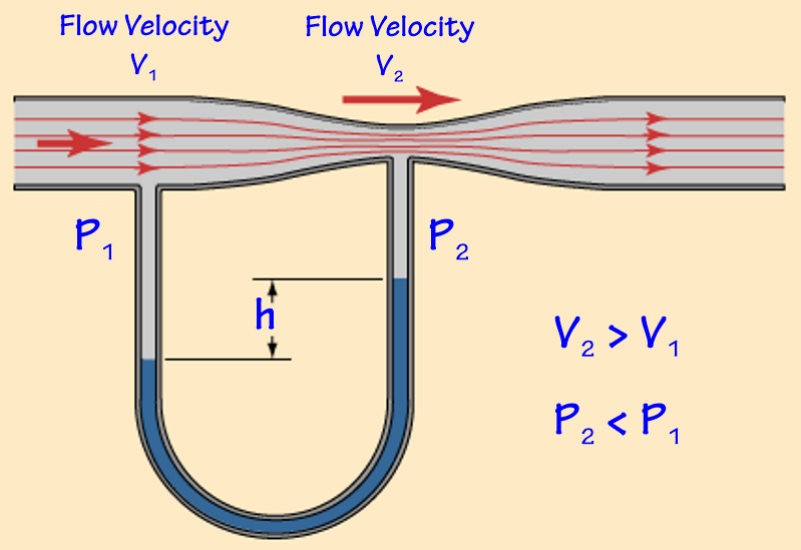
- based on the principal of ***conservation of energy***, the total energy of a fluid flow is given by,

$$E = PV + mgh + \tfrac{1}{2}mv^2$$

where,

PV	=	the potential energy of ***pressure***
mgh	=	the potential energy due to ***gravity***
$\tfrac{1}{2}mv^2$	=	the kinetic energy of ***motion***

- thus, as the velocity of flow increases passing though a narrowing and the velocity increases, so the pressure decreases



- also, for a system to work efficiently, laminar flow is important as turbulence would allow flow energy to be lost as heat
- this is the principal of operation of a venturi, where the opening of a side tube leads to the entrainment of another fluid
- the ***entrainment ratio*** ER is defined as,

$$ER = \text{Entrained Flow} / \text{Driving Flow}$$

- when there is no opening on the side of a narrowing in a tube, a region of low pressure is established and the stream tends to adhere to the wall
- if the tube then diverges, the stream may adhere to either wall, diverting flow to one or other lumen, the ***Coanda effect***
- valves can be constructed on this mechanism using fluid logic, a control nozzle being located just distal to the divergence of the lumen
- unfortunately these are wasteful and noisy

THE GAS LAWS

■ *Boyle's Law*

- at a constant temperature, the volume of a given mass of gas varies inversely with its absolute pressure, or,

$$PV = k_1$$

■ *Charles's Law*

- at a constant pressure, the volume of a given mass of gas varies proportionately to its absolute temperature, or,

$$V/T = k_2$$

■ *The Third Perfect Gas Law*

- at a constant volume, the absolute pressure of a given mass of gas varies proportionately to its absolute temperature, or,

$$P/T = k_3$$

■ *Dalton's Law of Partial Pressures*

- in a mixture of gases, the pressure exerted by each gas is equal to the pressure which would be exerted if that gas alone were present

■ *Avogadro's Hypothesis*

- equal volumes of gases, at the same temperature and pressure contain equal numbers of molecules

■ *Henry's Law*

- at a constant temperature, the amount of a given gas dissolved in a given liquid is directly proportional to the partial pressure of that gas in equilibrium with that liquid

STP: T = 273.15 K (0°C)
P = 101.325 kPa (760 mmHg)

■ *A Mole (mol)*

- is the quantity of any substance containing the same number of particles as there are atoms in 0.012 kg of ^{12}Carbon,

  $1 \text{ mol} \approx 6.022 \times 10^{23}$ ***Avogadro's number***
- for any gas at STP, $1 \text{ mol} \approx 22.4 \text{ litre}$

■ *Universal Gas Constant (R)*

- for 1 mol of any perfect gas, $R = PV/T$
- where n = number of mol of gas, $PV = nRT$

■ *Critical Temperature*

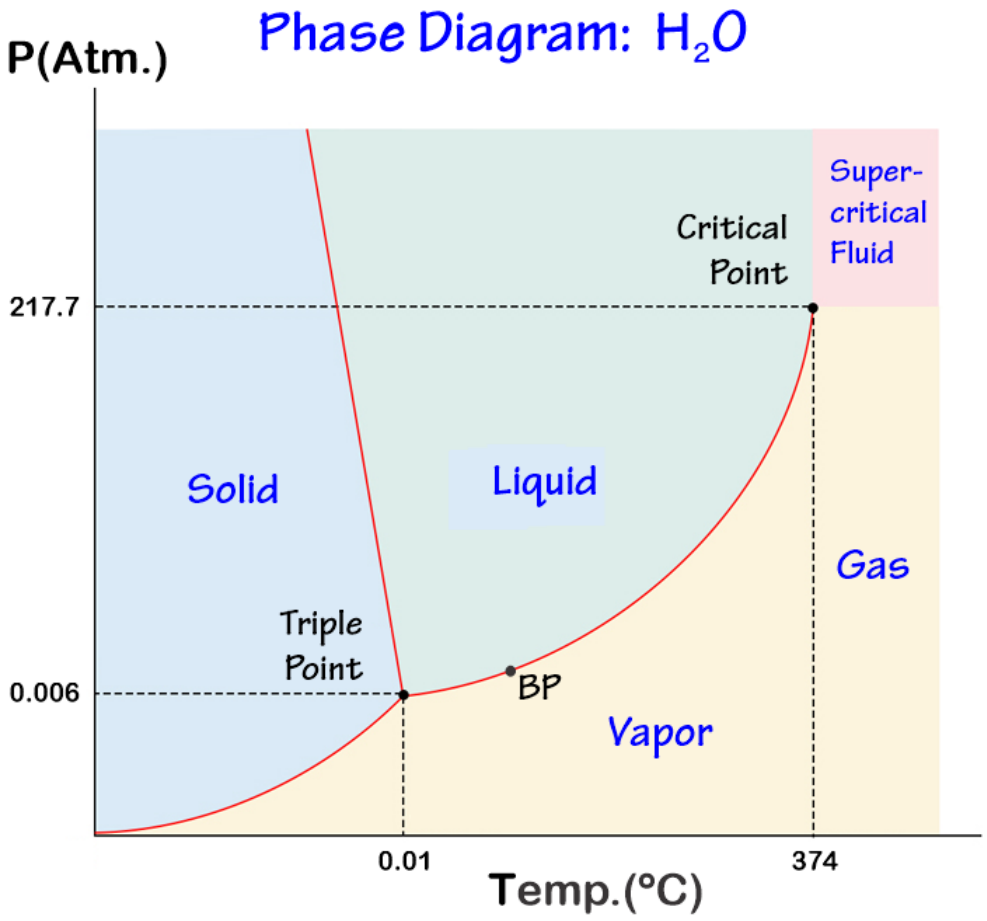
- is the temperature above which a gas cannot be liquified by pressure alone
  - i. N_2O = 36.5 °C
  - ii. O_2 = -119 °C

■ *Critical Pressure*

- is the pressure at which a gas liquifies at its critical T
  - **i. N_2O ≈ 73 bar @ 36.4 °C**
  - ii. N_2O ≈ 52 bar @ 20.0 °C

Def'n: ***A Gas***: a substance in the gaseous phase ***above*** its critical T

A Vapour: a substance in the gaseous phase ***below*** its critical T



NB: **TP_{H2O}: 0.01°C (273.16K)**

FP_{H2O} : 0°C (273.15K) @ 1.0 Atm

BP_{H2O} : 100°C @ 1.0 Atm

■ *Pseudo-Critical Temperature*

- for a mixture of gases at a ***specific pressure***, the ***specific temperature*** at which the individual gases may separate from the gaseous phase

1. N_2O 50% / O_2 50% = - 5.5 °C for cylinders (most likely at 117 bar)
2. N_2O 50% / O_2 50% = - 30 °C for piped gas

Filling Ratio: = $\frac{\text{the mass of the gas in the cylinder}}{\text{the mass of water which would fill the cylinder}}$

N_2O = **0.65** (UK/AUS)

■ *Adiabatic Change*

- the change of physical state of a gas, without the transfer of heat energy to or from the surrounding environment
- in rapid ***expansion***, energy is required to overcome Van der Waal's forces of attraction, as this energy cannot be gained from the surroundings, it is taken from the kinetic energy of the molecules → basis of the ***cryoprobe***
- in rapid ***compression***, the energy level between molecules is reduced, as this energy cannot be dissipated to the surroundings, it is transferred to the kinetic energy of the molecules

SOLUBILITY

■ *Bunsen Solubility Coefficient*

Def'n: the volume of gas, corrected to ***STP***, which dissolves in one unit volume of the liquid at the temperature concerned, where the partial pressure of the gas concerned is 1 atmosphere

■ *Ostwald Solubility Coefficient*

Def'n: the volume of gas which dissolves in one unit volume of the liquid at the temperature concerned

i. the ***temperature*** must be specified

ii. it is independent of ***pressure***

- as the pressure rises the number of molecules of gas in the liquid phase increases, however, when measured at the higher pressure the volume is the same

■ *Partition Coefficient*

Def'n: the ratio of the amount of a substance present in one phase as compared with than in another, the two phases being of ***equal volume***, the ***temperature*** must be specified and the phases in ***equilibrium***

DIFFUSION & OSMOSIS

■ *Diffusion*

- the spontaneous movement of molecules or other particles in ***solution***, owing to their random ***thermal motion***, to reach a uniform concentration throughout the solvent

■ *Fick's Law*

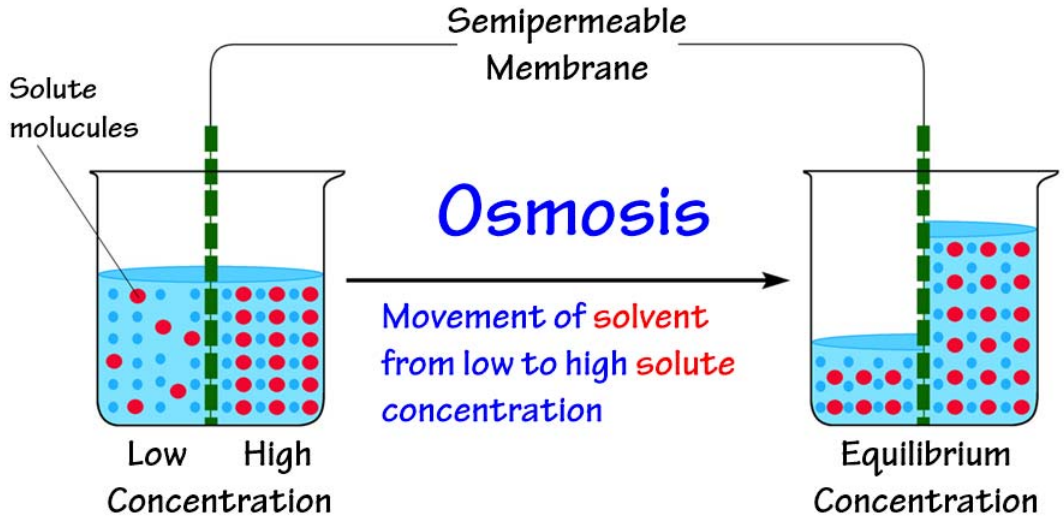
- the rate of diffusion of a substance across a unit area is proportional to the ***concentration gradient*** for that substance
- further, the diffusion of gas across a membrane, or into or out of a liquid, is proportional to the gases solubility in the liquid
- CO_2 being more soluble than O_2 diffuses far more rapidly across the alveolar membrane and into the RBC
- N_2O being far more soluble than N_2 may diffuse into and expand closed cavities during induction of anaesthesia

■ *Graham's Law*

- the rate of diffusion of a gas is inversely proportional to the square root of the ***molecular weight***
- this only applies to simple models and is inaccurate when dealing with complex biological membranes

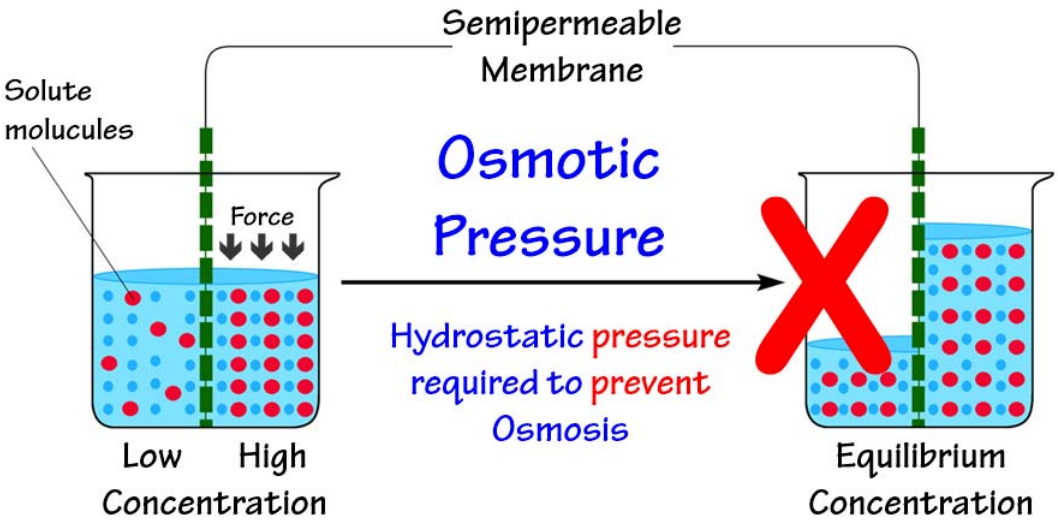
■ *Osmosis*

- the movement of ***solvent*** across a semipermeable membrane, down a ***thermodynamic activity*** gradient for that ***solvent***



■ *Osmotic Pressure*

- the pressure which would be required to ***prevent*** the movement of ***solven***t across a ***semipermeable membrane***, down a thermodynamic activity gradient for than solvent



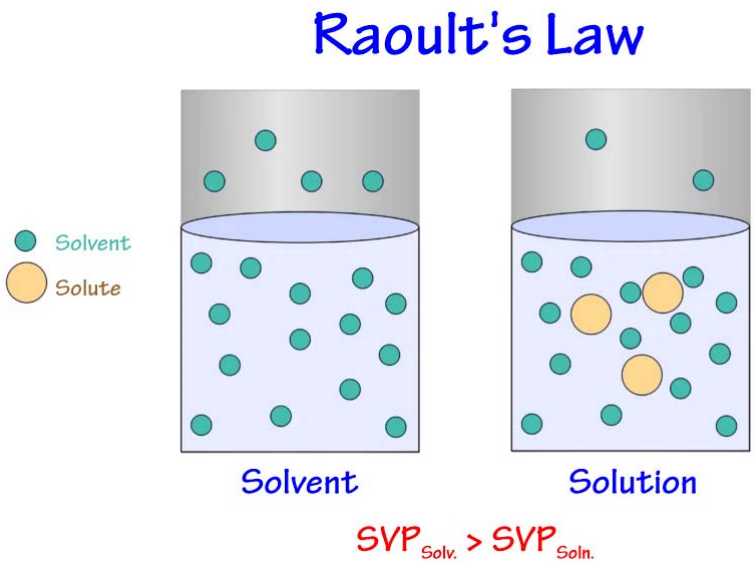
- 1 mol of any solute dissolved in 22.4 litres of solution at 0°C will generate an osmotic pressure of 1 atmosphere, *or*
- 1 mol in 1 L will produce an osmotic pressure of 22.4 atmospheres
- plasma osmotic pressure: 280 msom/L → (0.28 x 22.4 x 760) ≈ 4,770 mmHg
- actual values are higher: **> 5,000 mmHg**
    - T = 37°C
    - Gibbs-Donnan

- in mixed solutions the osmotic pressure is the sum of the individual molarities
- over 99% of the plasma osmolarity is due to ***electrolytes***, the contribution of the plasma proteins being only ~ 1 mosmol/l
- normal rbc's lyse at osmolarities ≤ 200 mosmol/l
- as capillaries are relatively impermeable to protein, this generates an osmotic pressure difference between the plasma and the interstitial fluid,
  the plasma ***oncotic pressure*** **~ 26 mmHg**

■ *Osmolality*

- the number of osmotically active particles (osmoles) per ***kilogram*** of solvent
- the depression of the freezing point of a solution is directly proportional to the osmolality,
  - → 1 mol of a solute / 1 kg of water
  - → ↓ freezing point ≈ **1.86°C**

- the presence of increased amounts of solute also lowers the vapour pressure of the solvent, viz.



