Frequent problems in ECG instrumentation

normal

sudden big shock: eg de-fibrillation of the patient (you apply a large voltage, it's also at the input of ampop which saturates, and slowly recovers with 1°timeconstant which should be kept short as possible)

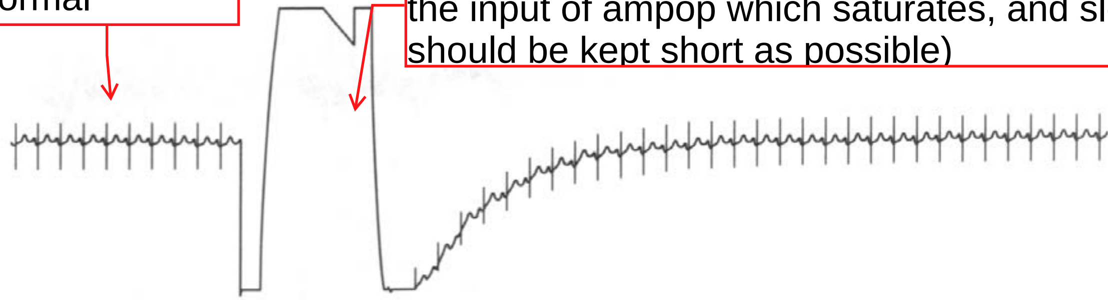


Figure 6.8 Effect of a voltage transient on an ECG recorded on an electrocardiograph in which the transient causes the amplifier to saturate, and a finite period of time is required for the charge to bleed off enough to bring the ECG back into the amplifier's active region of operation. This is followed by a first-order recovery of the system.

saturation and recovery transient of the baseline following the application of a large signal (e.g. defibrillation)

interference, in particular by power line

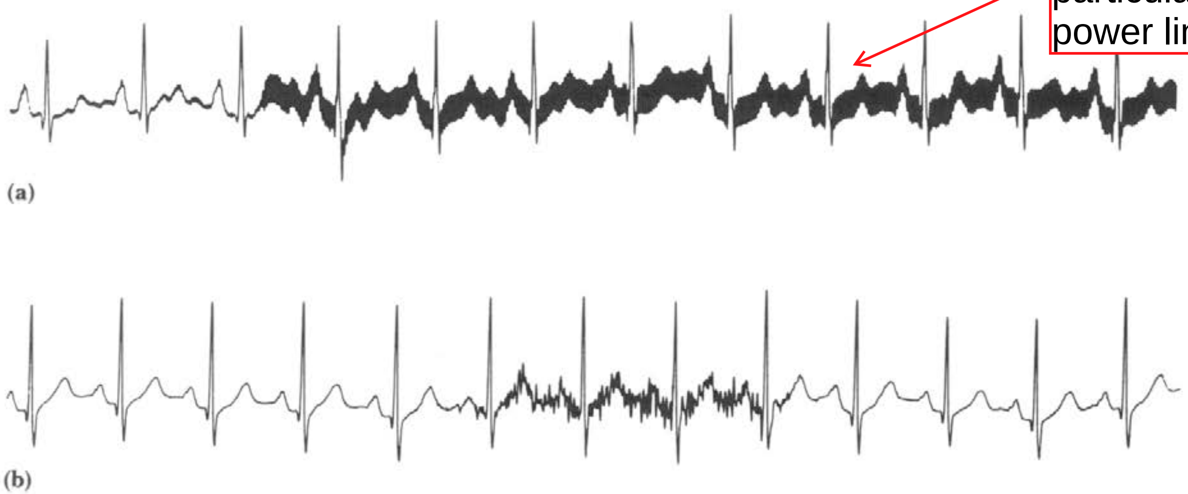


Figure 6.9 (a) 60-Hz power-line interference. (b) Electromyographic interference on the ECG. Severe 60-Hz interference is also shown on the bottom tracing in Figure 4.13.

interferences produced by the 50Hz power line or by other instrumentation

1. E.FIELD IRRADIATION ON CABLES/INSTRUMENT

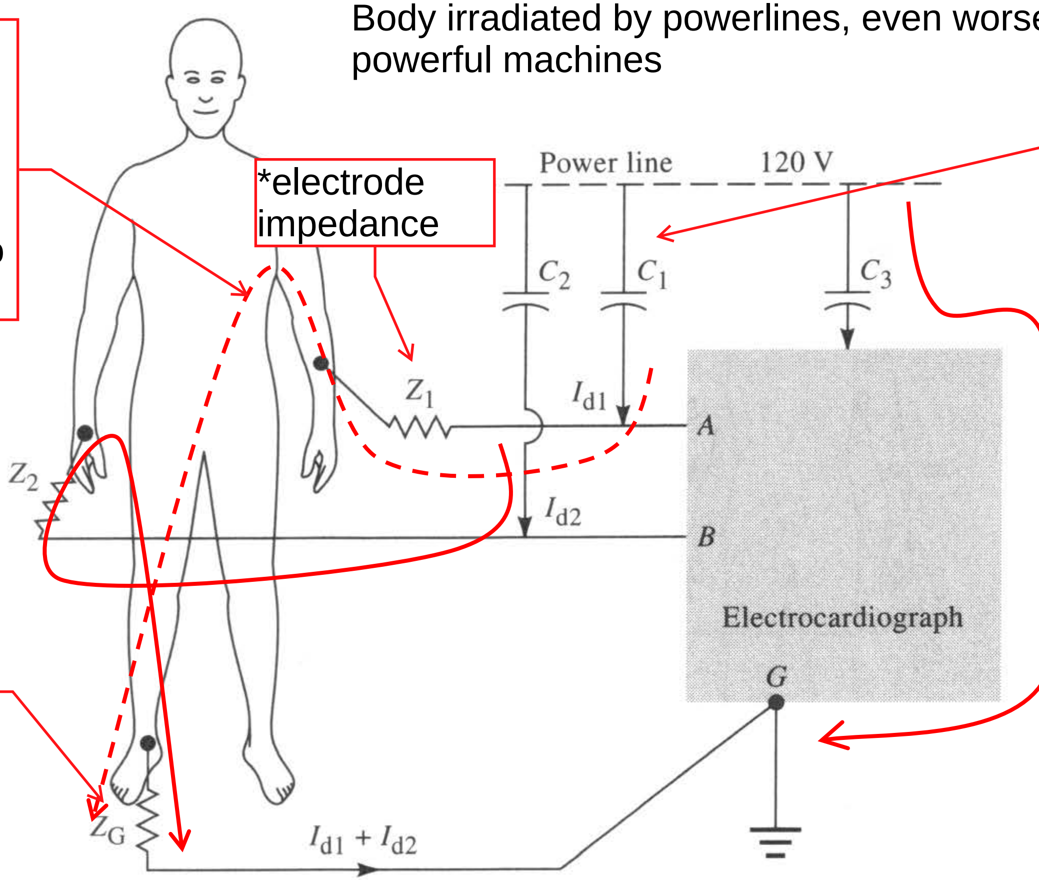
Coupling of the power line (220V, 50Hz) on the ECG

