BENG 186B: Principles of Bioinstrumentation Design

Lecture 12

Biopotential Amplifiers: Problems and Solutions

References

Webster, Ch. 6 (Sec. 6.3-6.6).

Common-mode and interference reduction   Ch. 6

- Interference : unwanted signals , present with the wanted signals , and considered as "noise"

The goal in good bioinstrumentation design is to maximize the SIGNAL-TO-NOISE RATIO (SNR) :

$$SNR = \frac{S}{N} = \frac{\text{signal power}}{\text{noise power}} \quad \text{(at the output)}$$

Expressed in decibels (dB) :

$$SNR\ (dB) = 10 \log_{10} \frac{S}{N} = 20 \log_{10} \frac{\text{signal amplitude}}{\text{noise amplitude}}$$

because: power is proportional to the square amplitude (magnitude)

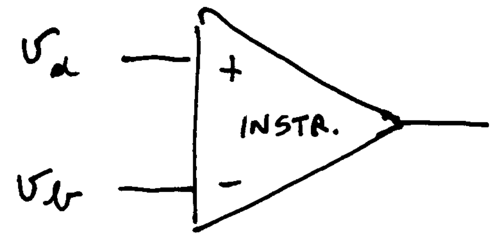
<u>Example</u>: ECG signal = 1 mV pp (peak-to-peak amplitude)
electrode noise = 1 μV pp

$\Rightarrow SNR = 20 \log_{10} 1{,}000 = 60$ dB (pretty good)

<u>Note</u>: can express amplitude in pp (peak-to-peak) or rms (root mean square), but be consistent with signal & noise.

- Common-mode rejection :

Most sources of interference are COMMON-MODE : they appear with equal strength at all terminals of the instrument. A good instrument design is DIFFERENTIAL and eliminates the common-mode component by subtraction.



DIFFERENTIAL BIOAMPLIFIER

$$v_o = \underbrace{A_d}_{\text{DIFFERENTIAL GAIN}} \cdot v_d + \underbrace{A_c}_{\text{COMMON-MODE GAIN}} \cdot v_{cm}$$

where $v_d = v_a - v_b$: DIFFERENTIAL signal of interest

$v_{cm} = \dfrac{v_a + v_b}{2}$: COMMON-MODE component NOT of interest (drift, noise)

Common-mode rejection ratio : $CMRR = |A_d| / |A_c|$

Expressed in dB : $CMRR_{(dB)} = 20 \log_{10} \left(|A_d| / |A_c| \right)$

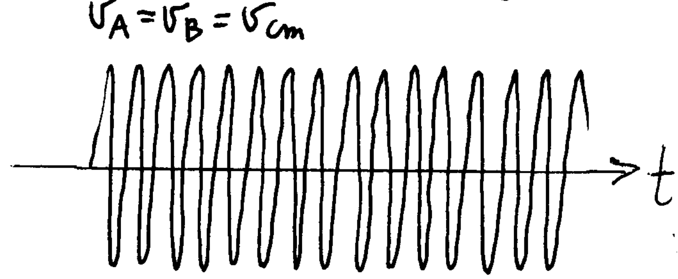
$= A_d\,(dB) - A_c\,(dB)$

Typically would like $CMRR\,(dB) \geq 80\,dB$

(OK for most differential amplifiers)

Good CMRR is critical for attaining a reasonable SNR when amplifying a weak differential signal, such as ECG, subject to substantial common-mode noise, such as 60 Hz line noise:

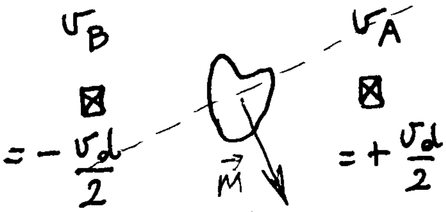
- Still heart in floating (or incompletely grounded) body:

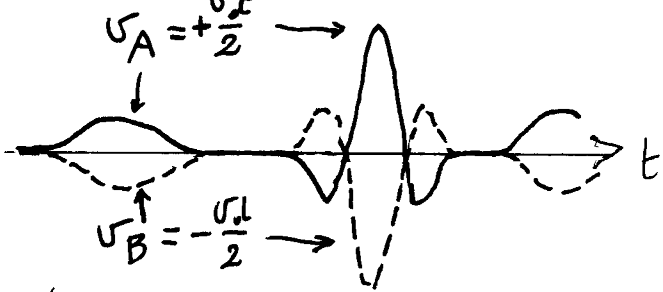


All electrodes pick up the same common-mode voltage v_{cm} due to the body's high volume conduction

Typical 60 Hz line noise as common-mode voltage

- Active heart in perfectly grounded body:

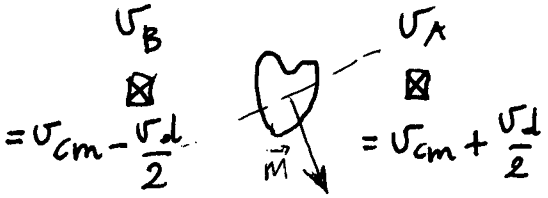


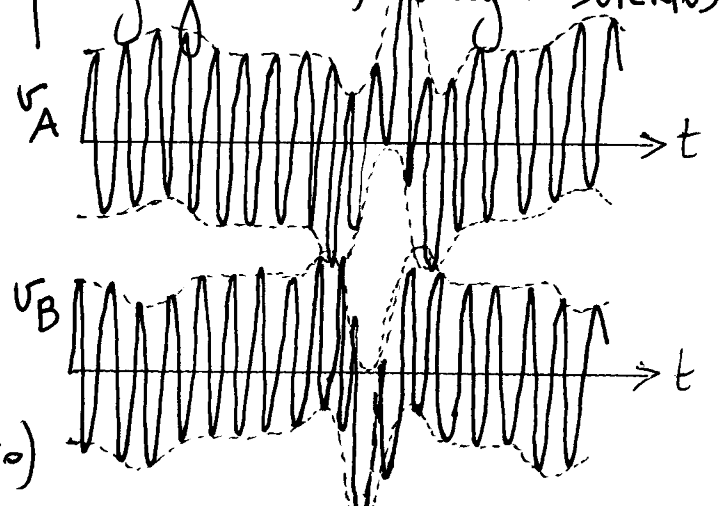


Electrodes pick up ECG leads differentially, relative to the body ground at same potential as the instrument ground.

(NOTE: v_A and v_B are not necessarily complementary, but the average of all (RA, LA, LL) electrodes is zero here.)

- Active heart in actual (incompletely grounded) body: SUPERPOSITION





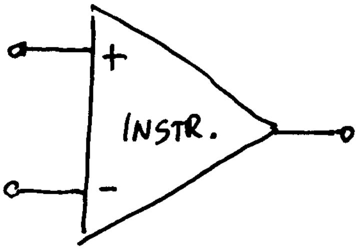
low SNR!

(typically $\ll 1$, or $SNR_{(dB)} < 0$)

- Effect of differential amplification with high CMRR:

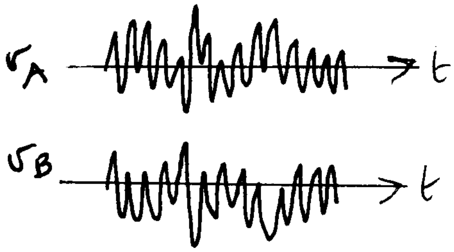
$v_A = v_{cm} + \frac{v_d}{2}$

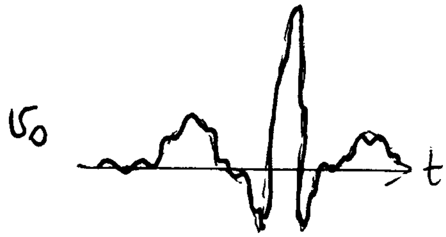
$v_B = v_{cm} - \frac{v_d}{2}$



$v_o = A_d v_d + A_c v_{cm}$

$CMRR = \left|\frac{A_d}{A_c}\right| \gg 1$





$$SNR_{in} = \frac{|\frac{1}{2} v_d|}{|v_{cm}|}$$

(either v_A or v_B)