■ *Raoult's Law*

- the depression or lowering of the ***vapour pressure*** of a ***solvent*** is proportional to the molar concentration of the ***solute***
- as the presence of a solute decreases the vapour pressure, making the solvent less volatile, so the boiling point is raised

NB: these phenomena,

i. depression of ***freezing point***
ii. depression of ***vapour pressure***, and
iii. elevation of ***boiling point***

being related to osmolality are termed ***colligative properties*** of a solution

■ *Azeotrope*

- is a ***mixture***, from which the component liquids vaporise in the same proportions as the molar ratios in the mixture
- ***ether & halothane*** form an azeotrope when the volume and the molar concentration ratios are both **1:2**
- alcohol & water form an azeotrope when the volume % alcohol is ~ 96%
- this makes it impossible to prepare alcohol solutions > 96% by ***fractional distillation***

SPECTROPHOTOMETRY

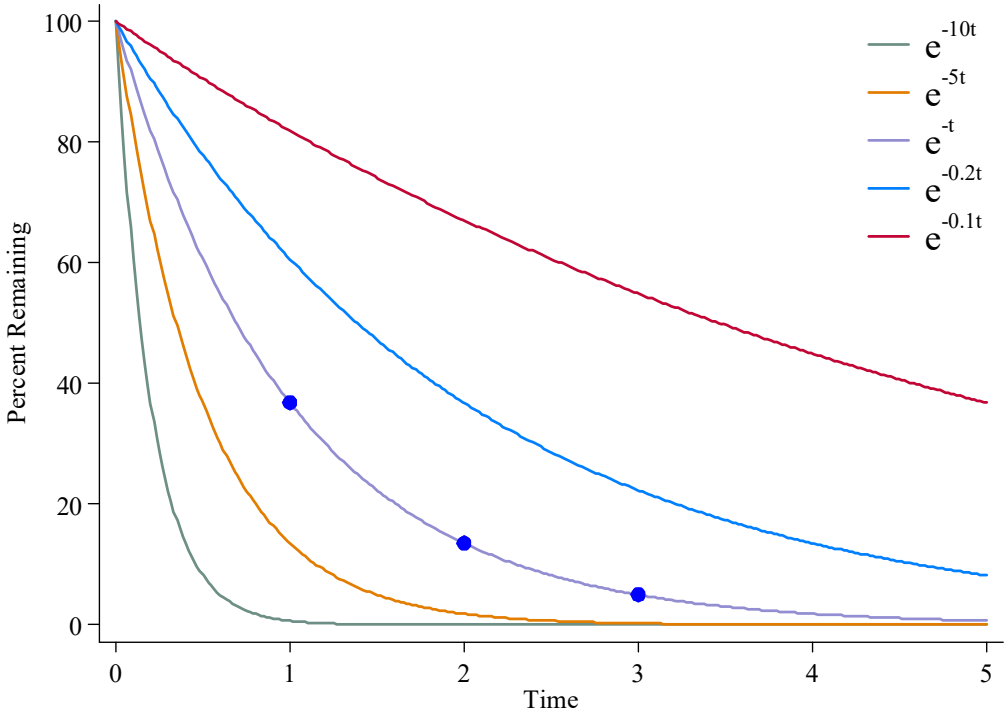
Exponential Decay

- a process where a constant ***number of discrete items*** of a substance are removed per unit time
- represented by the following ***differential equation***, viz:

$$\frac{dN}{dT} = -\lambda N$$

- applying implicit differentiation, giving the equation:

$$N_t = N_0.e^{-\lambda t}$$



■ *Half Life*

Def'n: time taken for the number/amount to fall to half the current level

$$N_{t_{1/2}} = N_0.\frac{1}{2}$$

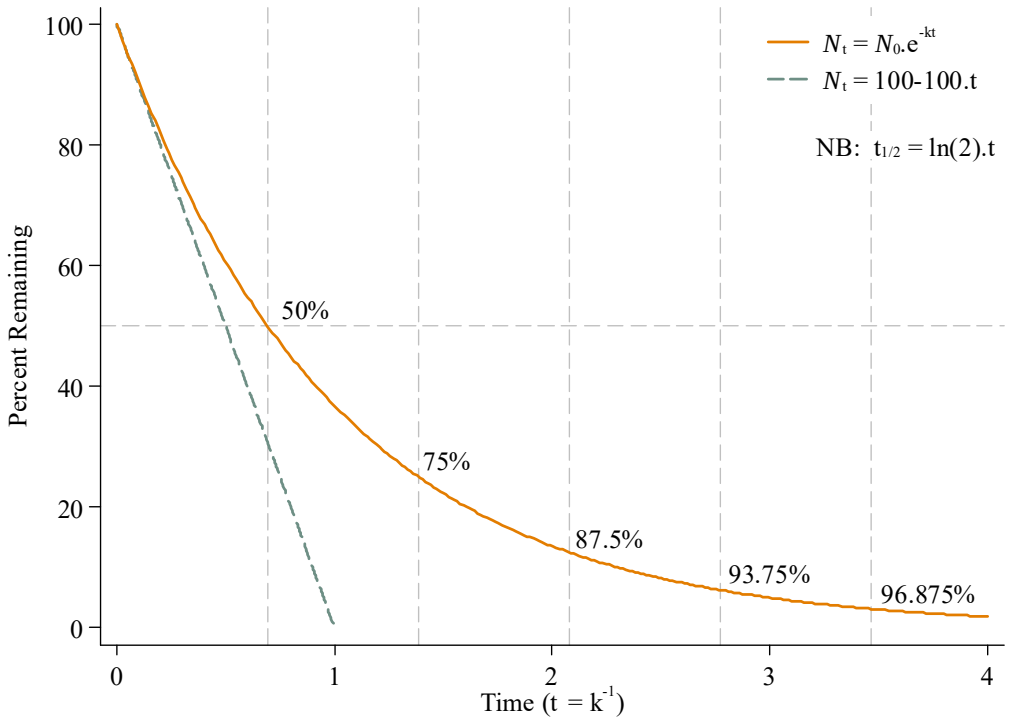
$$N_0.\frac{1}{2} = N_0.e^{-\lambda t_{½}}$$

$$\frac{1}{2} = e^{-\lambda t_{½}}$$

$$\ln(\frac{1}{2}) = -\lambda t_{½}$$

$$\lambda t_{½} = -\ln(\frac{1}{2}) = \ln(2)$$

$$t_{½} = \ln(2)/\lambda$$



■ *Time Constant*

Def'n: the ***time taken*** for a process to,

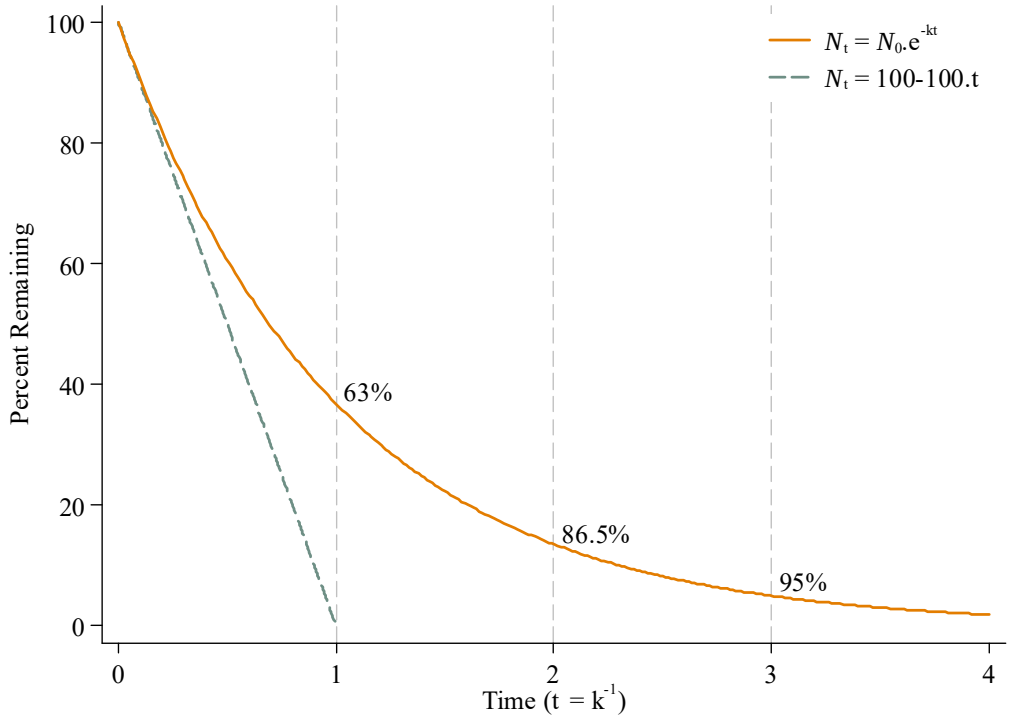
- run to ***completion*** if the initial rate of change continued unabated, *or*
- reach **≈ 63%** of its final (asymptotic) value, *or*

Def'n: the ***inverse*** of the elimination rate constant (λ),

$$\tau = 1/\lambda$$

- as, $t_{½} = ln(2)/\lambda$ so,

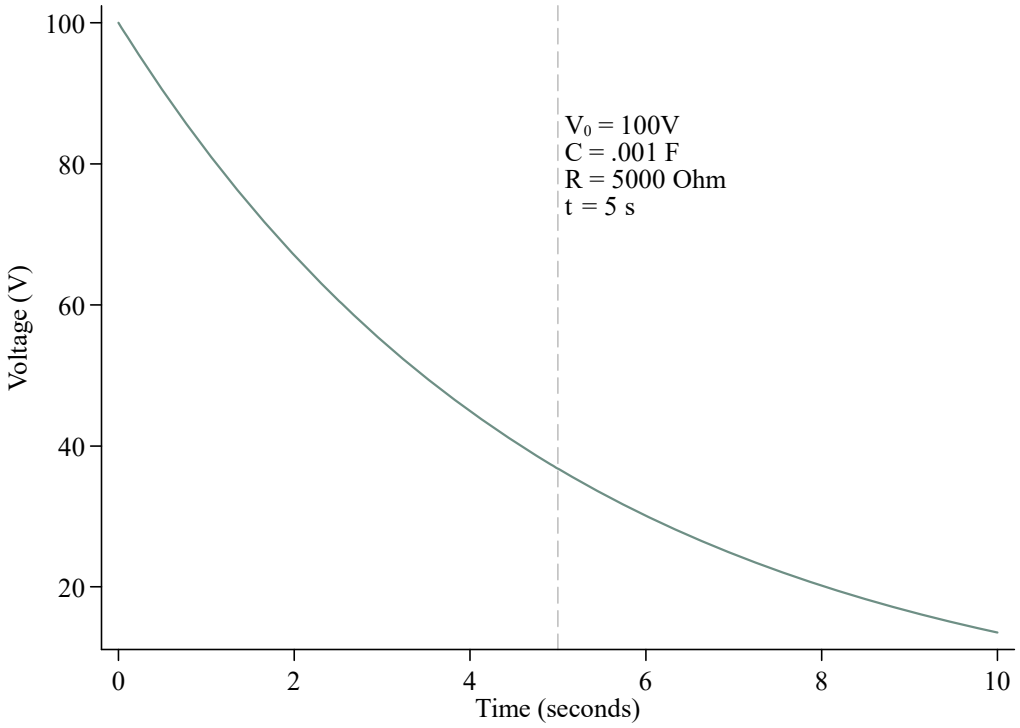
$$t_{½} = 0.693 \cdot \tau$$



- in physics and engineering, it characterizes the frequency response of a first-order, linear time-invariant (LTI) systems (RxL and RxC)
- classic physiology example is the time-constant of the lung,

$$\tau = R_L.C_L$$

- taken from electrical theory for voltage decay for a capacitor (Farads) discharging through a fixed resistance (Ohms):



■ *Exponential Decay*

- percent remaining after 'x' period intervals:

Half-lives	Time Constants
1. 50%	1. 37%
2. 25%	2. 13.5%
3. 12.5%	3. 5%
4. 6.25%	
5. 3.125%	

■ *Exponential Examples*

1. Atomic decay:

$$N_t = N_0.e^{-\lambda t}$$

2. Plasma drug levels:

$$C_t = C_0.e^{-k_{el}t}$$

3. Voltage decay (RC):

$$V_t = V_0.e^{-t/(R.C)}$$

4. Absorption of light (Beer-Lambert Law):

$$I_t = I_0.e^{-\alpha CL}$$

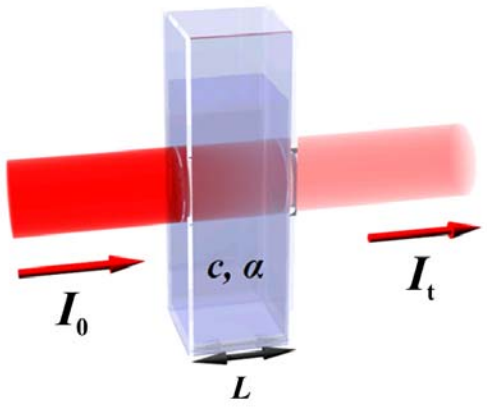
5. Exponential ***wash-in*** *where C_{In} = inflow/final concentration, e.g. alveolar

$$C_t = C_{In}(1 - e^{-\lambda t})$$

■ *Beer-Lambert Law*

1. Lambert's Law c1760 *Pierre Bouguer 1729
   - absorption ∝ ***length*** of light absorbing path
2. Beer's Law c1852
   - absorption ∝ ***concentration*** of absorbing species in solution
3. Combined as the Beer-Lambert Law
   - first used to determine the [Hb] the 1930's
   - solving for C,

$$C = \ln(I_0/I_t) / \alpha L$$



■ *Beer-Lambert: Pre-requisits*

1. I_0 measured
2. I_t measured
3. α known for given solute at specified wave-length (λ)
4. L path-length known
5. No particles in solution → ***no scattering***
6. No other light absorbing species present, *or*

   The number of ***wavelengths*** (λ_1, λ_2, λ_3, λ_4... λ_n) ≥ number (n) of ***solutes*** present

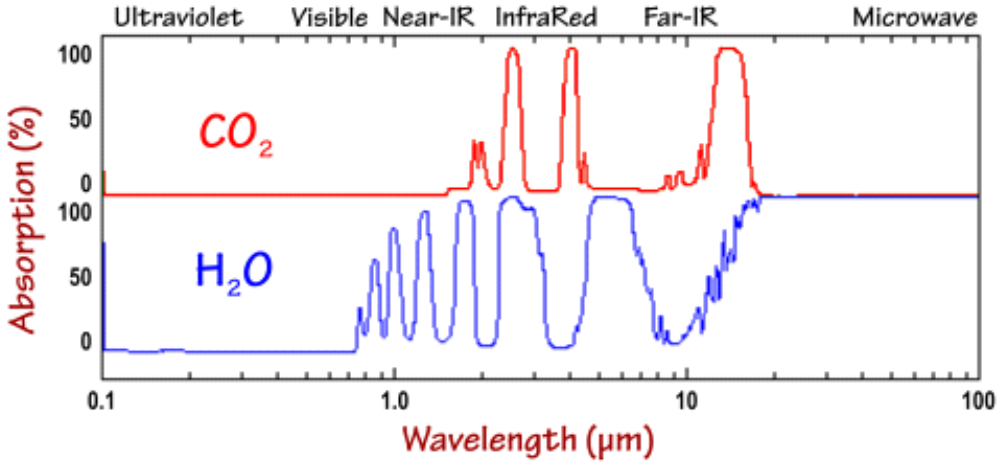
   → solve simultaneous B-L equations

■ *Beer-Lambert: Clinical Uses*

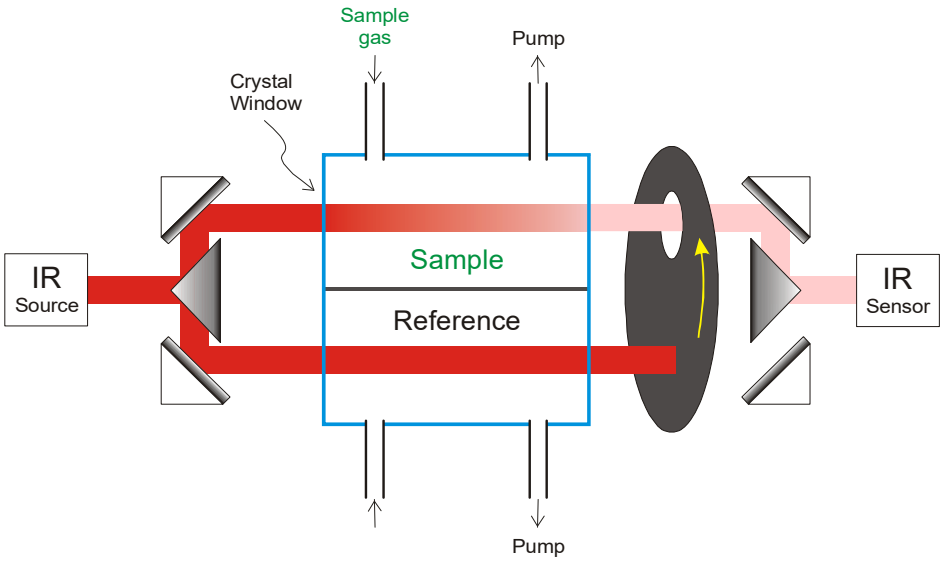
1. CO_2 in gas samples
2. Co-oximetry - ABG analysis
3. Hemocue® [Hb]
4. Volatile anaesthetic agent monitoring (UV)

Capnometry (IR)

- polyatomic gases ≥ 2 molecules (CO_2, H_2O, N_2O) → absorb IR radiation (> 1 µm)
- absorbance peak for CO_2 ≈ 4.28 µm



- chosen to minimize interference from H_2O in respiratory gases
- 1st IR CO_2 apparatus by ***Luft*** in 1943
- ASA closed claims > **90%** of anaesthetic mishaps preventable by **$ETCO_2$ & SpO_2**