Body irradiated by powerlines, even worse in hospital where there are powerful machines

if you do nothing, the currents Id1 and Id2 (dont flow in the input of ampop, it's high impedance) flow to the patient

*electrode impedance

irradiation effect on the instrument: modeleded by capacitor coupled. Id1 and Id2, on the wires



OK:
Faraday cage

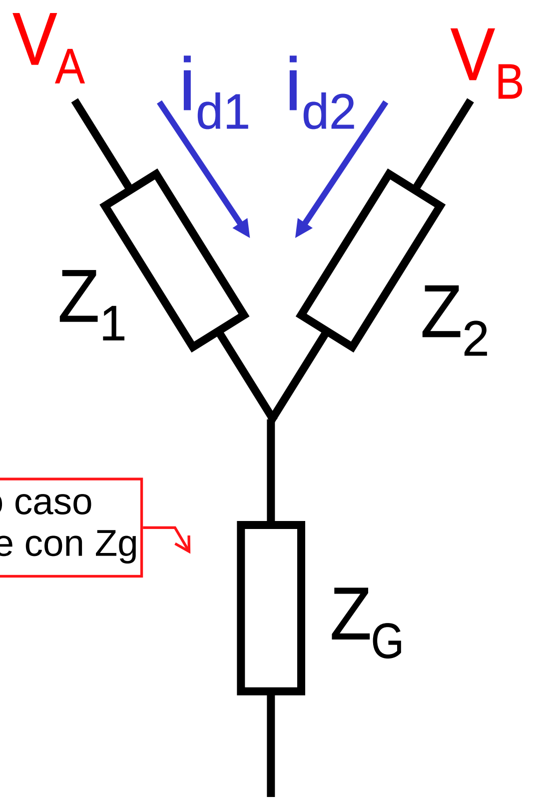
goes to ground with the RL. The patient is connected to gnd wit the Zg so that he's not shorted to ground in case of failure

it's shielded, so usually the current flow to ground through the metallic case

Figure 6.10 A mechanism of electric-field pickup of an electrocardiograph resulting from the power line. Coupling capacitance between the hot side of the power line and lead wires causes current to flow through skin–electrode impedances on its way to ground.

2 effect of irrdation:
1- instrumentation 2- on the patient

E.FIELD => WIRES => CURRENT => VdFake (Z1 != Z2)



in questo caso
partizione con Zg

$$V_A - V_B = i_{d1}Z_1 - i_{d2}Z_2$$

if $i_{d1} \approx i_{d2}$

$$V_A - V_B = i_{d1}(Z_1 - Z_2)$$

a differential signal appears at the amplifier input

e.g. $i_{d1} \approx 6nA$

$Z_1 - Z_2 \approx 20\ k\Omega$

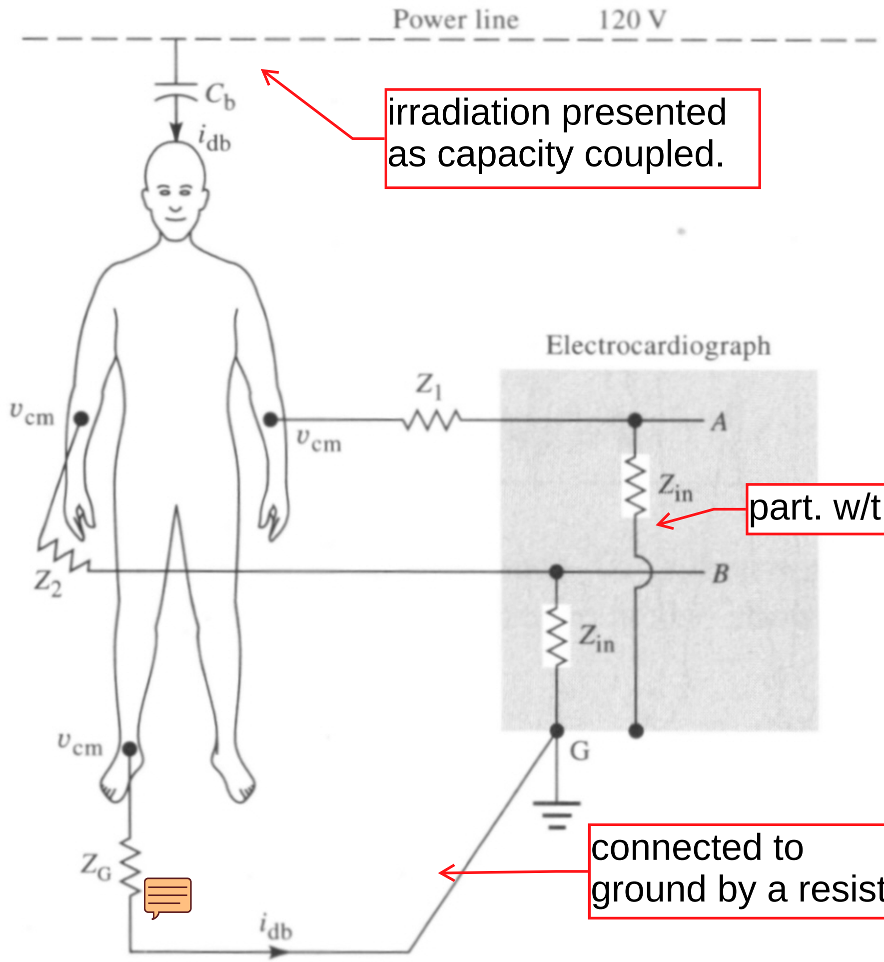
$\Rightarrow\ V_A - V_B = 120\mu V$

hints:

- shield the inputs
- reduce the electrodes impedance

Modeled. [audio]. Problem: if the impedance would be equal no effect, but since we cant be sure that electrode impedance will be equal, the voltage difference is not 0. number for example. 120uV IS NOT small compared with ecg voltage. So it's important to shield and **reduce el. impedance** (you'll reduce also the relative difference)

2. irradiation into the patient



irradiation presented as capacity coupled.

part. w/t Zin

connected to ground by a resistor

Figure 6.11 Current flows from the power line through the bod impedance, thus creating a common-mode voltage everywhere Z_{in} is not only resistive but, as a result of RF bypass capacitors at input, has a reactive component as well.

$$V_{cm} = i_{db} Z_G$$

$$V_A - V_B = V_{cm} \times \left(\frac{Z_{in}}{Z_{in}+Z_1} - \frac{Z_{in}}{Z_{in}+Z_2} \right)$$

approx

$$Z_1, Z_2 << Z_{in}$$

voltage diff due to cm:

$$V_A - V_B = V_{cm} \times$$

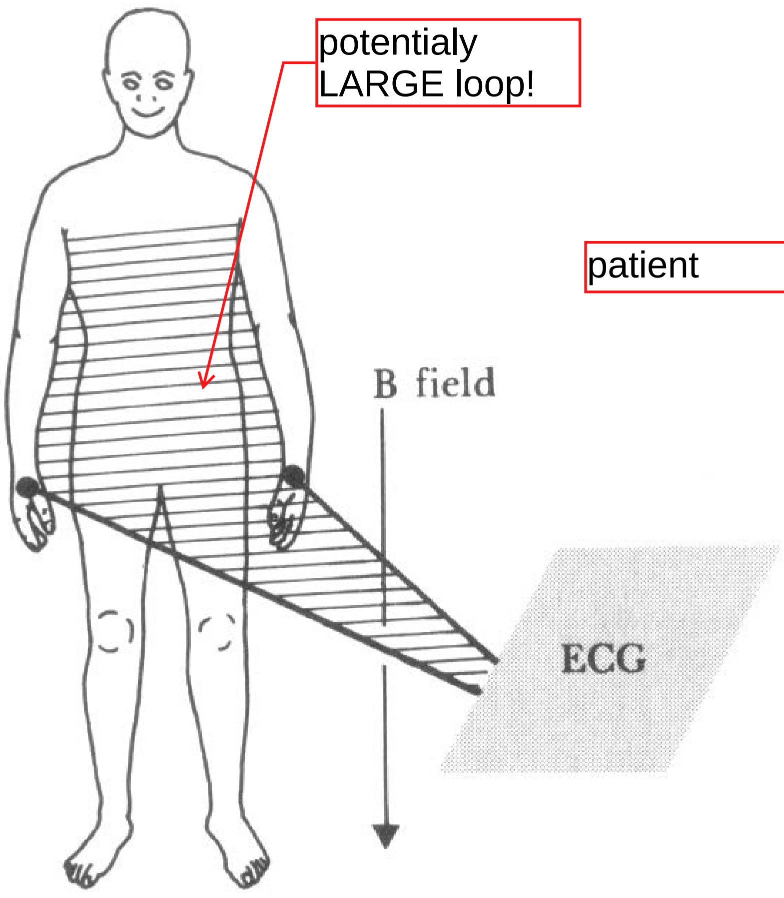
$$\times \frac{Z_2 - Z_1}{Z_{in}}$$

large

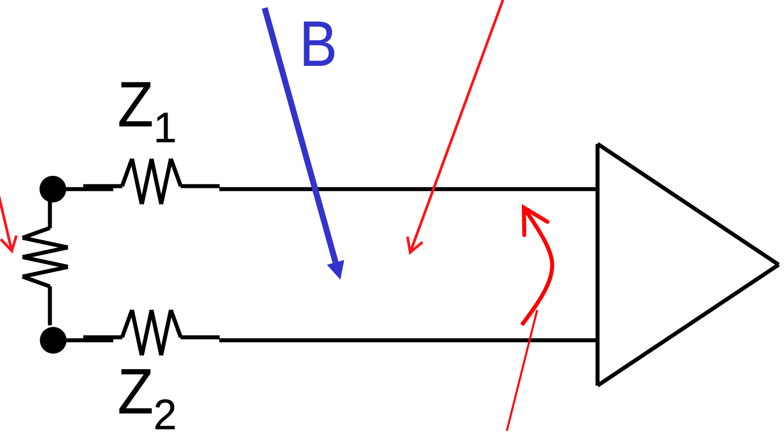
differential signal created by the common mode through unbalance impedances

If we do nothing the current flows though Zg and creates a ddp. This voltage is common all over the body (CM) and can be few volts. This will be at the input of the ampop. You don't have directly this, but its partition with Z1/Zin and Z2/ZIN. With Zin as large as possible they might be negligeble (the partition). If not we have that formula.