$$SNR_{out} = \frac{|A_d \cdot v_d|}{|A_c \cdot v_{cm}|} = \underbrace{\frac{A_d}{A_c}}_{CMRR} \cdot \underbrace{\frac{v_d}{v_{cm}}}_{2\, SNR_{in}}$$

$\Rightarrow$ $SNR_{out} = 2 \cdot CMRR \cdot SNR_{in}$

or $SNR_{out}\,(dB) = 6\,dB + CMRR\,(dB) + SNR_{in}\,(dB)$

$\Rightarrow$ CMRR helps boost the SNR directly!

<u>Example</u>: $v_{cm} = 100\ mV_{pp}$ typically observed (without active grounding)

$v_d = 1\ mV_{pp}$ typical lead II ECG

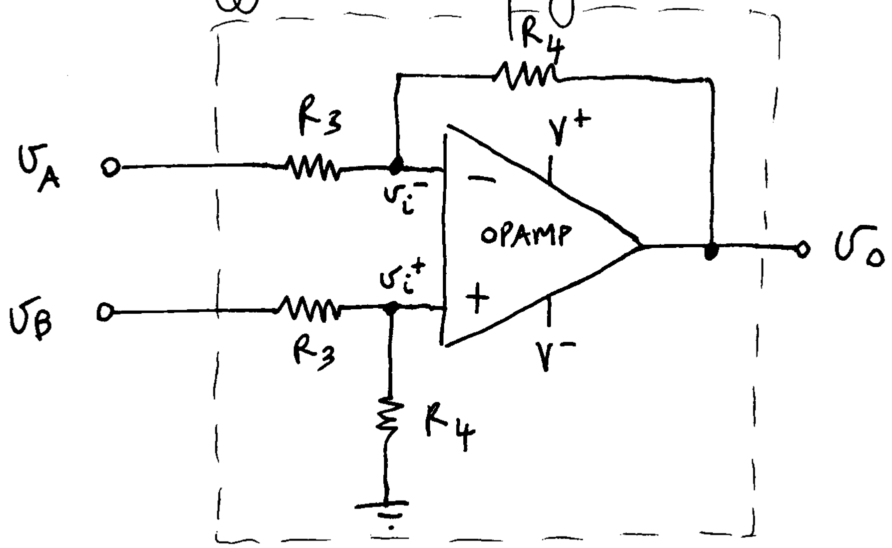
$\rightarrow$ $SNR_{in} = -46\ dB$!!

$\Rightarrow$ Need $CMRR \geqslant 80\ dB$ to obtain $SNR_{out} \geqslant 40\ dB$

(the least useful)

How to build an amplifier with high CMRR ?

TRIAL 1 : Differential amplifier (Sec. 3.4)



(NOTE : Can make the output trivially NON-INVERTING by swapping the v_A and v_B inputs.)