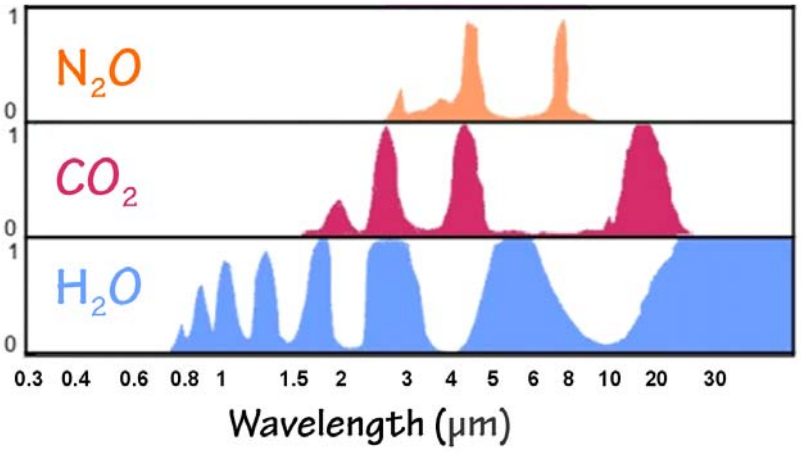
- glass absorbs IR radiation, ∴ chamber windows made of sodium chloride/bromide crystal
- calibration achieved by filling the chamber with a CO_2 free gas, or by splitting the incident beam and passing this through a reference chamber
- reference beam also compensates for variations in IR source output
- the sample chamber is made small → ↓ time constant (τ)
  → response time ≈ 100 ms
- enabling end-tidal CO_2 estimation & continuous waveform analysis

■ *Classification*

1. Side-stream → sensor in main unit
   - gas aspirated from circuit
   - sampling flow rate may be - high > 400 mL/min, or
     - low < 400 mL/min
   - optimal gas flow ≈ 50-200 mL/min
   - ensuring reliability with both adults and children
   - exhaust gases contain anaesthetic agents & should be scavenged
2. Main stream → sensor/curvette within the circuit
   - heated > 39°C to prevent water condensation
   - tend to be bulky
   - add dead space
   - expensive if dropped & broken
   - no mixing of gases occurs during sampling → more rapid response time

■ *Sources of Error*

1. H_2O
2. N_2O
   - absorbance peak ≈ 4.5 µm
   - minimized by "narrow bandwith filter"
   - now more commonly LED source → ≈ monochromatic



3. P_{Atm} / $P_{Circuit}$
4. Other
   - O_2
   - alinearity
   - volatile anaesthetic agents

NB: both H_2O & N_2O → ***collision broadening*** of CO_2 absorbance peak
→ ↑ PCO_2 readings

■ *Photo-Acoustic PCO_2*

- first described by Alexander Graham Bell (circa 1880)
- thin discs emitted sound when exposed to a beam of sunlight that was rapidly interrupted with a rotating slotted disk
- the absorbance of IR light by CO_2 → gas expansion / ↑P
- IR light pulsed at acoustic frequencies → energy absorbed detected by a microphone
- claimed advantages over IR spectrometry,
  - light absorption measured directly without the need for a reference chamber
    → no zero point drift
  - higher accuracy
  - increased reliability
  - reduced maintenance & reduced need for calibration

Co-oximetry: ABG Hb-Saturation

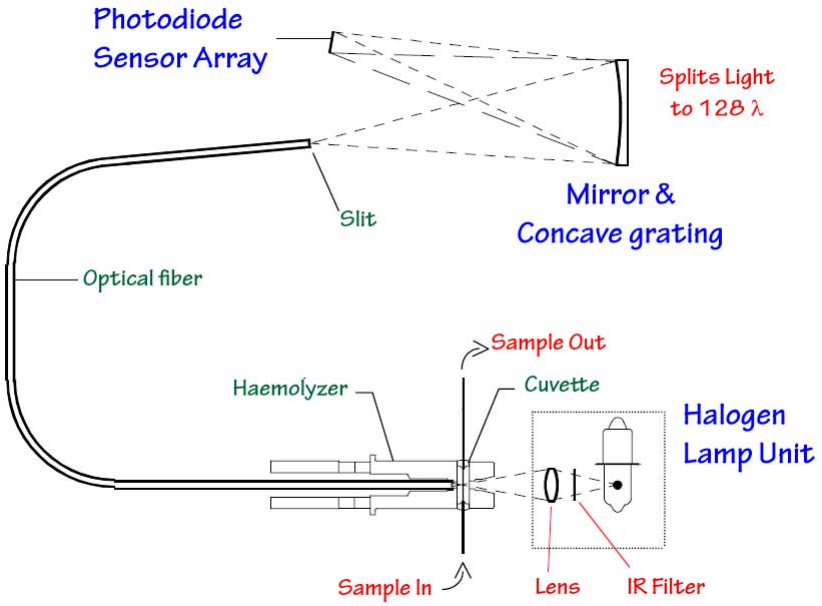
- CaO_2 was originally measured ***volumetrically*** by the method of Van Slyke and Neill
- functional oxygen ***saturation*** **sO_2** is defined as the CaO_2 / the oxygen capacity
  → includes O_2Hb and small quantity of dissolved O_2
- now taken as:

$$sO_2(\%) = 100 \times \left(\frac{cO_2Hb}{cHb + cO_2Hb}\right)$$

- normal adult blood contains four species of Hb: {Hb, O_2Hb, metHb, COHb}
- the later two are normally found only in low concentrations, except in disease states, and are ineffective in the transport of oxygen
- ***fractional*** **SaO_2** includes both met-Hb and CO-Hb in the denominator

$$FO_2Hb(\%) = 100 \times \left(\frac{cO_2Hb}{cHb + cO_2Hb + cCOHb + cMetHb}\right)$$

- early ABG machines employed 4-wavelengths and solved 4x B-L equations
- modern devices use multiple wavelengths (ABL800 → **128λ**)
  - range λ - 478-672 nm
    - within visible light spectrum (400-700nm)
  - multiple λ - for each Hb species, and
    - non-Hb light absorbing species (bilirubin, urea, creat., etc.)



Oximetry

- Kramer first estimated SaO_2 in animals in the 1930's
- Karl Matthes (1936) estimated SaO_2 using red and blue-green transillumination of the ear
- Millikan *et al*. in the 1940's
    a. coined the term ***oximeter***
    b. WWII project ∝ loss of consciousness of RAF pilots during dog-fights
    c. developed a lightweight smaller version of Matthes' design
    d. transillumination of the earlobe with 2 wavelengths of light
        i. red & green filters covering Kramer's photo-cells
        ii. signal from the green filter later shown to be IR
    e. oximeter which measured SaO_2

- two major technical problems:
    1. many non-Hb light absorbers in tissue
    2. tissues also contain capillary & venous blood, c.f. arterial

- these were overcome by:
    1. first measuring a ***bloodless baseline***
        - absorbance while earlobe compressed to remove all blood
    2. then earlobe heated (43ºC) to "***arterialize***" the blood

- this device was shown to accurately predict intra-operative desaturations, however, due to the technical difficulties was never adopted on mass

- 1970s Hewlett-Packard (now Agilent) developed an **8×λ** device solving simultaneous B-L equations
    → however, still needed ***empiric*** calibration coefficients for accuracy

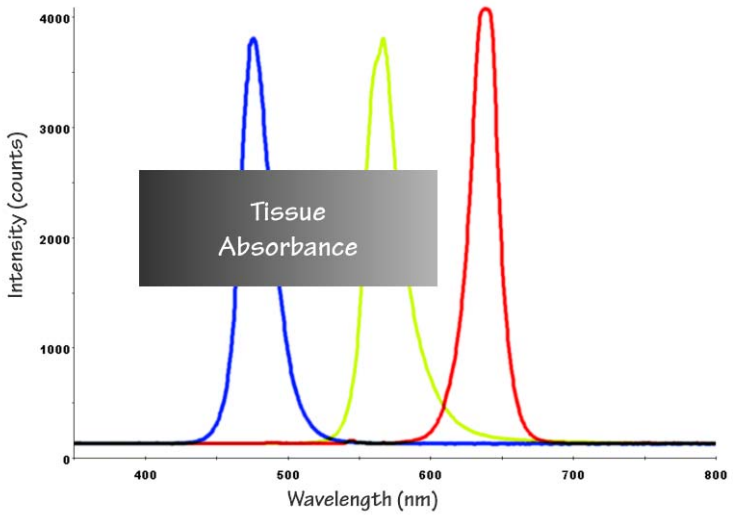
Pulse Oximetry: SpO_2

- in the mid 1970's, the Japanese engineer Takuo Aoyagi working at Nihon Khoden:
  a. studying CO estimation by dye-dilution (indocyanine green) → ear densitometry
  b. noted interference from ***pulsatile components*** of the red & IR absorbances
  c. interference related to the SaO_2 / $Sv'O_2$
  d. first prototype built 1974 and patented by N-K in 1978
     - T. Aoyagi and M. Kishi, "Optical device for measuring arterial oxygen saturation" Japan Patent, 53-26437, 1978
     - interesting that lab "assistant" was included on patent, along with manager?!

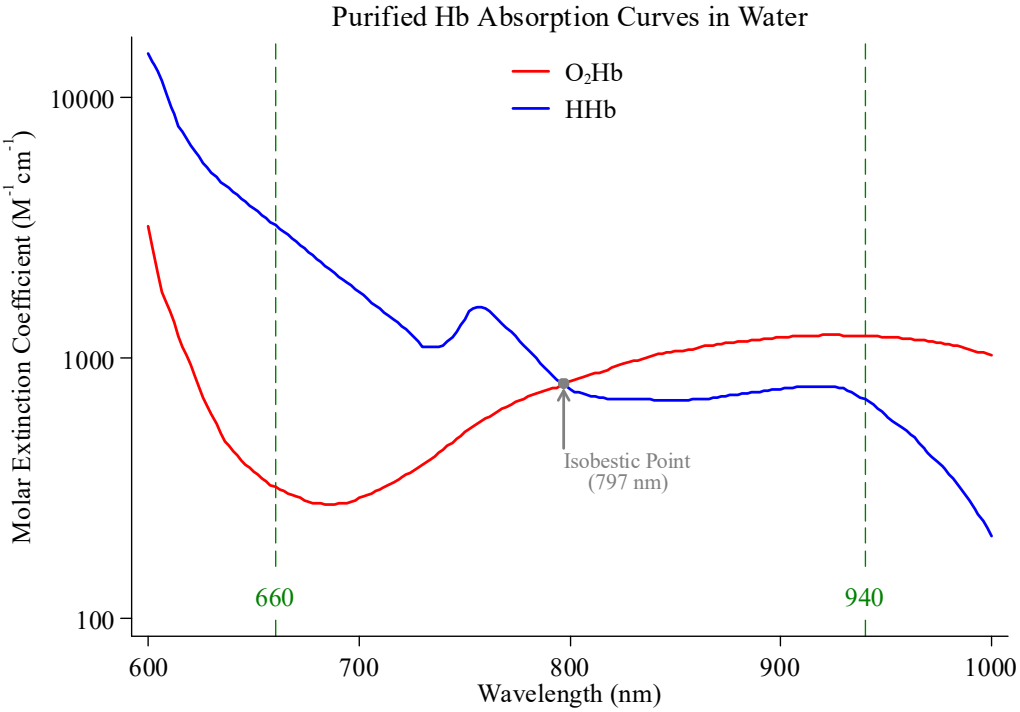
  NB: pre-requisites for the B-L ***cannot*** be met and the relationship is empiric!

     - do not know/measure I_o
     - path-length unknown
     - multiple other light absorbers and particles → scattering

- relationship holds where measurement with ≥ 2 wavelenghts of light:
  a. sufficient intensity to trans-illuminate finger/body tissue
  b. wavelengths in the red → IR range avoid tissue/H_2O absorbance
     i. Red (600-700nm)
     ii. Near IR (700-900nm) | Far IR (900-1000nm)
     iii. c.f. ABL800 oximetry (478-672nm) → tissue absorbance irrelevant



  c. sufficiently different absorbance at $\lambda_{1/2}$ for O_2Hb/Hb
     - ideally isobestic point (equal absorbance) & greatest separation
     - isobestic point often quoted as 806nm, actually ≈ 797 nm
     - **MEDAL**: **M**ore than **E**ight-hundred nm, **D**eoxy-hb **A**bsorbs **L**ess



d. in 1970s, the only commercially available and suitable LEDs

   i. red: $\lambda =$ 660 nm

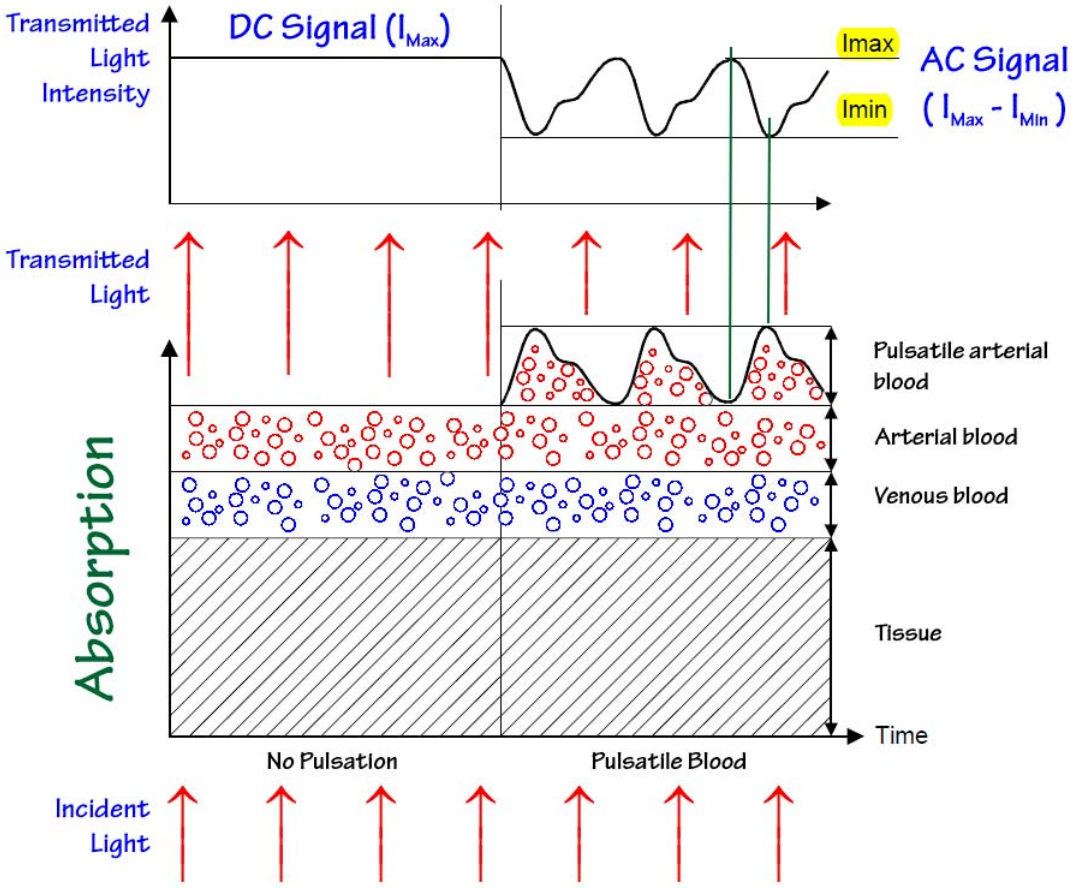
   ii. IR: $\lambda =$ 940 nm

e. attempts to improve performance with up to **5x λ** → no benefit

   - Aoyagi himself spent years trying to improve accuracy to no avail

f. modern machines may use slightly different $\lambda_{1/2}$ as LED technology has advanced and multiple additional wavelengths now available

- signal is divided into two components,
  a. DC = tissue + capillary blood + venous blood + non-pulsatile arterial blood
  b. AC = pulsatile arterial blood **≈ 1-2% of I_{Max}**