Magnetic coupling

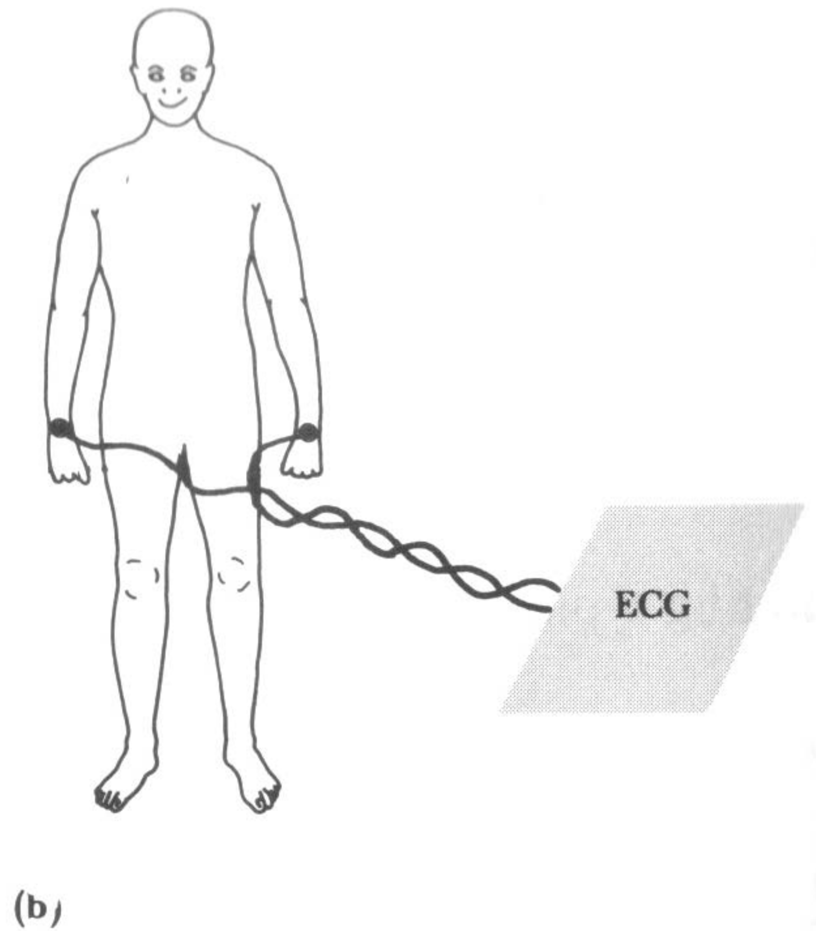


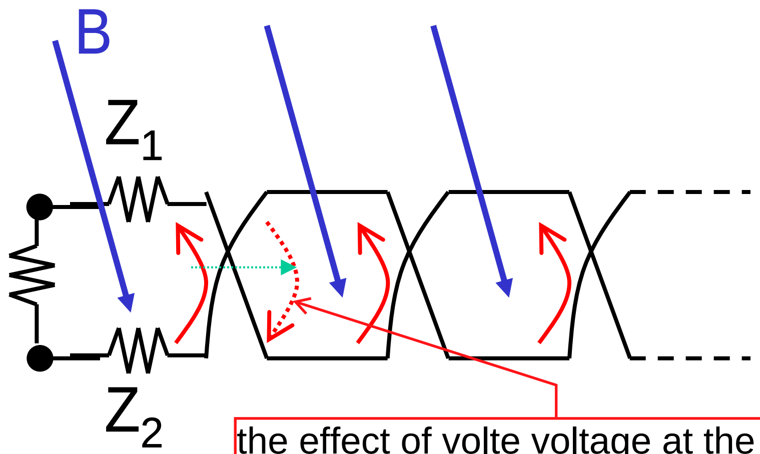
(a)



We also have magnetic field, not only electro-field. A loop is created, which is "conductive", if the magnetic field is variable and is coupled, we have an induced potential difference.

Usual way: **TWISTING PAIRS**





the effect of volte voltage at the second circle FLIPS! So the two voltages ideally compensate each other

With twisted pair cables:

1) whorls are smaller
2) potential differences created in the different whorls tend to compensate each other

We create many small circles that helps. Every two circles we compensate the potential difference created.

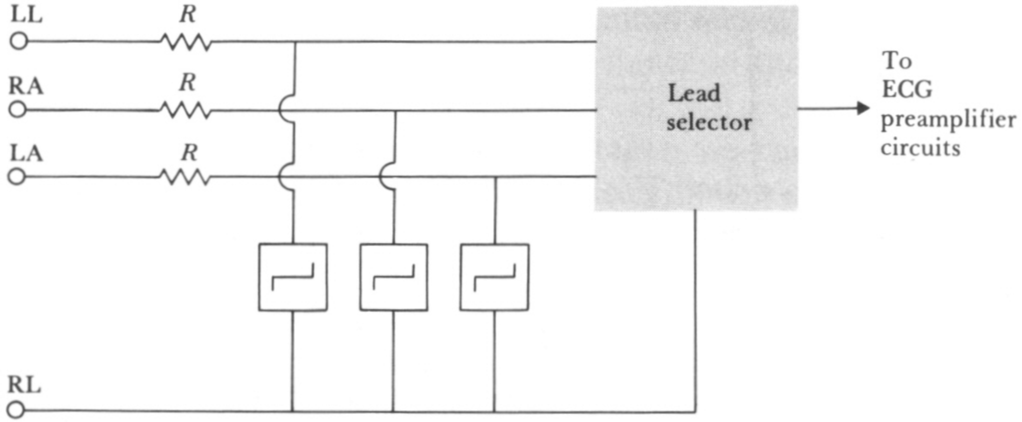


Figure 6.13 A voltage-protection scheme at the input of an electrocardiograph to protect the machine from high-voltage transients. Circuit elements connected across limb leads on left-hand side are voltage-limiting devices.

Components for the over-voltage protection (of the instrument)

input voltage cannot exceed

Flipped diode

Zener back to back

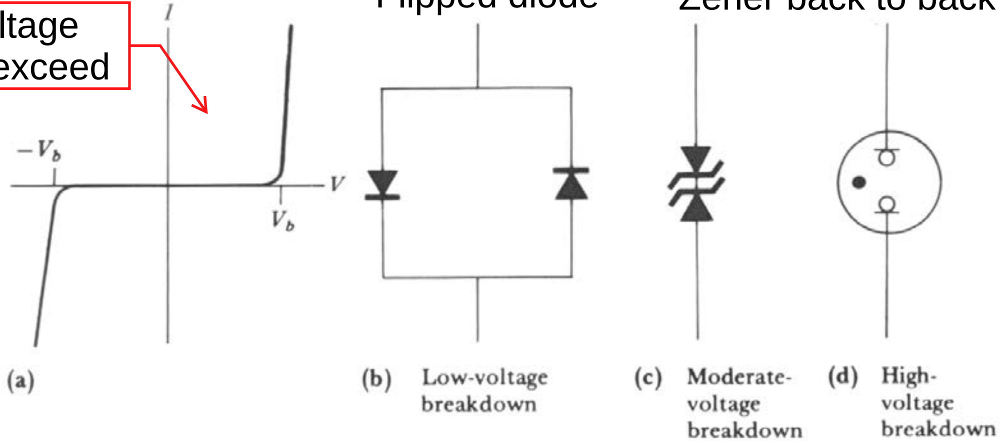


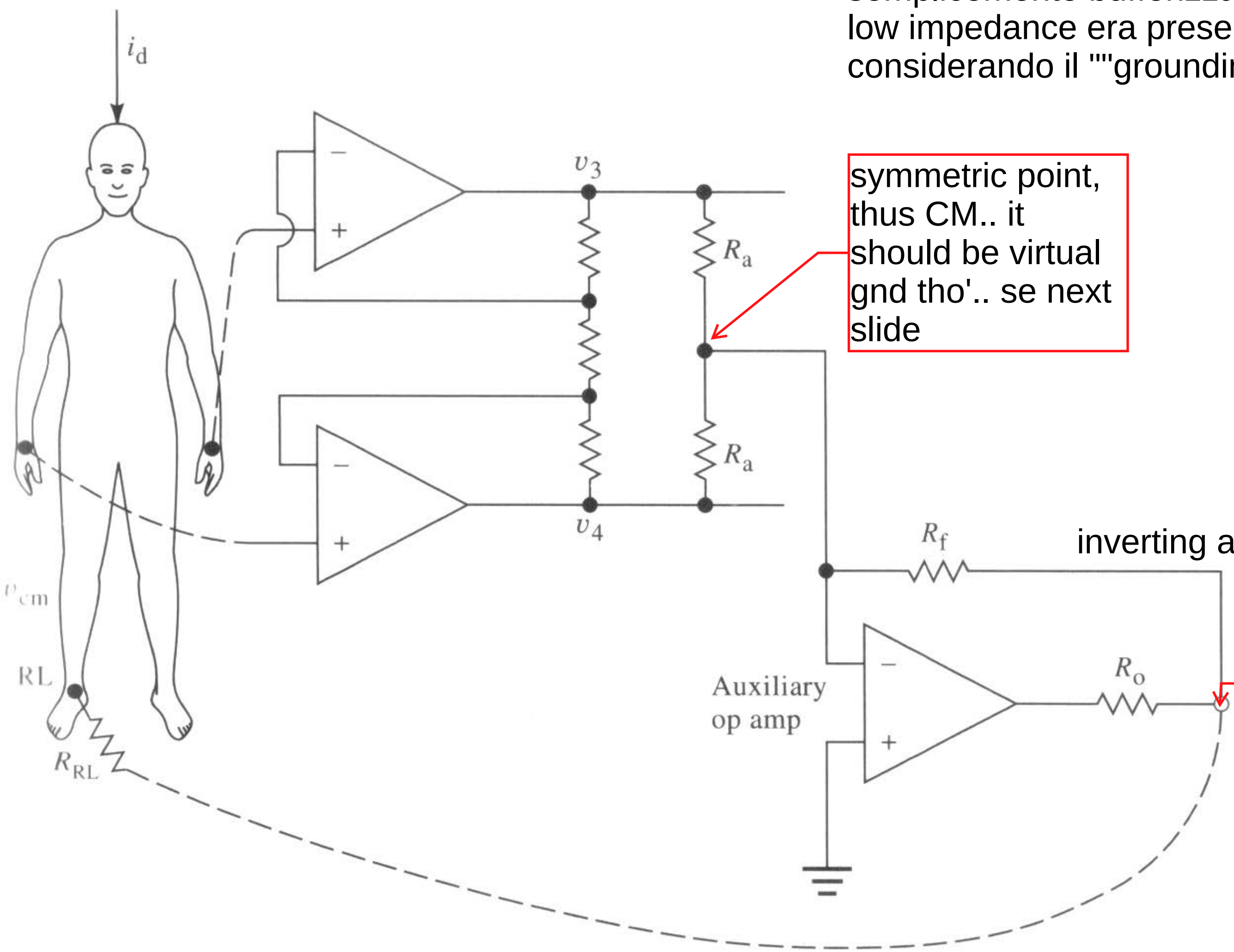
Figure 6.14 **Voltage-limiting devices** (a) Current-voltage characteristics of a voltage-limiting device. (b) Parallel silicon-diode voltage-limiting circuit. (c) Back-to-back silicon Zener-diode voltage-limiting circuit. (d) Gas-discharge tube (neon light) voltge-limiting circuit element.

If the patient experiences defibrillation, if you do nothing it destroys the ampop. We need protection devices. It's just a protection of differential voltage at the input of ampop.

Right-leg driver circuit to reduce the common-mode voltage

Minimize CM

*nello shielding ed uso come ref, veniva semplicemente bufferizzata (con x1). La low impedance era presente anche considerando il ""grounding"" dello shield