Ref: Datex Ohmeda E-modules Technical Reference Manual

- for each wavelength, the oximeter determines the AC/DC fraction
  → independent of the incident light intensity = ***"pulse added absorbance"***
- then the ratio (R) of these is calculated, commonly incorrectly stated as,

$$R = \frac{(AC/DC\ absorbance)_{Red}}{(AC/DC\ absorbance)_{IR}}$$

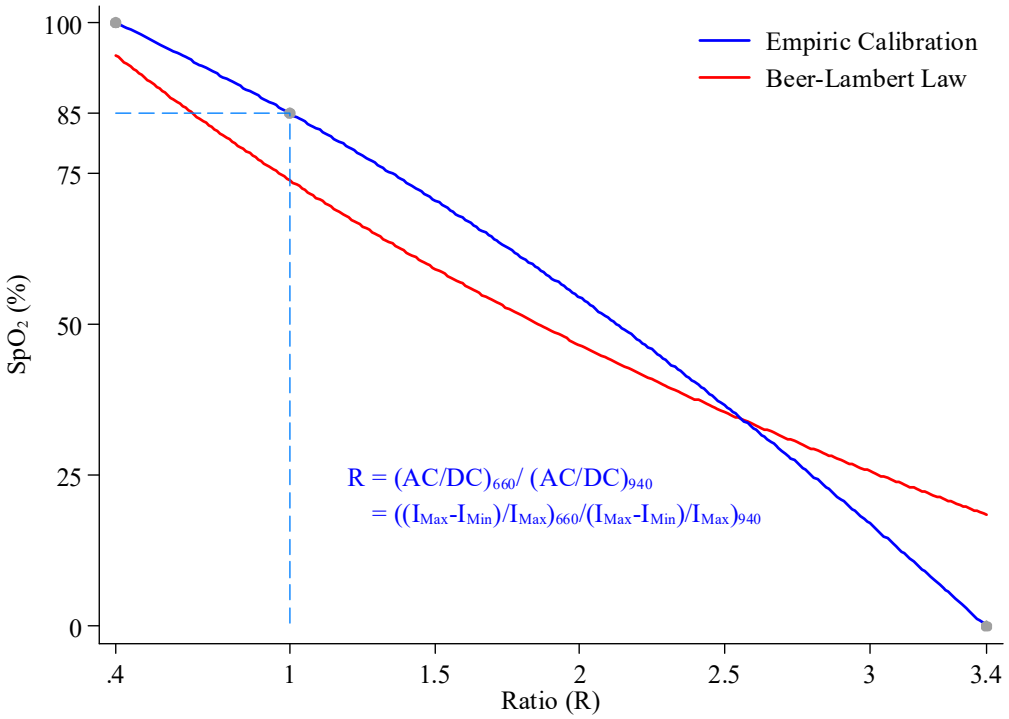
$$= A_{660nm} / A_{940nm}$$

NB: this is a misnomer and most descriptions incorrectly state 'absorbance'

- ***absorbance*** requires knowing incident I_0 which is ***unknown***
- as shown above, the only measurement is ***transmitted intensity*** I_t
- pulse added "absorbance" → ↓ I_t ≈ 1-2% of DC intensity (I_{Max})

- more correctly, AC/DC should reference measured light ***intensity***:

$$R = ((I_{Max} - I_{Min})/I_{Max})_{660}/((I_{Max} - I_{Min})/I_{Max})_{940}$$



- calibration values vary with device and algorithm (esp. LED $\lambda_{1/2}$), but commonly:
  a. $SaO_2 = 100\%$ R = 0.4
  b. $SaO_2 = 85\%$ R = 1.0
  c. $SaO_2 = 0\%$ R = 3.4

- empiric calibration assumes only 2 major light absorbing Hb species in arterial blood:
  a. MetHb or COHb will contribute to the AC component
  b. CO absorbs very little light at 940nm (IR) but ≈ O_2Hb at 660nm (red)
     - → ↑ [CO-Hb]→ $SpO_2 \approx SaO_2 + (0.9 \times COHb)$
  c. MetHb absorbance is high at both wavelengths → ↑ AC_{660nm} & AC_{940nm}
     - ∴ → R ≈ 1.0
       $SpO_2 \approx 85\%$
  d. HbF has a greater affinity for O_2 than HbA
     - however, the absorbance coefficient ≈ identical
     - ∴ HbF does not affect the SpO_2 reading
     - only important if the aim of therapy is a specific P_{aO2}, c.f. a specific SaO_2

- ***ambient light*** is also registered by the photo-detector diodes:
    a. interference reduced by ***cycling***: red only → IR only → both off
    b. repeated at 480-1000 Hz (not a multiple of 50Hz)
    c. enables subtraction of ambient light (even when this is oscillating)
    d. despite this filtering, ambient light can produce interference so the sensor is usually covered with an opaque material

- ***low signal strength*** limits estimation of AC components (≈ 1-2% DC)
    a. automatic gain circuitry → amplification of low signal strengths
    b. allows SpO_2 estimation, however → ↓ ***signal to noise ratio*** (SNR)
    c. some meters give "low signal strength" warnings to alleviate this problem
    d. laser-Doppler flow studies show oximeters will estimate SpO_2 down to ≈ 8% of the control pulse strength, over a wide range of CO values

- ***patient motion artifact*** is another cause for low signal-noise
    a. most oximeters employ signal averaging circuitry to adjust for this
    b. however, increasing the signal averaging time → ↓ device response time

- the final source of error is LED wavelength variability,
    a. specified λ ± 10nm
    β. → probe variation in accuracy

- manufacturers claim accuracies around, *e.g. Datex-Ohmeda
    a. SpO_2 ≈ 100% to 70% → **± 2%**
       ± 3% with patient motion
    b. SpO_2 < 70% → unspecified

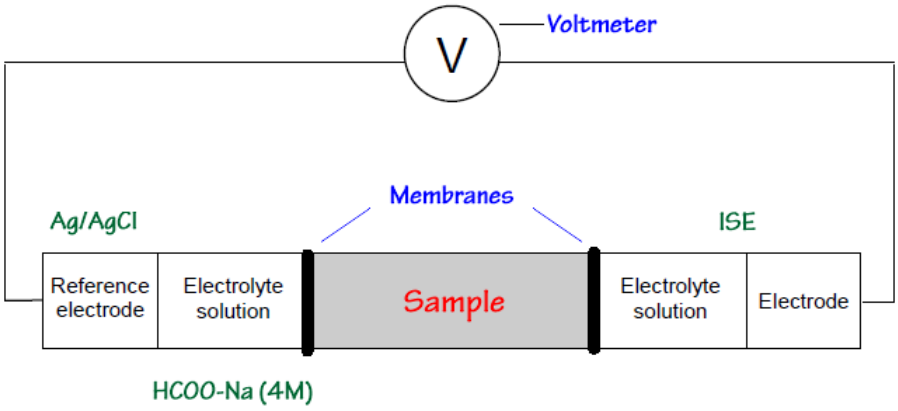
■ *Limitations of Pulse Oximetry*

a. SpO_2 does not indicate oxygenation unless [Hb] is known
b. insensitive to directional changes in P_{aO2} above 80 mmHg
c. due to automatic gain, oximetry is insensitive to perfusion
d. errors of saturation estimation
    i. signal to noise ratio - low perfusion pressure
       - motion artifact
    ii. light artifact
    iii. dyshaemoglobins
    iv. intravenous dyes
    v. probe variability errors

BLOOD-GAS MEASUREMENT

Potentiometric Method

- the potential of an electrode chain is recorded using a voltmeter, and related to the concentration of the sample *Nernst equation ***not V_m equilibrium!**
- an ***electrode chain*** circuit consists of a sample, electrode, reference electrode, voltmeter, membranes, and electrolyte solutions:



- every element in the electrode chain contributes to the total δV:
  - with appropriate electrolyte solutions, both electrodes have separate potentials
  - membrane junctions between the sample and electrolyte solutions also have separate potentials

NB: The potentiometric principle is applied to

pH, pCO_2, and electrolyte electrodes

(*Radiometer ABL800 Flex Manual)

$$E_{Sample} = E_0 + \frac{2.3RT}{zF} \cdot \log(a_x)$$

where:

E_0	=	reference electrode potential	
R	=	gas constant	(≈ 8.314 $J.K^{-1}.mol^{-1}$)
T	=	absolute temperature	(≈ 310K / 37°C)
z	=	ion charge/valence	
F	=	Faraday constant	(≈ 96487 $coulomb.mol^{-1}$)
a_x	=	***activity*** of x	

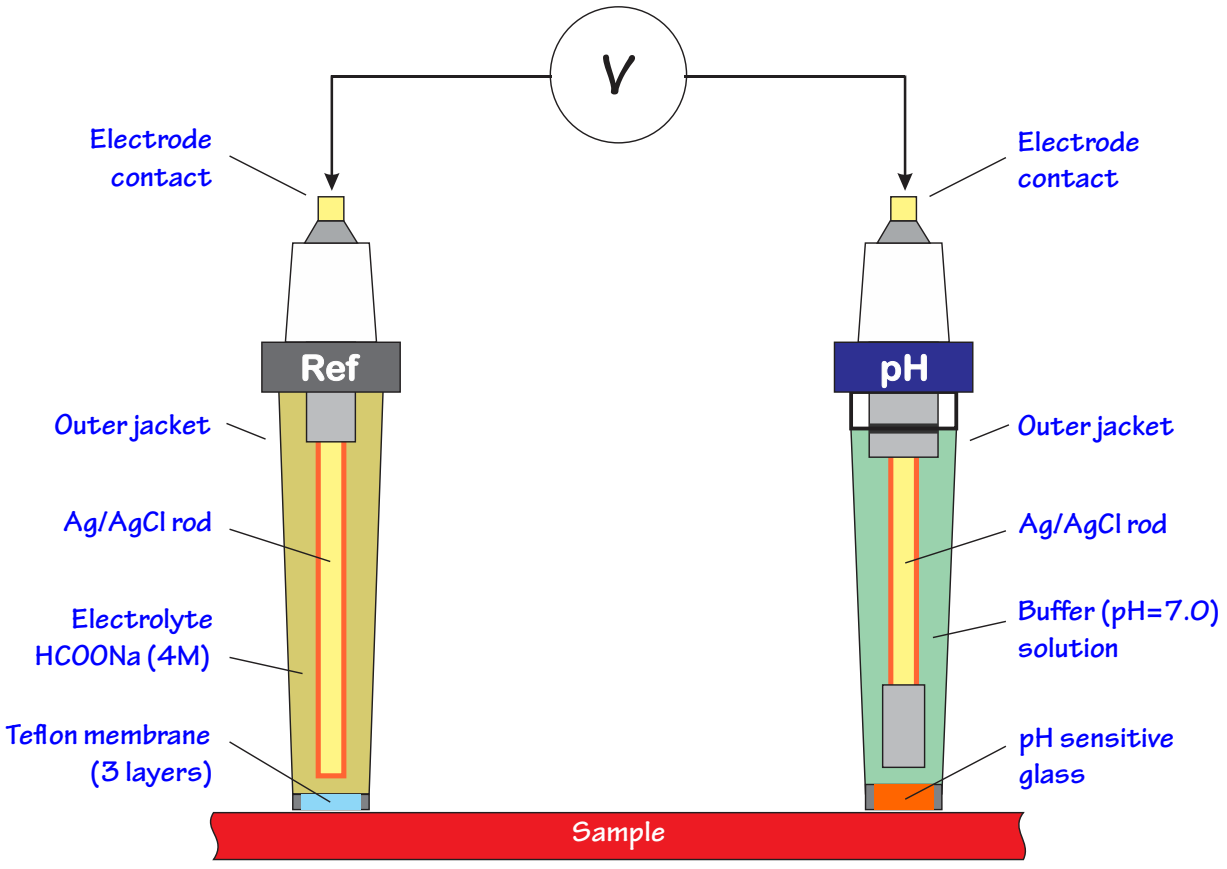
- and the ***activity*** $a_x = \gamma c_x$ where γ the ***activity coefficient*** = 1 for ideal systems
- forms basis for all ***ion-specific electrodes (ISE)***

Measurement of pH

Def'n: pH = $-\log_{10}[H^+]$ *actually H^+ ***activity*** but for dilute solutions $\cong [H^+]$

$\approx 7.4 \pm 0.04$ arterial blood 37°C

- the circuit consists of,
  - a. a ***reference electrode***
    - Ag/AgCl - most common, *or*
      Hg/Hg_2Cl_2 (calomel)
    - + an electrolyte solution - KCl or HCOONa (sodium formate)
  - b. an ***ion specific electrode***
    - capillary tube of pH sensitive glass $\rightarrow \delta V$
    - a reference buffer solution + Ag/AgCl electrode
  - c. a surrounding water jacket at 37°C
  - d. a voltmeter: **$dV_{Glass} \approx -61.5\ mV \times pH$**



- the electrodes are metal/metal chloride, which are then in contact with electrolyte containing Cl^- to maintain their stability
- the pH difference across the glass produces a potential in proportion to the $[H^+]$ difference

- temperature is important as acids/bases dissociate at higher temperatures altering the pH
- approximated by the formula of ***Rosenthal***,

$$\delta pH \approx \delta T°C \times -0.015$$

- as these are "half-cells", there is ***drift*** and each component contributes a small δV, so regular ***calibration*** is essential
- the calibration solutions give the following two points:

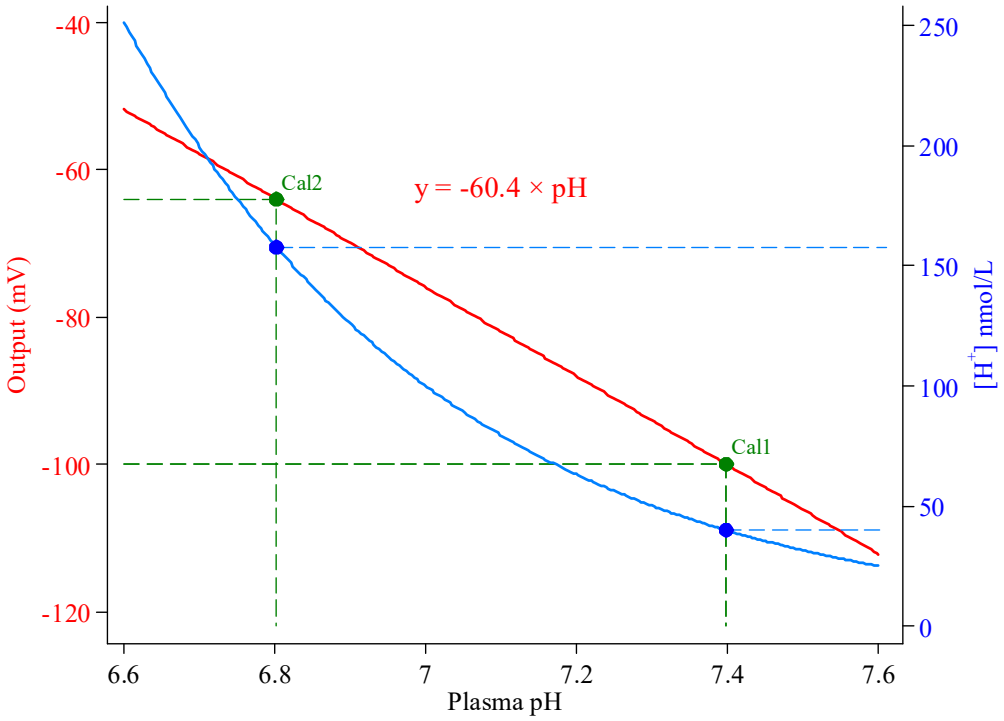
1. Cal 1: pH = 7.398 -100 mV
2. Cal 2: pH = 6.802 -64 mV

- within the range 6.30 to 8.00 the pH electrode is linear
- estimated pH is adjusted by the measured ***sensitivity*** (δV/δpH) during the previous 2-point cal

$$\mathbf{pH_{Sample} = pH_{Cal1} + (E(pH_{Sample}) - E(pH_{Cal1})) / (-61.5 \times Sen(pH))}$$

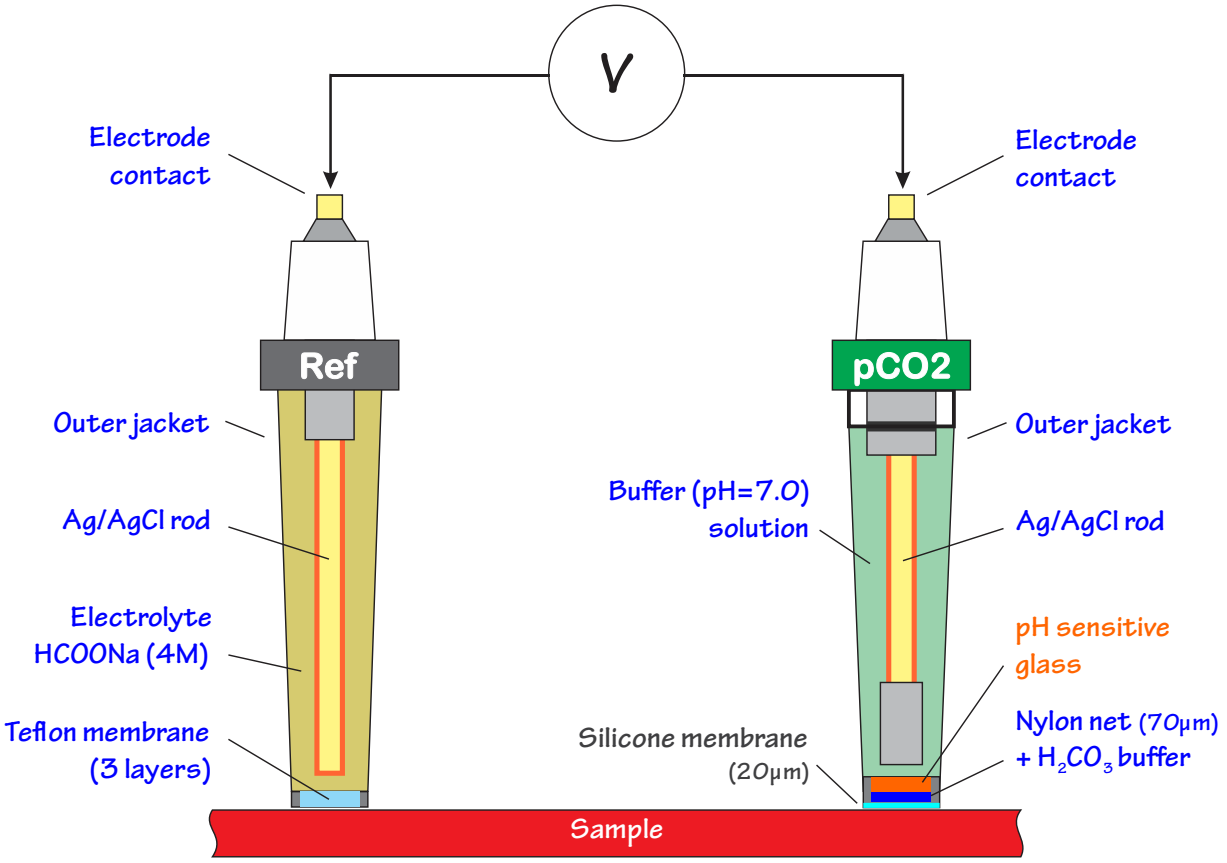
- where,
  - pH_{Cal1} = pH of Cal1 solution
  - $E(pH_{Cal1})$ = potential of the pH electrode chain from a Cal1 calibration
  - Sens(pH) = sensitivity of the pH electrode chain from previous 2-point cal

Ref: Radiometer ABL800 Flex Laboratory Manual



Measurement of P_{CO2}

- the normal $P_{aCO2} \approx$ 40 mmHg (5.3 kPa)
- measurements are based on pH, due to the dissociation of carbonic acid
- the P_{CO2} is therefore related to the $[H^+]$
- the ***Severinghaus CO_2 Electrode*** provides a direct measure of P_{CO2} from the change in pH
- the circuit consists of,
    a. a closed cylinder of pH sensitive glass in the centre
    b. 2 electrodes, 1 inside, the other outside the cylinder
    c. a surrounding solution of sodium bicarbonate
    d. a thin film of bicarbonate impregnated nylon mesh covering the end of the cylinder
    e. a thin, CO_2 permeable membrane covering the end of the electrode



- at the end of the electrode CO_2 diffuses from the blood sample through the membrane into the nylon mesh and by the formation of carbonic acid lowers the pH of the bicarbonate solution
- this change in pH alters the δV across the glass
- as pH changes such that,

$$\delta pH \sim \delta \log_{10} P_{CO2}$$

- the output of the voltmeter can be calibrated in terms of P_{CO2}
- should the end membrane be perforated, then it ceases to be a semipermeable membrane to CO_2 and the reading will be erroneous
- the electrode has an accuracy ≈ 1 mmHg
- the response time ≈ 2-3 mins
- as for the pH electrode, the CO_2 electrode must be kept at 37°C and regularly calibrated with known concentrations of CO_2

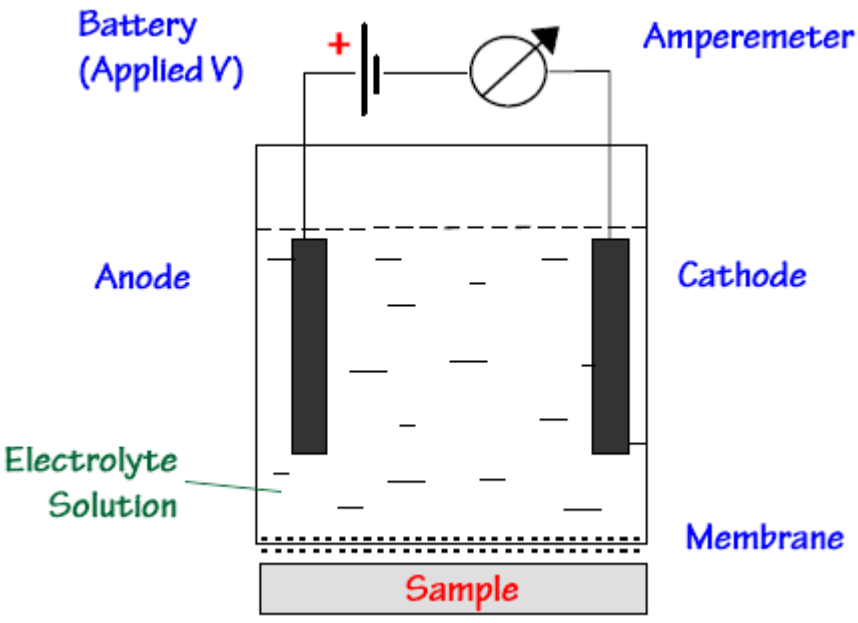
1. Gas 1: 5.61 % CO_2
2. Gas 2: 11.22 % CO_2

NB: $pCO_2 = FCO_2 \times (B_{Gas} - pH_2O)$ B_{Gas} = pressure inside chamber
$pH_2O = VP_{H2O}$ at 37°C (≈6.28 kPa)

- transcutaneous electrodes, similar to the Clark electrode have been used for continuous PCO_2 monitoring

Amperometric Method

- the current flowing through an electrode chain, is proportional to the concentration of the substance being oxidized or reduced at an electrode in the chain
- the chain in amperometric measurements consists of the sample, 2 electrodes (anode and cathode), an amperemeter, a voltage source, the membranes, and the electrolyte solutions:



Measurement of P_{O2}

- methods of measurement include:
    1. Clarke electrode - polarographic (ABG)
    2. Paramagnetic
    3. Raman scattering
    4. Others:
        i. Fuel cell
        ii. Optode - photoluminescence quenching
        iii. Mass spectrometer

■ *Clarke Electrode*

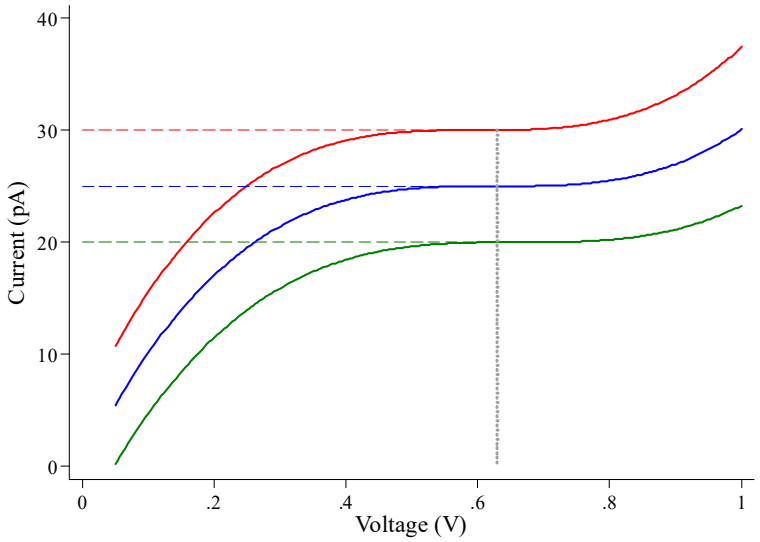
- Leyland Clarke developed an early bubble-oxygenator for cardio-pulmonary bypass
- journal editor refused to publish, as he was unable to provide data on efferent blood PO_2
- in response (1956), he developed the ***polarographic*** oxygen electrode
- prior to this the P_{O2} had not been measured
- the Severinghaus CO_2 electrode was developed in 1958 and ABG analysis was revolutionized
- the amperometric circuit consists of,
    a. DC voltage source (630mV)
    b. ammeter (pA)
    c. platinum cathode (reduction): $O_2 + 2H_2O + 4e^- \rightarrow 4OH^-$
    d. Ag/AgCl anode (oxidation): $4Ag + 4Cl^- \rightarrow 4AgCl + 4e^-$
    e. electrolyte solution (KCl)
    f. O_2-permeable membrane

- as for any resistive circuit:

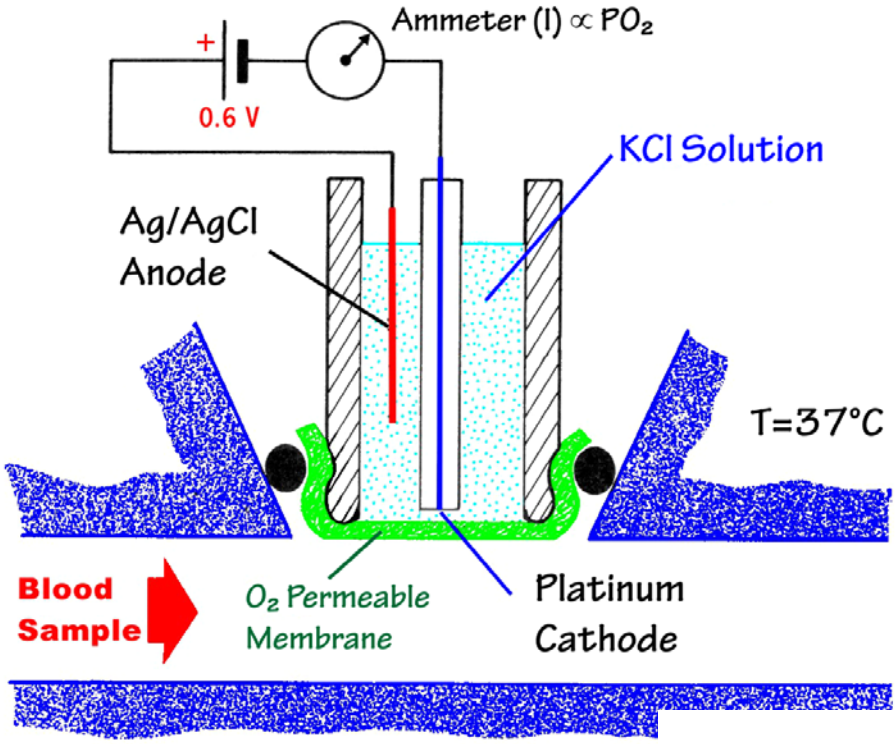
$$\uparrow I \propto \uparrow V$$

- in the above circuit there exists a ***plateau voltage*** range over which the current does not increase with increasing voltage
- however,

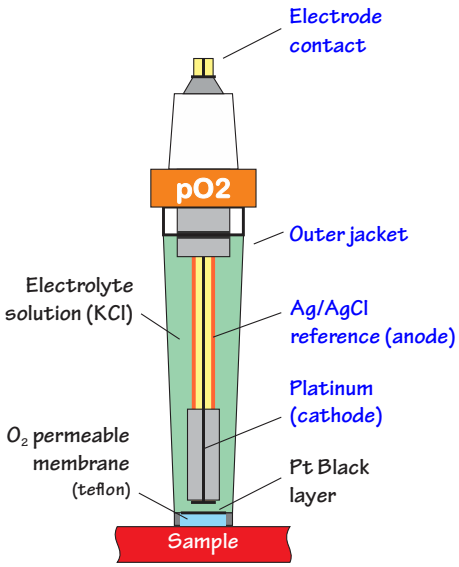
$$\uparrow I \propto \uparrow P_{O2}$$



- the current flow (pico-A) being in direct proportion to the consumption of oxygen
- the platinum electrode cannot be inserted directly into the blood stream as protein deposits form an affect its accuracy



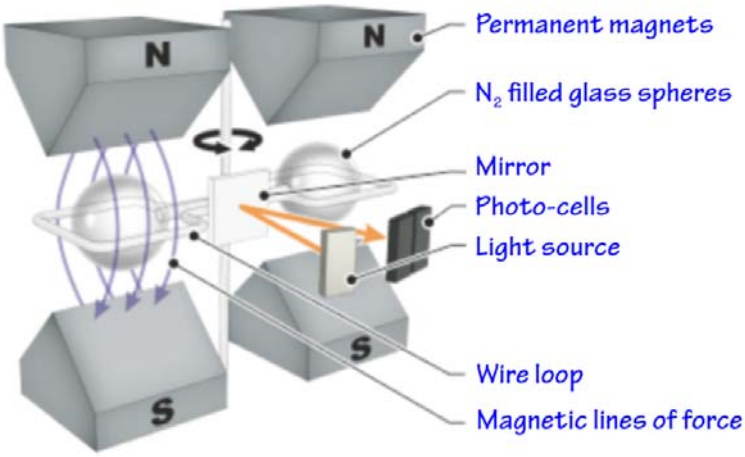
- regular calibration:
    1. electrode ***sensitivity*** established by:
        i. gas 1 19.76% O_2
        ii. gas 2 0% O_2
    2. ***zero-point*** (gas2) and ***drift***
        - 40s required for O_2 washout



- not all Ag^+ ions are removed from solution
  → deposits on cathode (require periodical removal)
-

■ *Paramagnetic Oxygen Analysis*

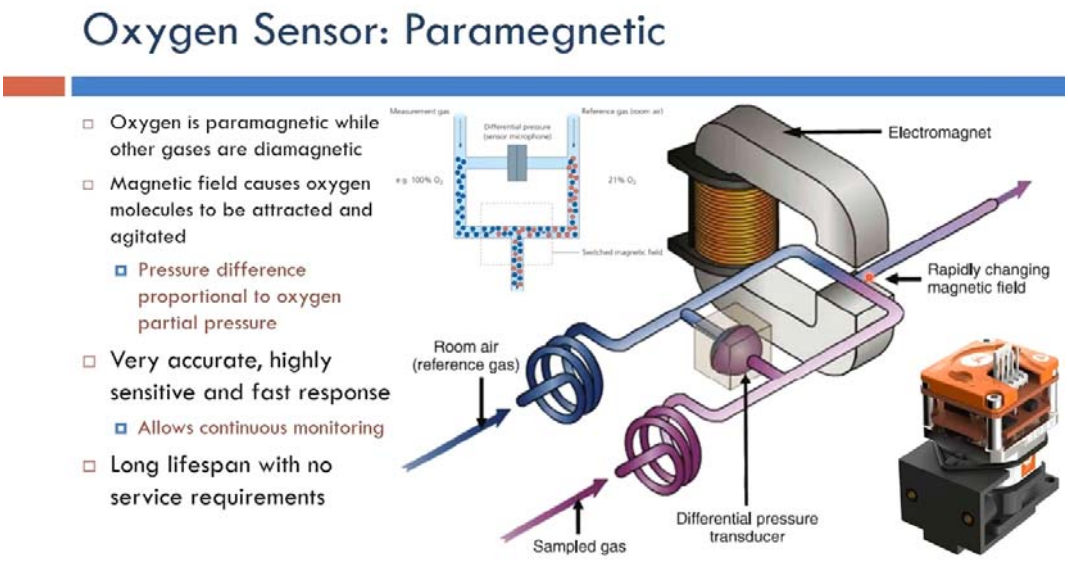
- O_2 is paramagnetic and is therefore attracted into a magnetic field
- this is due to the unpaired outer shell electrons of the oxygen molecule
- most other gases, such as N_2, are weakly diamagnetic and are repelled from a magnetic field
- actually measure ***fractional concentration*** of oxygen $\rightarrow$ $PO_2 = P_{bar} \times FO_2$
- early systems used deflection of nitrogen containing glass spheres in a dumbbell arrangement:



- originally described by Pauling (J Am Chern Soc. **1946**; 68: 795-8)
- these indicate either by direct rotation of a pointer or deflection of light, or may be arranged in a null deflection system
- require calibration before use with 100% N_2 and 100% O_2
- the presence of water vapour biases the result, therefore gases should be dried through silica gel before analysis

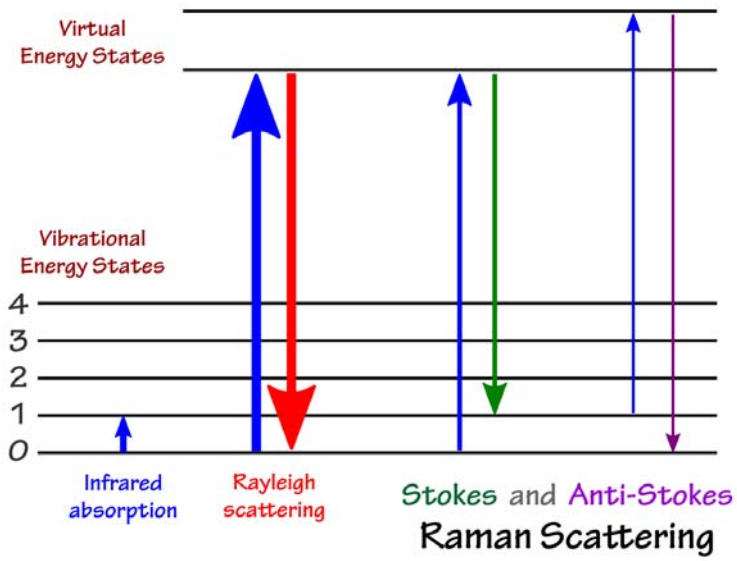
■ *Hummel Cell*

- magneto-pneumatic/acoustic method, described by Hummel (Chemie-Ing-Technol. **1968**; 40:947-51)
- cell where 2 gases were mixed in a homogenous alternating magnetic field
- measured pressure difference in the 2 conduits upstream of the cell – ***pneumatic bridge configuration***
- signal amplitude directly proportional to the difference in oxygen partial pressure ($\propto FO_2$)