symmetric point, thus CM.. it should be virtual gnd tho'.. se next slide

inverting ampop

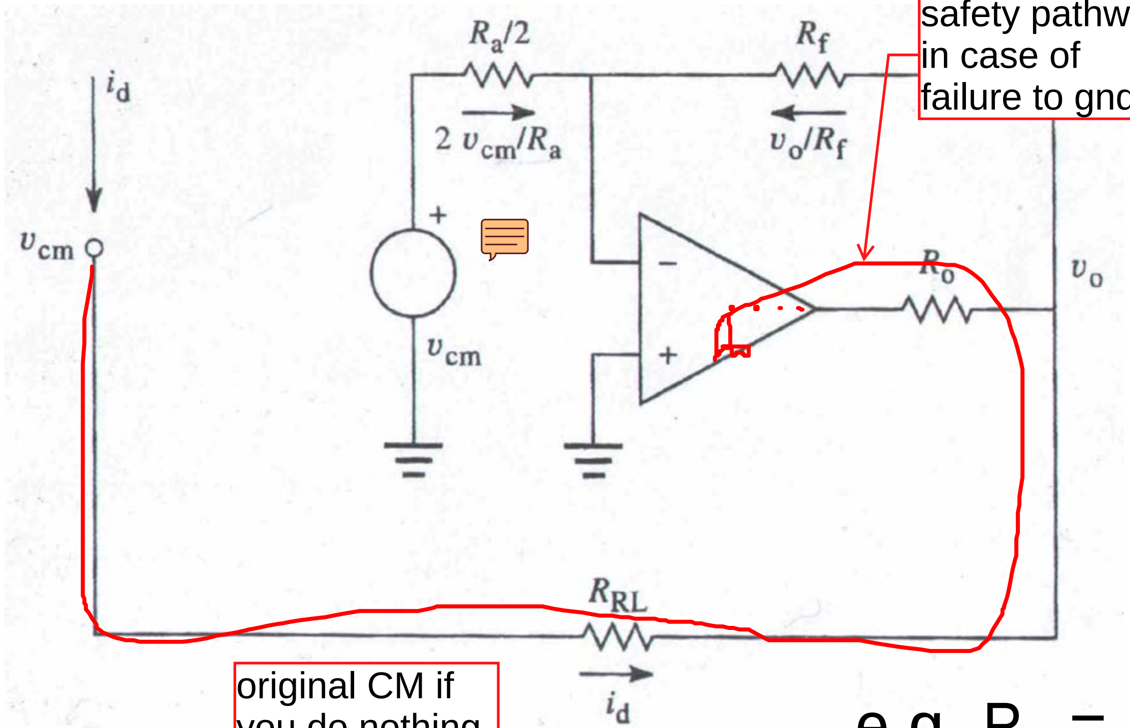
low impedance

CM is a problem, if I do nothing it's id*Rrl. We create a negative loop. Instead of putting Rrl to ground, it's driven by an ampop. It **amplifies** and inverts the CM. Rrl is STILL put at low impedance, it's connected to the output of an ampop, which thanks for the loop effect it's l-impedance.

equivalent thevenin [audio], *con thevenin viene, con sov. effetti no(!?)

explanation of passages

safety pathway
in case of
failure to gnd



$$\frac{2v_{cm}}{R_a} + \frac{v_o}{R_f} = 0$$

$$v_o = -\frac{2R_f}{R_a}v_{cm}$$

vcm-vo=id*Rl

$$v_{cm} = \boxed{R_{RL}i_d} + v_o$$

original CM
*la id è fissa!!

original CM if
you do nothing

$$\boxed{v_{cm} = \frac{R_{RL}i_d}{1 + 2R_f/R_a}}$$

improvement factor
= To loop gain!

e.g. R_a = 25kΩ, R_{RL} = 100kΩ

$$\frac{100\ \mathrm{k\Omega}}{1 + \dfrac{2 \times 5\ \mathrm{M\Omega}}{25\ \mathrm{k\Omega}}} = 249\ \Omega$$

i_d

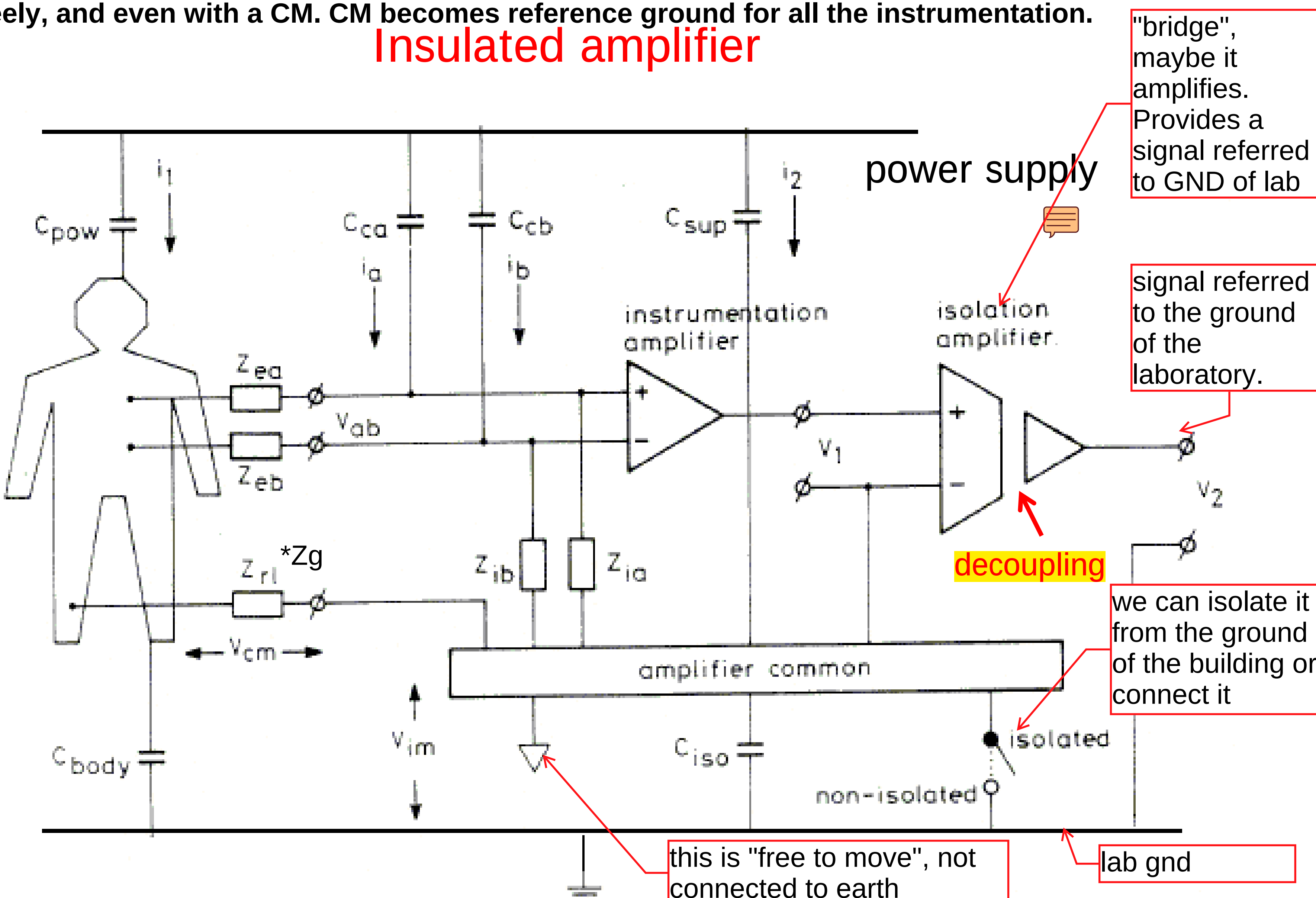
$$v_{cm} = 249\ \Omega \times 0.2\ \mu\mathrm{A} = 50\ \mu\mathrm{V}$$

(ref. 100kΩ x 0.2μA = 20mV)

eg, impedance of 100k is transformed to 249ohm. Transformation of impedance Rrl. Resistor Rout: it's not playing any role, it's output R of ampop, doesnt change the formula because vo is made """0""" impedance, even if we have R0. We put it tho' for SAFETY reasons, it guarantees a resistive path towards ground. If the loop is open, not working, for eg in DEFRIBILLATION: indeed if it's SATURATED not in linear region, IT'S NOT WORKING

2. amplifier common: LOCAL GROUND not connected to laboratory gnd. We leave this common electrode to float freely, and even with a CM. CM becomes reference ground for all the instrumentation.