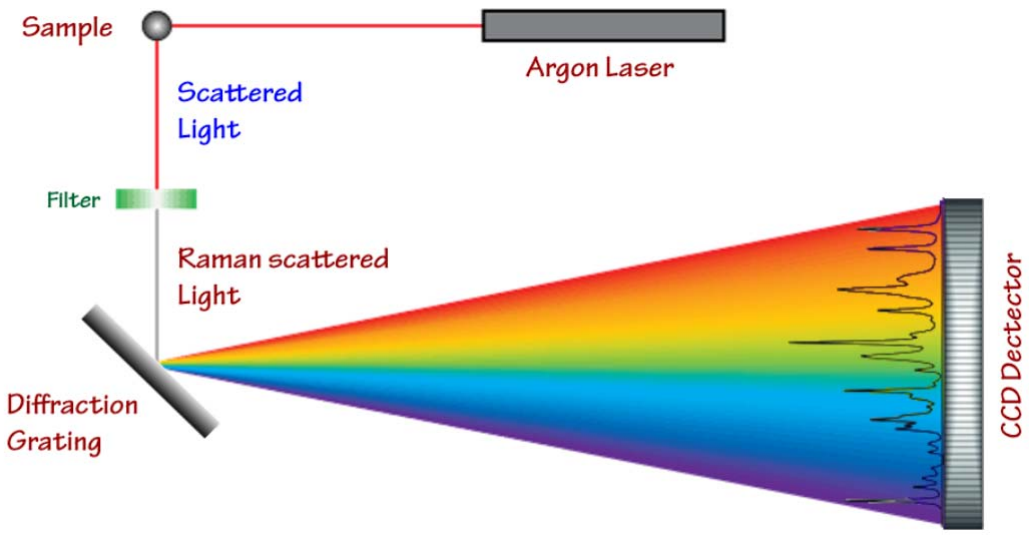


■ *Raman Spectrography*

- Sir C.V. Raman (1930 Nobel prize: 1st non-white!)
- ***Raman scattering*** occurs with high intensity argon laser light
- absorbed energy produces unstable energy states (rotational & vibration)



- emitted low energy light, Raman light → measured at 90° to the laser path
- can identify all molecule types in a gas sample → chemical "fingerprints"
- incorporated into Datex (RASCAL™) monitors
- can identify & quantify CO_2 and inhalational agents



■ *Oxygen Fuel Cell*

- amperometric circuit consists of,
  a. ammeter
  b. gold mesh cathode
  c. lead anode
  d. compensating thermistor
  e. electrolyte solution (KCl) and O_2-permeable membrane

- the same reaction, c.f. Clarke, takes place at the ***cathode***,

$$O_2 + 2H_2O + 4e^- \rightarrow 4OH^-$$

- current flow depends upon the uptake of oxygen at the cathode
- the reaction at the ***anode*** is as follows,

$$Pb + 2(OH^-) \rightarrow PbO + H_2O + 2e^-$$

- unlike the Clarke electrode, the fuel cells requires no external power source, acting as an oxygen dependent battery
- like other batteries, the fuel cell will eventually expire
- the output is affected by ***temperature***, as is that of the Clarke electrode, however compensation may be achieved by means of a parallel thermistor
- the typical response time is ≈ 20-30 s

Other O_2 Measurement

▪ *P_{O2} Optode*

- based on the principle of ***photoluminescence quenching***
- when light shines on luminescent material, electrons are excited to higher energy states and on their return emit light at characteristic wavelengths
- this excited electron can also return to its original energy state by interacting with an oxygen molecule, increasing the vibrational and rotational energy of the later
- for such photoluminescent quenching dyes, the amount of oxygen present can be related to the luminescent intensity by the ***Stern-Volmer equation***,

$$I_{P_{O2}} = \frac{I_0}{1+(k.P_{O2})}$$

where, I = the luminescent intensity at a P_{O2}
I_0 = the intensity in the absence of O_2
k = the ***quenching constant*** for the dye

- the advantages of this system are its simplicity and size, which allow intra-arterial insertion and measurement
- pH-sensitive dyes are also available, therefore , a three optode sensor can measure P_{O2}, P_{CO2} and pH simultaneously

▪ *Cytochrome aa3 Saturation Monitoring*

- this enzyme is distal in the cytochrome oxidase chain and contains copper
- when oxidized this enzyme has an absorbance peak ≈ 830 nm in the near IR range
- as this wavelength is absorbed by both Hb & O_2-Hb, simultaneous estimation of these must be carried out and 3 wavelengths must be used
- the device for measuring this, the ***Niros scope*** = near infrared oxygen sufficiency scope
- uses powerful laser diodes with sufficient light intensity to penetrate the skull

Oxygenation

- under normal conditions, the oxygen cascade results in an interstitial P_{O2} between 20-40 mmHg and an intracellular $P_{O2} \approx 20$ mmHg
- mitochondrial enzyme systems are designed to function at a $P_{O2} \approx$ 2-3 mmHg, therefore there is usually an excess of oxygen
- ***hypoxia*** could therefore be defined as a ***mitochondrial*** **$P_{O2} < 3$ mmHg**
- in the classic study of Comroe & Botelho (1947), after 7,204 observations, it was found that trained observers were unable to detect any degree of cyanosis until the arterial $SaO_2 < 85\%$
- for the detection of cyanosis ≈ 5g of reduced Hb must be present
- with a normal haematocrit this corresponds to a **$SaO_2 \approx$ 60-70%**
- in the presence of anaemia, the saturation must be considerably lower

▪ *Arterial Oxygen Content*

Def'n: volume of oxygen, in mL, contained in 100 mL of blood at 1 atmosphere, at 37°C
= volume percent

$$CaO_2 \approx (1.37 \times [Hb] \times SaO_2) + (0.0034 \times P_{aO2})$$
$$\approx 20 \text{ vol\%}$$

- the ideal value for the carriage of oxygen by Hb of **1.39 mL/g** is not reach in vitro due to the presence of dyshaemoglobins
- thus, for the measurement of content three variables must be known, SaO_2, P_{aO2} and [Hb]
- however, SaO_2 is a function of P_{aO2} as expressed by the Hb-O_2 dissociation curve
- three key points on this standard curve are,
  i. 90% → 60 mmHg
  ii. 75% → 40 mmHg
  iii. 50% → 26.2 mmHg = **P_{50}**
- the curve is displaced to the ***right*** by 4 factors,
  1. increasing $[H^+]$ (decreasing pH)
  2. increasing temperature
  3. increasing CO_2
  4. increasing 2,3-DPG
- it is displaced to the ***left*** by Hb-F, metHb and CO-Hb

■ *Oxygen Delivery - Flux*

Def'n: O_2 Flux (mL.O_2/min) = CO (L/min) × CaO_2 (mL.O_2/dL) × 10

- the normal CO is taken from the cardiac index, CI = CO/BSA
  ≈ 3.0-3.4 L/min/m^2
- this gives an average O_2 flux ≈ 640 mL/m^2/min
- the average BSA for a 70 kg male = 1.8 m^2 → CO ≈ 5.75 L/min
  → O_2 flux ≈ 1150 mL/min

- the normal MRO_2 is stable for a given individual at rest and ranges from 115-165 mL/m^2/min
- the mixed venous oxygen roughly reflects global tissue oxygenation
- the normal value corresponds with
  i. $Cv'O_2$ = 12-15 vol.%
  ii. $Pv'O_2$ = 40-46 mmHg
  iii. $Sv'O_2$ = 72-78 %

- however, different vascular beds have different extraction ratios and the mixed venous P_{O2} does not reflect regional ischaemia

■ *Hypoxia*

- hypoxia is defined as inadequate tissue oxygenation
- therefore this may result from either,
  a. ischaemia - inadequate CO
  b. hypoxaemia - decreased CaO_2
     i. hypoxaemic hypoxaemia - decreased P_{aO2} & SaO_2
     ii. anaemic hypoxaemia - decreased [Hb]
     iii. toxic hypoxaemia - decreased SaO_2
     - P_{aO2} & [Hb] normal

Def'n: a pathological abnormality of oxygen delivery or utilization, resulting in inadequate tissue (mitochondrial) oxidative phosphorylation

(personnal definition - MEF)

TEMPERATURE

■ *Heat*

Def'n: a form of energy, being the state of ***thermal agitation*** of the molecules of a substance, which may be transferred by,

- ***conduction*** through a substance
- ***convection*** by a substance, and
- ***radiation*** as electromagnetic waves

■ *Temperature*

Def'n: the physical state of a substance which determines whether or not the substance is in ***thermal equilibrium*** with its surroundings,

heat energy being transferred from a region of higher temperature to a region of lower temperature

• alterations in the temperature of a substance, through the addition or removal of heat energy, also leads to alterations of the physical properties of the substance

• thus, mercury expands when heated and this was used by Fahrenheit to construct the first temperature scale → freezing point (32°F) & boiling point (212°F) of pure water

1. Kelvin
   - SI unit of thermodynamic temperature
   - 1954: equal to **1/273.16** of the absolute temperature of the ***triple point*** of water

     → the temperature at which ice, water and water vapour are all in ***equilibrium***
   - 2019: 1K change of thermodynamic temperature corresponds to a change in the thermal energy, k_BT, of exactly 1.380649×10^{-23} joules

     → where $k_{Boltzmann} = 1.380649\times10^{-23}$ J/K
2. Celsius scale
   - Temperature (K) = Temperature (°C) + 273.15
   - ∴ the triple point of water is **0.01 °C**

■ *Heat Capacity*

Def'n: heat required to raise the temperature of a given object by 1 K (J/kg/K)

- for a human the approximated mean value ≈ 3.5 kJ/kg/K
- thus, the heat capacity for a 70 kg person ≈ 245 kJ

■ *Specific Heat Capacity*

Def'n: the heat required to raise the temperature of 1 kg of a substance 1 K (J/kg/K)

i. water SHC ≈ 4.18 kJ/kg/K or, 1 kcal/kg/K

ii. blood SHC ≈ 3.6 kJ/kg/K

- 2000 mL of blood at 5°C → 35°C,
  would require ≈ 2 kg × 3.6 kJ/kg/°C × (35-5)°C ≈ 216 kJ
- this would result in the person's temperature falling by ≈ 1°C

■ *Specific Heat Capacity: Gases*

- gases have very low SHC's which are usually expressed per unit ***volume*** rather than per kg,
  - Air ≈ 1.01 kJ/kg/K
  - Air ≈ 1.20 J/L/K (ie. ≈ 1/1000th)

- therefore, only very small amounts of heat are gained or lost when the temperature of a small volume of gas is altered
- for an intubated patient with a tracheal temperature of 34°C, a minute ventilation of 7.0 L/min and a room temperature of 20°C, the heat lost from the patient would be,

Heat Loss ≈ 7.0 L/min × 1.2 J/L/°C × 14°C
≈ 118 J/min
≈ 1.96 W

- this is insignificant compared with the basal heat production of 80 W
- however, greater losses are encountered if the air must be humidified due to the latent heat of vaporisation of water

■ *Specific Latent Heat*

Def'n: the heat required to convert 1 kg of a substance from one phase to another at a given temperature = ***latent heat of vapourization***
= ***latent heat of fusion***

- the LHV of water @ 100°C = 2.26 MJ/kg
  @ 37°C = 2.42 MJ/kg
- therefore, the lower the temperature the greater the latent heat required
- as temperature rises, the latent heat falls until ultimately it reaches zero at a point which corresponds with the ***critical temperature***

Measurement - Non-electrical

a. thermal expansion
   i. gas - Bourdon guage (pressure calibrated to δT)
   ii. liquid - mercury, alcohol
   iii. solid - bi-metallic strip

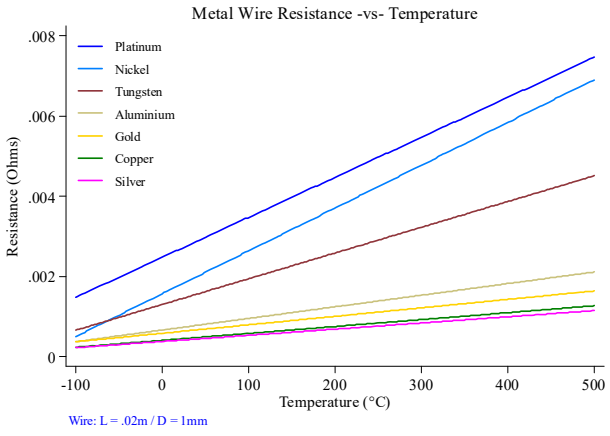
b. mercury thermometers
   - accurate, reliable, cheap
   - readily made in maximum reading form
   - easily made into a thermostat / switch
   - low coefficient of expansion and requires 2-3 mins to reach thermal equilibrium
   - unsuitable for insertion in certain orifices

c. alcohol thermometers
   - cheaper than mercury
   - useful for very low temperatures, mercury → solid at -39°C
   - unsuitable for high temperatures as alcohol boils at 78.5°C
   - expansion also tends to be less linear than mercury

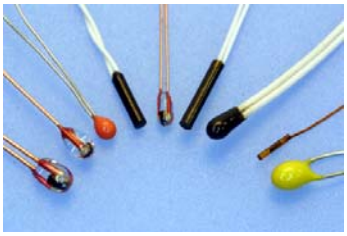
d. bimetallic strips
   - rugged / durable
   - suited to high temperatures and hostile environments

Measurement - Electrical

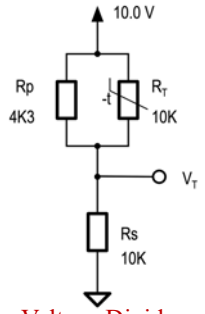
a. resistance thermometer
   - metal wire ≈ linear **↑R ∝ ↑T**
   - platinum - very linear, higher δR/δT, large T-range, inert/stable
   - accuracy improved by incorporation into a Wheatstone bridge circuit



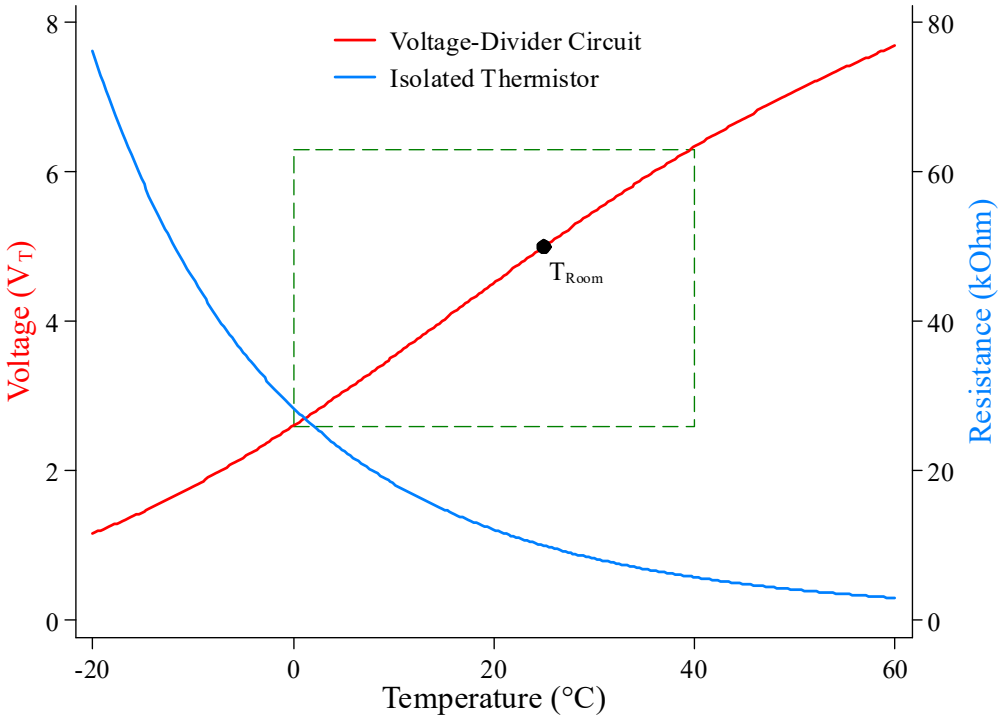
b. thermistor



- made from a small bead of metal oxide
- negative temperature coefficient (**NTC**)
  > positive (PTC)
  → ↓R ***exponentially*** ∝ ↑T
- rapid thermal equilibration
- small and may be used almost anywhere
- devices over range -100 to 300°C
- narrow reference range and require different thermistors for different scales
- accuracy/linearity improved in a Wheatstone bridge or ***voltage-divider*** circuit
- calibration may be changed by exposure to severe temperatures, eg. sterilization



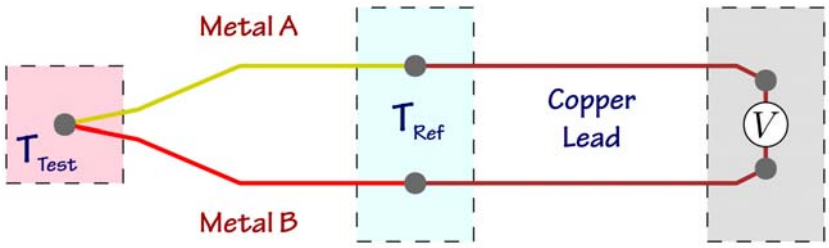
Voltage Divider



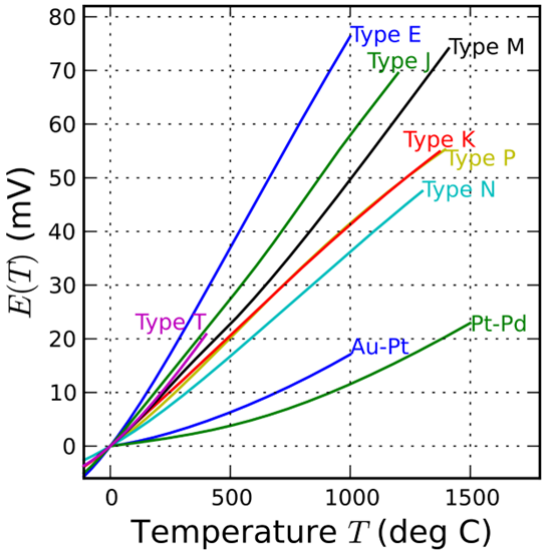
c. thermocouple

- based on the Seebeck effect
- junction of two dissimilar metals → small voltage is produced,

$$\delta V \propto \delta T$$



- constant ***reference temperature*** at the second junction of the electrical circuit
- metals such as copper and constantan (Type T: Cu+Ni)
- can be used for a larger range and higher temperatures c.f. thermisters
- may be made exceeding small and introduced almost anywhere