Insulated amplifier

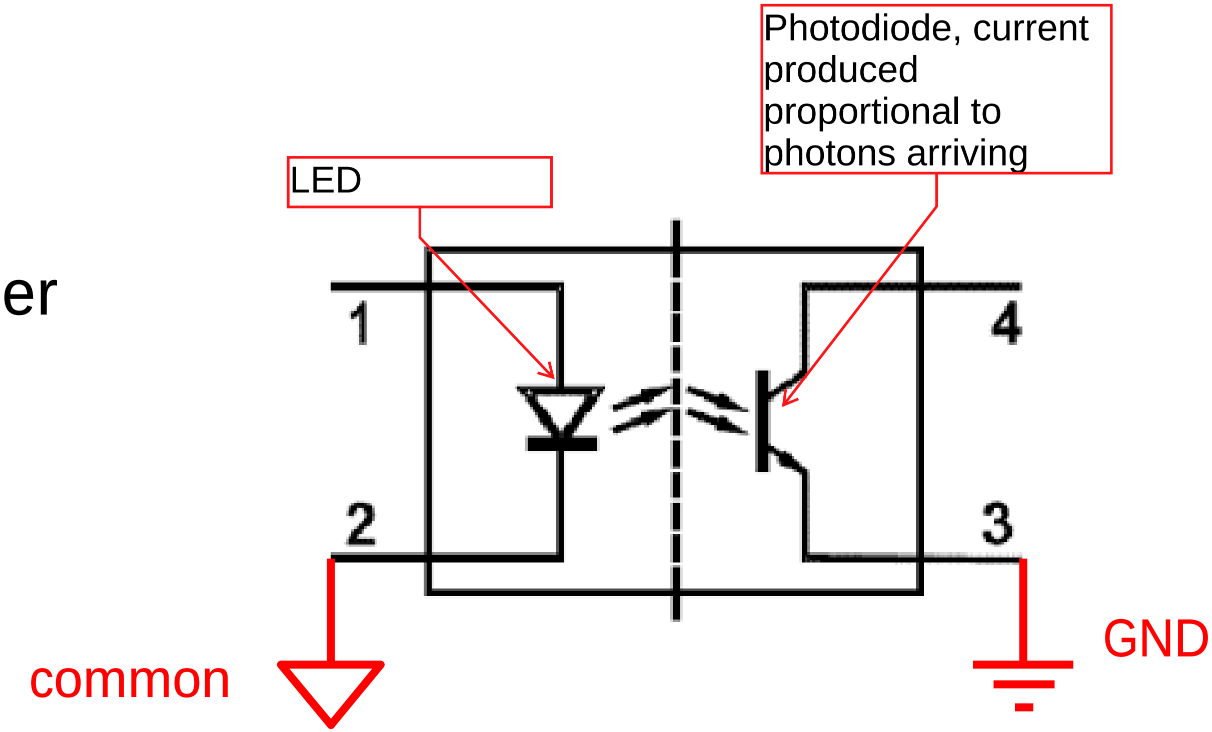


1. We have seen that if we have a very large CM, and we bias the ampop with +-Vcc, and the CM >Vcc theoretically it should reject the CM, but it's not gonna work if the input > Vcc. eg cm 20V, alimentazione +-15V. So we decided to refer the power supply to CM: we sample the CM and we connect the power supply to this floating CM. This concept already seen can be extended to all the input stages of the instrument: we create AMPLIFIER COMMON

Examples of decoupling solutions

1) AC coupling

2) Optocoupler



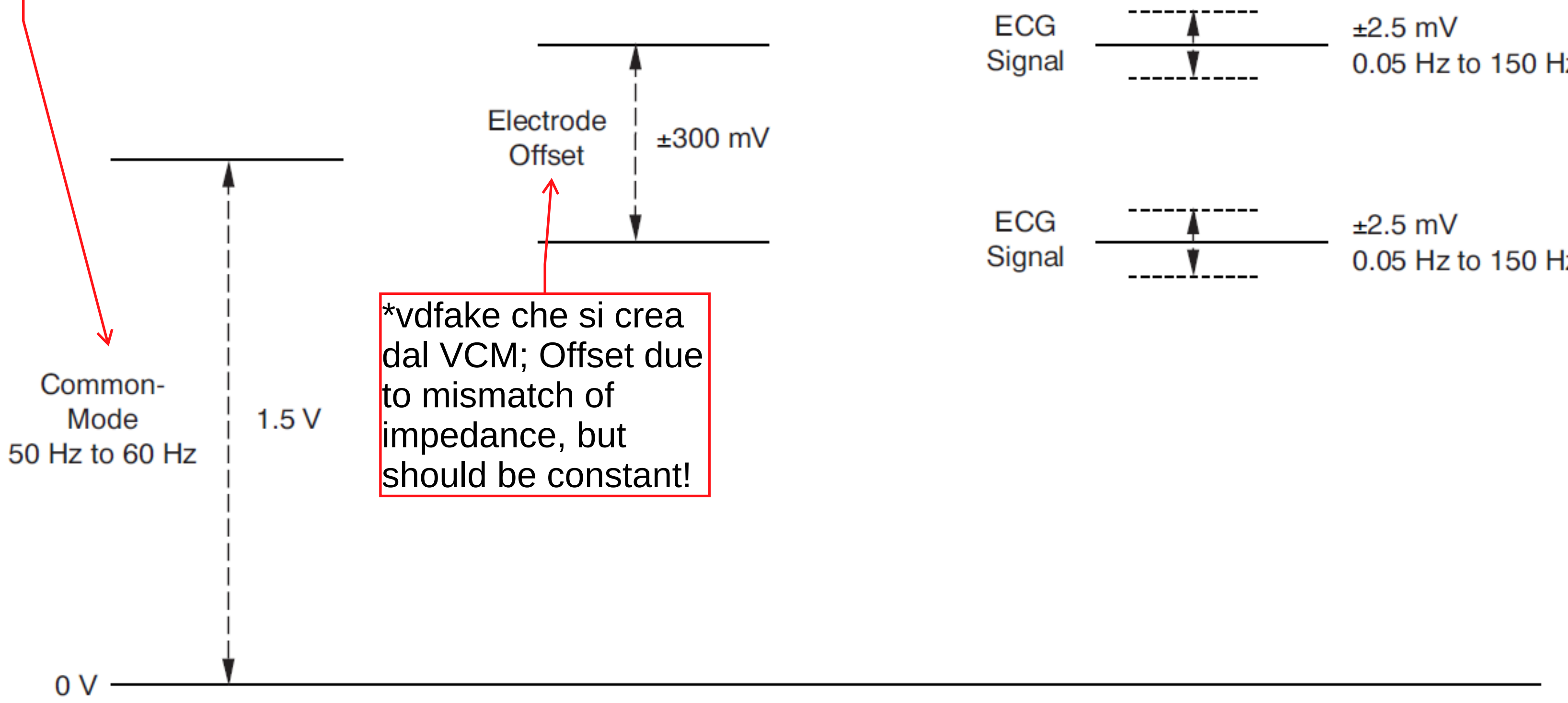
Popoular solution to decoupling.
1) CAPACITOR, DC it transfer the signal but breaks the DC signal
2) OPTOCOUPLER: creates the bridge betweeen the two reference with optical light, because it's insentivie to grounding reference.

Reminder of what we have at the input

ECG signal characteristics

small signal
and their BW

CM, we rely on
CMRR



*vdfake che si crea
dal VCM; Offset due
to mismatch of
impedance, but
should be constant!

2 stage ECG amplifier

Example of ECG amplifier

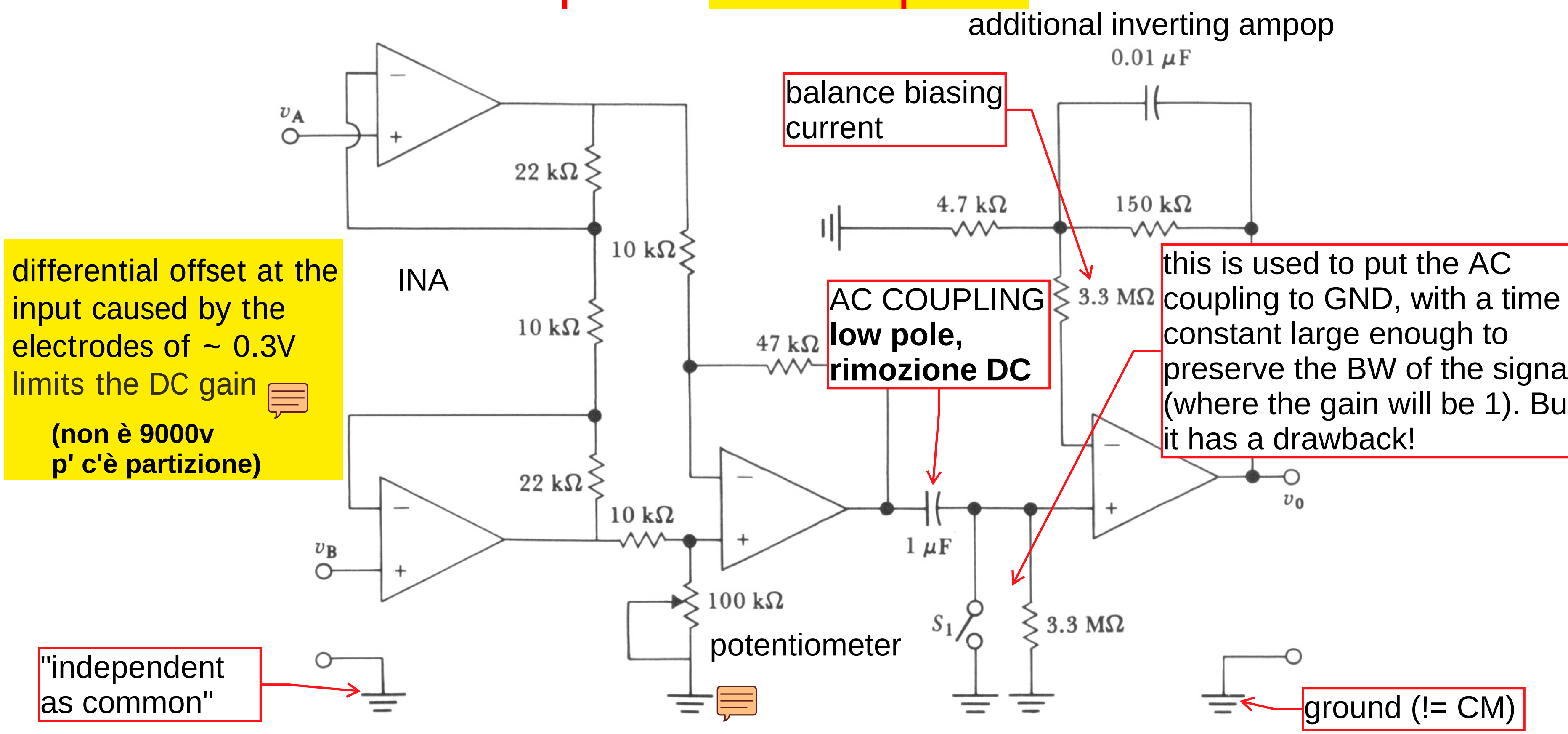


Figure 6.18 This ECG amplifier has a gain of 25 in the dc-coupled stages. The high-pass filter feeds a noninverting-amplifier stage that has a gain of 32. The total gain is $25 \times 32 = 800$. When μA 776 op amps were used, the circuit was found to have a CMRR of 86 dB at 100 Hz and a noise level of 40 mV peak to peak at the output. The frequency response was 0.04–150 Hz for ±3 dB and was flat over 4–40 Hz.

The gain is not distributed only on the first INA, becuase we have electrode offset quite high! the ina is sensitive to these! *è differenziale. we would saturate the amplifier :(. The output of INA would be saturated, we **split the gain in two stages**, INA=> 25, second stage=>32. Overall desired 800, but now the first stage doesnt saturate. (yes, 25*0.3 is already a lot, but then again **the DC voltage is removed by the AC coupling**)