Body Temperature

- humans, like all mammals and birds, are ***homeothermic*** and control their body temperature within a narrow range = **37 ± 0.5 °C**
- normal circadian rhythm varies temperature ± 0.4 °C, being lowest in the early am. and highest in the evening
- also varied with the menstrual cycle, basal temperature increasing in the second half of the cycle after ovulation
- body is divided into zones,

a.	central core	≈ 37 °C
b.	intermediate zone	
c.	shell (≈ 2.5cm)	≈ 32-53 °C

▪ *Heat Production*

- in the average male under resting conditions ≈ 50 $W.m^{-2}$, or ≈ 80 W total
- increases of the BMR occur after food, with exercise etc.
- also, the BMR rises when there is an increase in the core temperature
- there is no mechanism for a reduction in heat production to compensate for overheating
- increased heat production can be achieved by shivering and voluntary muscular activity

▪ *Heat Loss*

- there are four routes of heat loss from the body,

a.	radiation	≈ 40%	
b.	convection	≈ 30%	
c.	evaporation	≈ 20%	
d.	respiration	≈ 10%	- humidification 8% - heating of air 2%

- ***conduction*** is not an important means of heat loss in humans as gases are poor conductors
- radiation is predominantly in the ***infrared*** spectrum and is determined by the temperature difference between the body and surrounding objects
- the amount of heat loss by evaporation may be increased up to 10x by sweating (LHV)
- all of these mechanisms depend upon the surface area of skin exposed to the environment
- thus, if this area is reduced heat loss is minimized

■ *Latent Heat in Anaesthesia*

- vaporisation of ethyl chloride → skin cooling and local anaesthesia
- vaporisation of volatile anaesthetics results in cooling & lowering of saturated vapour pressure
- compensatory mechanisms are then required to ensure a constant vapour pressure
- rapid emptying of a N_2O cylinder results in cooling and a steady decrease in the cylinder pressure
- this returns to 52 bar if the cylinder is closed and allowed to reheat
- carbon dioxide and cyclopropane are also stored as liquids but the rate of use is too slow to significantly reduce the liquid temperature

- liquid oxygen is stored in containers at about -160°C as its critical temperature is -119°C
- the pressure inside the vessel is set at ≈ 7 bar which is the vapour pressure of oxygen at -160°C
- this is then passed through a superheating coil and regulated to a pipeline pressure of ≈ 4.1 bar
- no refrigeration is needed as the contents are kept cool by the LHV of the oxygen
- if no oxygen is used the temperature and pressure rise above the setting of a safety valve, oxygen is then blown off, cooling the remaining contents
- if the usage rate is greater than the rate of vaporisation, a low pressure valve allows liquid oxygen to flow directly into the superheating coil, increasing the rate of vaporisation

■ *Heat Lost From The Patient*

- for a person breathing dry gas at a minute ventilation of 7 L/min with an upper airway humidity of 34 mg/l, then

Total water vapourized	= 7.0 L/min x 34 mg = 0.238 g/min
Total LHV required	= 2.42 MJ/kg x 0.000238 kg/min = 576 J/min ≈ 9.6 W

- therefore,
  a. heat loss from respiration ≈ 11.6 W (≈ 15% basal heat production)
  b. losses from humidification ≈ 5x losses heating the air

HUMIDFICATION

■ *Absolute Humidity*

Def'n: the ***mass*** of water vapour (g) present in a given volume of air (m^3), numerically = mg/L

fully saturated air	20°C contains	≈ 17 mg/L	≈ 17.6 mmHg
	37°C contains	≈ 44 mg/L	≈ **47.1 mmHg**

■ *Relative Humidity*

Def'n: the ***ratio* (%)** of the mass of water vapour in a given volume of air to the mass required to fully saturate that volume of air at a given temperature

as ***mass*** is directly proportional to the number of moles present, then by the ideal gas equation,

$$\text{Relative humidity} = \frac{\text{actual vapour pressure}}{\text{saturated vapour pressure}}$$

Measurement of Humidity

1. Hair Hygrometer
   - hair elongates as the humidity rises
   - simple and cheap
   - only really accurate over the range 30-90%

2. Wet & Dry Bulb Hygrometer
   - the temperature of the wet bulb is reduced due to evaporation
   - the lower the humidity the greater the evaporative cooling and the greater the temperature difference $\rightarrow$ tables δT -vs- % humidity
   - air must be flowing over the wet bulb to prevent a local rise in the humidity

3. Regnault's Hygrometer
   - uses the principle that condensation occurs when the air is fully saturated at a given temperature = the ***dew point***
   - air is blown through a silver test tube containing ether, reducing the temperature by evaporation
   - the dew point is noted and from tables both the relative and absolute humidity can be established,

$$\text{Relative humidity} = \frac{\text{SVP at dew point}}{\text{SVP at ambient temp.}}$$

4. Other Methods
    i. electrical transducers - both resistance & capacitance
    ii. mass spectrometry
    iii. UV absorption spectroscopy

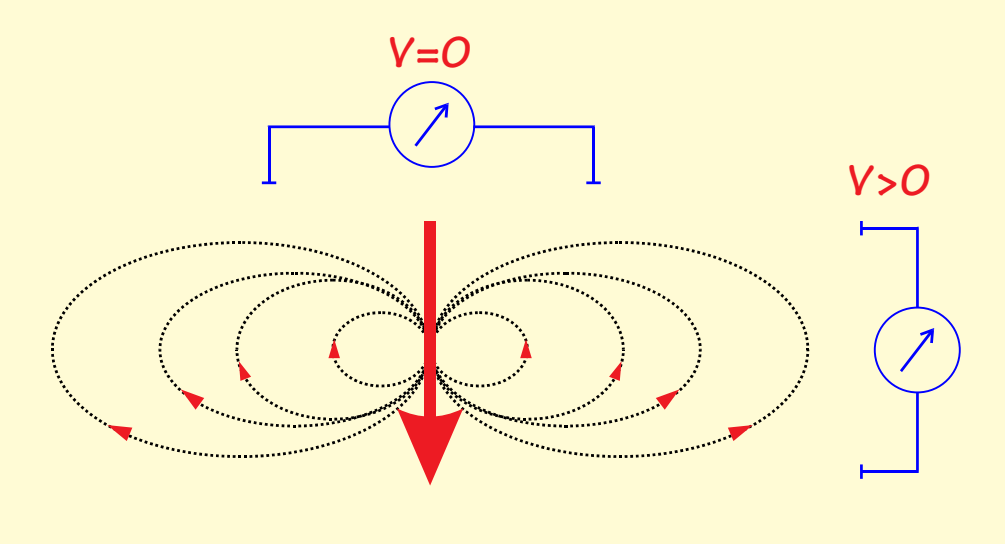
Types of Nebulizers

a. cold water bubble through
b. condenser
c. hot water bath
d. heated Bernoulli nebulizer and anvil
e. ultrasonic nebulizer

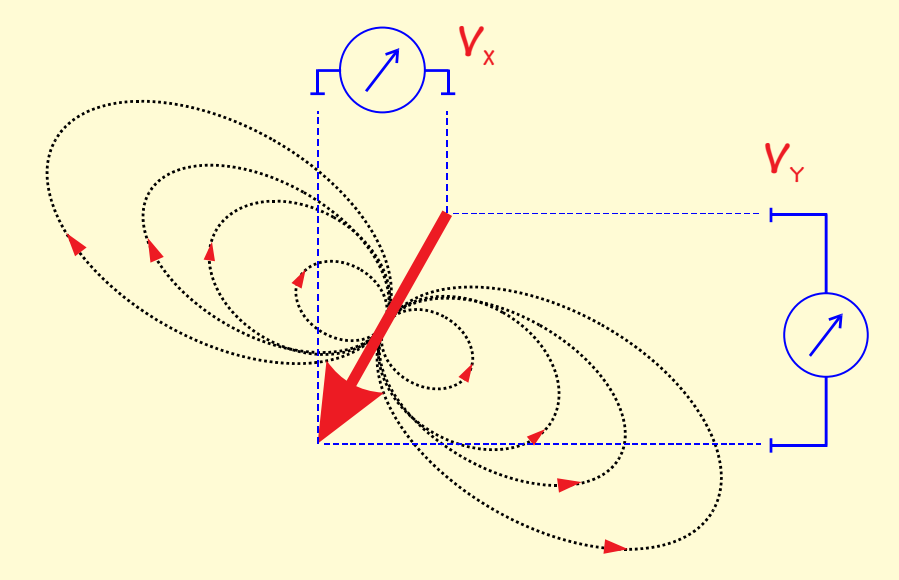
NB: * these are in order of increasing efficiency

Electrocardiography (ECG)

- any net current vector, of size and direction, will generate an electrical field



- this field will be seen as a potential (voltage) difference, measurable at a distance, in parallel with the direction of the current vector
- when the net vector is,
    - perpendicular → measured field will be equal on both sides, and V=0
    - at any angle → the parallel axis components will register, viz,

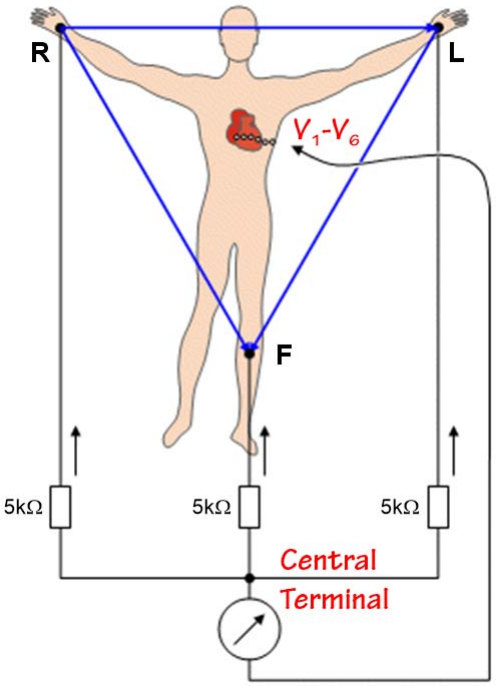


- during the cardiac cycle, multiple cells in the heart are depolarising/repolarising at any instant, acting as a single/summed current vector with net amplitude and direction
- the body acts as a "volume conductor" and these field potentials **≈ 5-10mV** at the surface

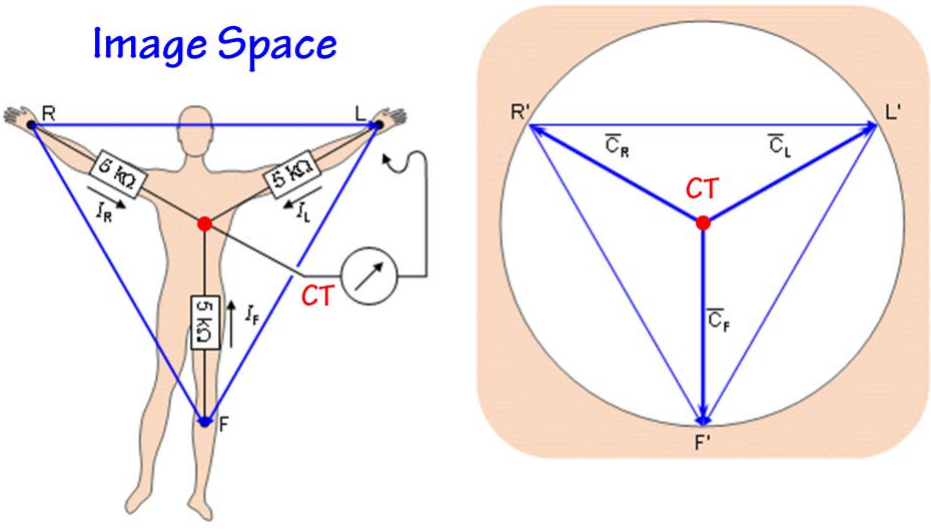
■ *Standard 12-Lead ECG*

- based on ***Einthoven's Triangle***
  1. Bipolar limb leads → I, II, III
  2. 'Unipolar' limb leads → aVR, aVL, aVF
  3. 'Unipolar' chest leads → $V_1 \ldots V_6$

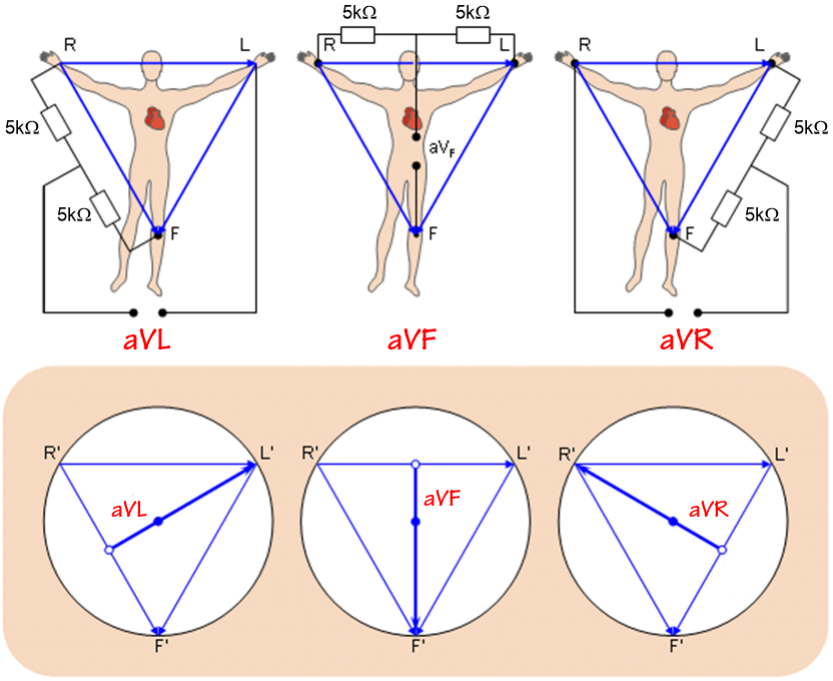
- the ***unipolar leads*** are not truly unipolar, rather the reference point is a summation/calculation from other leads:
  1. Wilson's ***central terminal*** → summation of RA, LA, LL
     - references the centre-point between these 3 points in space
       → mid-thoracic cavity, behind the heart
     - "joining" RA+LA+LL would abolish potentials for leads I, II & III
     - ∴ isolation resistors provide electrical summation whilst preserving differential voltages
     - given surface V ≈ 5-10mV, current from RA→LA ≈ 0.5-1.0 μA (5+5kΩ)
     - given voltages are rapidly changing, this current flow is trivial



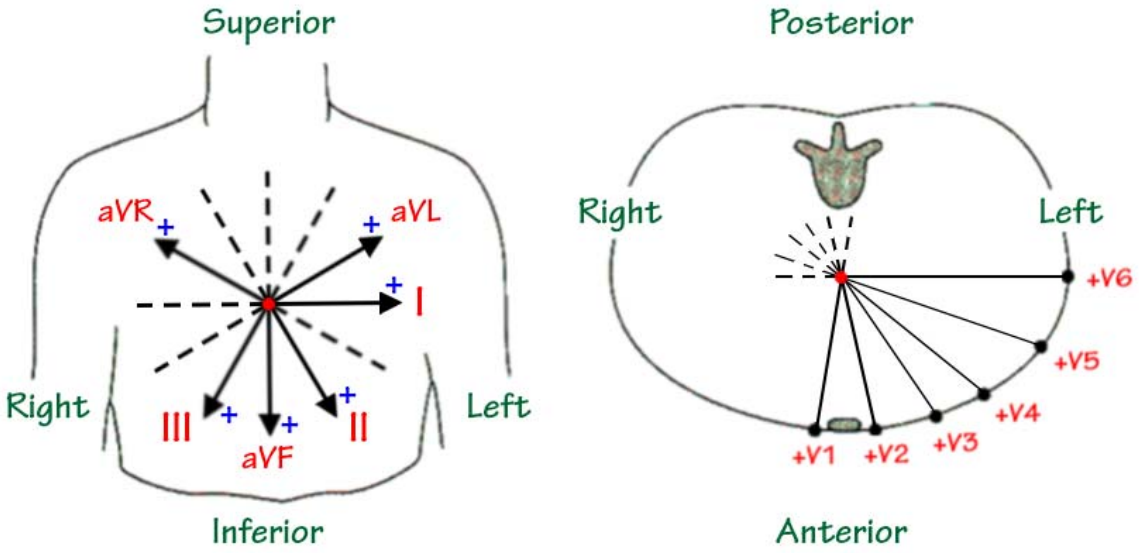
  2. ***augmented*** limb leads
     - limb vectors lie in the direction of the CT → RA, LA, LL



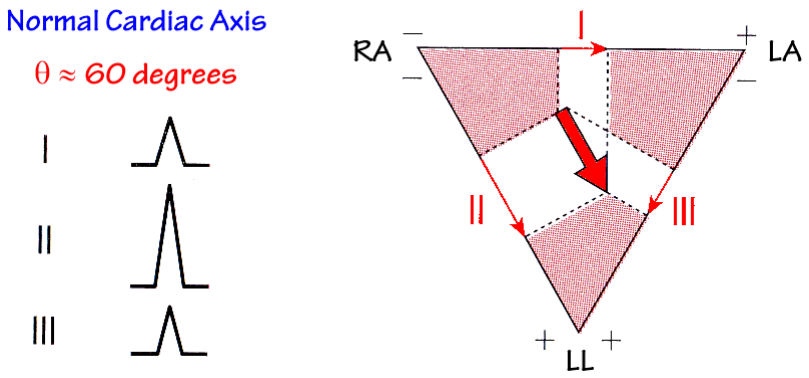
- however, these vectors are identical (but smaller) than those from the middle of the opposing triangle base
- measuring the vector over a greater refrence distance increases signal strength
- this was more important when first established due to very small voltages and electrical interference
- modern-day amplifiers can easily cope with 5-10mV, however, augmentation still improves signal-noise ratio (SNR)



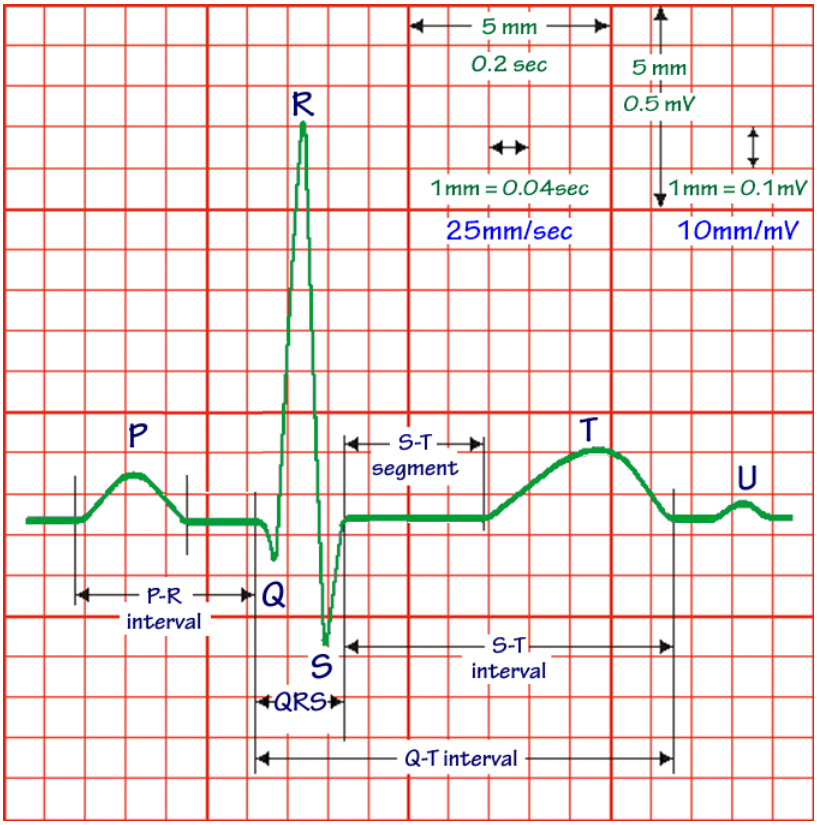
- the complete 12 lead ECG - ***vector image space*** as viewed from the CT



- the overall ***mean cardiac axis*** is given by the dominant vector components in I, II & III

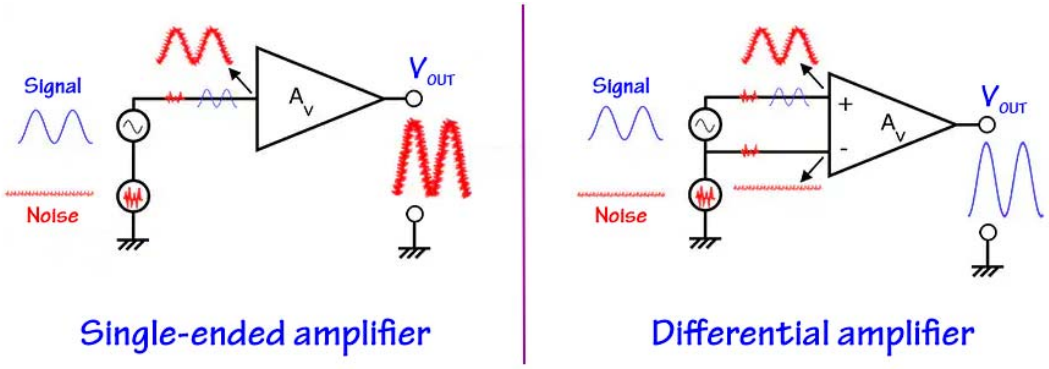


- the sequence of cardiac depolarisation/repolarisation captured in PQRST(u) as "seen" from multiple vector windows → 12 standard leads (II shown below):



■ *Signal Optimisation*

1. machine calibration → timing (mm/sec) & amplitude (δV/mm)
2. correct lead placement
3. minimisation of skin resistance
   - gel electrodes
   - removal of skin oil & hair
4. amplification
   - high gain ∝ small signal (5-10mV)
   - high input impedance
   - high common mode (noise) rejection ratio → differential amplification



5. frequency response & filtering
   - 0.5-40 Hz → ECG waveform analysis
   - 300-1.0 kHz → pacemaker detection
6. e limination of interference / noise
   - mains = 50Hz
   - muscle – large degree of overlap with ECG signal *f*

ENERGY & POWER

Def'n: ***work*** is done when the point of application of a force moves in the direction of that force;

one joule of work is done when a force of ***1 newton*** moves its point of application ***1 metre*** in the direction of the force

• as most work in the body is performed by muscular contraction, the amount of work is the product of the distance of shortening and the mean force exerted

• for fluid flow this may be converted to pressure and volume, viz.

$$W = F \cdot \vec{s}$$

but $P = F/A \rightarrow F = P.A$
and $V = A.s \rightarrow s = V/A$
thus, $W = P.A \times V/A$

or $\mathbf{W = P.V}$

Work of Breathing

• thus, for respiration the work performed is given by the area of a pressure-volume loop

• ie., the cumulative product of pressure-volume of air moved each instant,

$$W = \frac{\delta P.\delta V}{\delta t}$$

• this is required to overcome both ***elastic*** and ***non-elastic*** resistance to breathing,

a. elastic resistance ≈ 65%

b. non-elastic resistance ≈ 35% → 80% airway
20% viscous

• as airway resistance, or inspiratory flow rate increased, so would δP_{IP}, effectively sloping curve to right increasing total and viscous work

1. as respiratory ***frequency*** increases → flow rates & viscous drag increase
2. as ***tidal volume*** increases → elastic work area increases

NB: therefore, patients with stiff lungs → small shallow breaths
patients with airways obstruction → long deep breaths

as both of these patterns tend to ***decrease*** the work of breathing

Metabolic Work of Breathing (O_2 cost of breathing)

- expressed as mL of O_2 (additional O_2 consumption)/L ventilation
- this is low during quiet breathing
- increases with increasing ventilation, especially with pulmonary disease

O_2 cost of quiet breathing	≈ 0.5 to 1.0 mL.O_2/L ventilation
or,	≈ 1-2% of basal MRO_2 (250 mL/min)

■ *Mechanical efficiency*

$$= \frac{\text{useful work}}{\text{total energy expended } (O_2 \text{ used})} \times 100$$

$$\sim \quad 5 \text{ to } 10\%$$

■ *Power*

- is the ***rate of work***, measured in watts, 1 watt being 1 joule per second,

$$W = J.s^{-1}$$

- the power requirement of breathing depends upon the type of flow
- as work = P.V, so for
  a. laminar flow, where $P \propto V$, then power $\propto V^2$
  b. turbulent flow, where $P \propto V^2$, then power $\propto V^3$

- therefore, the power dissipation as fluid flows through a tube is proportional to the square and cube of the flow rate
- this does not allow for the ***kinetic*** component of fluid flow, however, for most physiological examples the kinetic energy component is negligible

Work of Myocardial Contraction

- similarly this is given by the area of a pressure-volume loop
- thus, work approximates 16 kPa (120 mmHg) x 60 mL,

$$\begin{aligned}\text{Work done} &\approx (16 \times 10^{3})\ \text{Pa} \times (60 \times 10^{-6})\ \text{m}^3 \\ &= 0.960\ \text{J} \\ &= \mathbf{960\ mJ}\end{aligned}$$

- therefore, each contraction requires just under 1 J of work

NB: if the HR = 60, the the power output of the LV = 1 J/s = **1 W**

- this can also be calculated from the mean pressure (12 kPa) and flow,

$$\begin{aligned}E' &= P \times V' \\ &\approx (12 \times 10^{3})\ \text{Pa} \times (5 \times 10^{-3}/60)\ \text{m}^3/\text{s} \\ &\approx 1\ \text{W}\end{aligned}$$

- for the RV this would be,

$$\begin{aligned}E' &\approx (2.4 \times 10^{3}) \times (5 \times 10^{-3}/60) \\ &\approx 0.2\ \text{W}\end{aligned}$$

- thus, the total power of the heart ≈ 1.2 W
- given the average efficiency of the heart ≈ 15%, then the total energy requirement of the heart would be 8W
- this approximates 10% of the basal MRO_2, which ≈ 80W
- energy is also required to provide the ***kinetic energy*** of flow, however this is small at rest
- as power is the ***product*** of pressure and flow, increases in either the mean arterial pressure or the cardiac output will significantly raise the myocardial MRO_2

GAS CHROMATOGRAPHY

Def'n: now used as a general term for analytical procedures that separate a mixture into its components as the mixture passes through a column

- clinical uses
    a. volatile anaesthetic agents
    b. barbiturates
    c. benzodiazepines
    d. phenothiazines
    e. steroids
    f. catecholamines

- the system has a stationary phase and a mobile phase
- for gaseous mixtures, the stationary phase of the column is frequently a material such as fine silica-alumina coated with polyethylene glycol or silicone oil
- through this column a flow of carrier gas is passed, such as argon or helium
- sample gases are then entered into the stream, and the speed with which they pass through the column is determined by their differential solubility between the two phases
- solubility is temperature dependent $\rightarrow$ apparatus is maintained at a constant temperature
- this system is often termed gas-liquid chromatography
- as the gases leave the column they pass through some form of detector, e.g.
    a. flame ionization detector - organic vapour
    b. thermal conductivity detector - inorganic vapour
    c. electron capture detector - halogenated vapours

- in a flame ionization detector, the gas is introduced into a hydrogen/air flame
- as the constituents of flames are ionized particles, the resistance of the flame will decrease in the presence of organic gas vapour
- if a constant potential (150V) is generated across the flame, then the current flow will show peaks as the individual components of the gas mixture enter the flame
- the thermal conductivity detector, also called a ***katharometer***, has a heated electrical resistance wire in the main stream of the gas flow
- as different gases have different thermal conductivities, as each component of the sample passes over the wire the temperature will fluctuate
- this system is more suited for the measurement of inorganic gases
- halogenated compounds can be detected with greater sensitivity by an electron capture detector
- a polarizing voltage is applied across an ionizing chamber, in which electrons are released by a radioactive cathode
- halogenated compounds accept electrons and decrease the current flow reaching the anode

NB: ** none of these detectors allows absolute identification of the component gasses, and some knowledge of the substituents is necessary prior to analysis

- the time between entry of the sample and the appearance of the component is the ***retention time***
- most samples will have numerous peaks with varying retention times
- with appropriate calibration the area of a peak can be used to calculate the quantity of the gas present in the mixture
- if the portal of entry of the sample is heated then injected liquids will be vapourized and these can also be analyzed
- it is useful for measuring very low concentrations of either gases or liquids
- however, continuous analysis is not possible and some knowledge of the sample must be available

Ultra-Violet Analysis

- halothane absorbs light in the UV spectrum
- therefore the concentration of halothane may be measured in accordance with Beer's law, as for end-tidal CO_2
- a reference is obtained with a beam splitter and a second chamber
- the sample and reference cells have quartz windows as glass absorbs UV light

Piezoelectric Gas Analysis - "Emma"

- a quartz crystal is coated with oil
- gasses are absorbed into the oil in proportion to their gas:oil partition coefficients and in accordance with Henry's law
- the presence of the gas alters the resonant frequency of the crystal which can be measured electronically
- these analyzers are not agent specific and will respond partially to water vapour

MASS SPECTROMETER

VAPORISERS

- SVPs of the volatile anaesthetics are many fold greater than their respective MAC's

Agent	Sat. Vapour $P_{20°C}$		MAC
Halothane	243 mmHg	32%	0.75 %
Enflurane	175 mmHg	23%	1.68 %
Isoflurane	251 mmHg	33%	1.15 %
Sevoflurane	160 mmHg	21%	1.71%

- reduction in the vapour pressure is achieved by dividing the gas flow from the meter into two streams, one bypassing the vapour chamber
- gas can flow through a vapourizer by two means,
    a. plenum vapourizers → gas is driven proximally
    b. draw-over vapourizers → distal "negative" pressure
        - either by the patient's respiratory efforts, or by mechanical means

Boyle's Bottle

- early, simple type of plenum vapourizer
- bypass and vapour streams determined by a rotatory valve
- the degree of saturation of vapour is highly dependent upon the flow rate
- with vaporisation, the temperature and saturated vapour pressure of the bottle fall
- output varies with both temperature and flow rate, making the device unsuited for calibration

▪ *Flow Dependence*

- this is abolished if all vapour passing through the chamber is fully saturated at all flow rates
    → concentration can be adjusted by the splitting ratio, and is independent of flow
- this requires a large surface area in the chamber, which may be achieved by,
    a. wicks → Floutec, Dräger, Abingdon
    b. scintered discs

- the ***splitting ratio*** depends on the relative resistances to flow through the two paths, and thus is affected by,
    a. laminar vs. turbulent flow
    b. physical properties of gases - viscosity and density

- problems are usually worse at low flow rates (<1 L/min)

• calibration will depend upon which carrier gas is used, and this should be undertaken to represent the clinical conditions of use

■ *Temperature Control*

• the glass used in the walls of Boyle's bottle is a poor conductor and little heat exchange occurs with the surroundings
• most modern vapourizers use metal cases with good thermal conductivity
• also, a heat reservoir of either metal or water may be used to delay temperature fluctuations
• these changes are however not eliminated and some form of compensation is required,

a. temperature measurement and concentration scales
b. temperature measurement and manual adjustment
c. temperature controlled valves
    i. bimetallic strips (Fluotec, PAC)
    ii. bellows valve (EMO, Abingdon, Ohio)
    iii. metallic rod valve
d. direct addition of the volatile liquid to the gas stream

• due to the high saturated pressures of the volatile agents, regular calibration is essential to prevent inadvertent overdosage

■ *IPPV*

• due to an intermittent fall in back pressure from the ventilator, gas from the vapour chamber may expand into the bypass channel, thereby increasing the concentration of agent delivered
• this is more likely to occur when the volume of the chamber is significantly larger than the bypass channel
• this may be solved by,

a. a pressurizing valve, ensuring the vaporiser pressure is always ~ the ventilator pressure
b. the volume of the vapour chamber = the bypass channel
c. increasing the length of the chamber inlet tube so no retrograde flow reaches the bypass channel

■ *Hyperbaric Conditions*

• the saturated vapour pressure is unaffected by the ambient temperature
• thus, for halothane, is still 32 kPa at 200 kPa ambient pressure
• since the splitting ratio is unchanged, the vapourizer will deliver 1/2 of the dialled percentage
• however, as the depth of anaesthesia is dependent upon the partial pressure of the agent, not the percentage, most vapourizers may be used with the usual settings at different ambient pressures

■ *Vapourizer Position*

- should be positioned between the flow meter block and the oxygen emergency flush control
- if there is an emergency gas flow cut-out actuated by failure of the oxygen supply then this should be down-stream of the vapourizer
- the control should be off in the clockwise position and both inlet and outlet to the chamber should be occluded

■ *Draw-Over Vapourizers*

- similar problems exist but in addition the internal resistance of the circuit must be low, so not as to add undue resistance to the patients breathing
- because they do not require gas supplies, they are ideal for "field" work
- the "EMO" is well established and has a bellows thermal compensatory device and a water reservoir for thermal stability
- it is designed for use with ether as this is produces less cardiorespiratory depression than the modern volatile agents
- the sample is passed through a molecular leak into an ionizing chamber
- the ionized particles are then accelerated and focussed into a beam which directed though a strong magnetic field
- depending upon their ***charge/mass ratio***, different molecules describe different arcs of travel
- these separated beams are then detected depending upon their position
- by varying the accelerating voltage, molecules of different masses can be made to describe the same arc $\rightarrow$ one detector
- alternatively, multiple detectors can be used
- an alternative means of manipulating the accelerated beam is the quadrupole
- here, 4 electrically charge rods are positioned around the beam such that only a molecule of a given charge/mass ration will remain undeflected
- some compounds fragment on ionization and analysis of the fragments can allow differentiation between molecules of the same charge/mass ratio
- this occurs with N_2O and CO_2, both of which have a MW = 44, however the nitrous oxide fragments into nitric oxide which allows differentiation