KCL@ v_i^- : $$\frac{v_A - v_i^-}{R_3} = \frac{v_i^- - v_o}{R_4}$$

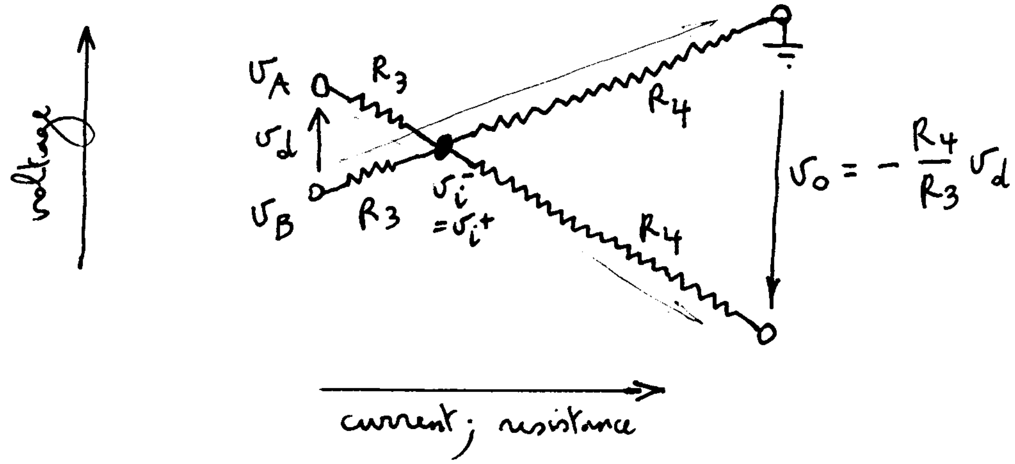
and $v_i^- = v_i^+$

IDEAL OPAMP

KCL@ v_i^+ : $$\frac{v_B - v_i^+}{R_3} = \frac{v_i^+ - 0}{R_4}$$

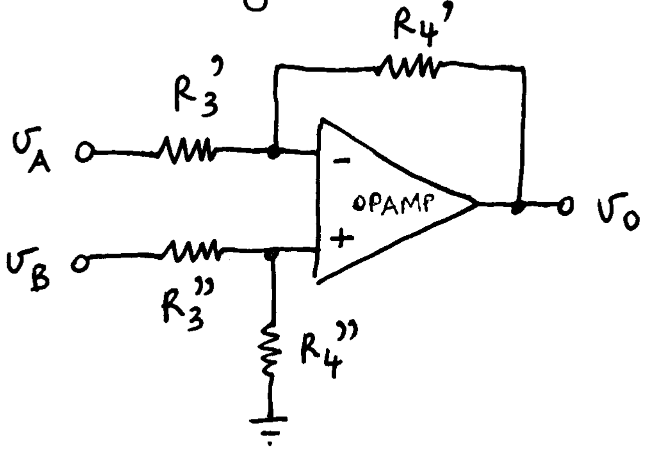
$$\Rightarrow \quad v_o = -\frac{R_4}{R_3}(v_A - v_B) \quad , \text{ or } \quad \begin{cases} A_d = -\dfrac{R_4}{R_3} \\ A_c = 0 \\ \text{CMRR} = \infty \quad (!) \end{cases}$$

Interpretation : voltage drop is linear in resistance :



Problems:

1. Sensitivity to TOLERANCE (relative accuracy) in resistance values:



e.g., R_3' and R_3'' : $R_3 \pm 1\%$

R_4' and R_4'' : $R_4 \pm 1\%$

nominal values (design) — relative accuracy (tolerance)

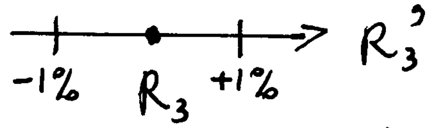
$$v_O = -\frac{R_4'}{R_3'}\cdot v_A + \frac{R_4''}{R_3'}\cdot\frac{R_3'+R_4'}{R_3''+R_4''}\cdot v_B$$

$$\Rightarrow \quad A_d = -\frac{1}{2}\frac{R_4'}{R_3'} - \frac{1}{2}\frac{R_4''}{R_3'}\frac{R_3'+R_4'}{R_3''+R_4''} \approx -\frac{R_4}{R_3}$$

$$\text{but} \quad A_C = -\frac{R_4'}{R_3'} + \frac{R_4''}{R_3'}\frac{R_3'+R_4'}{R_3''+R_4''} \neq 0 \; !$$

Corner analysis: worst effects are expected when the resistances are <u>at</u> their corner tolerance values, i.e:
<u>at</u> +1%, or <u>at</u> −1%, for ±1% tolerance interval:

e.g. for R_3':



$\longleftrightarrow$ guaranteed range of R_3' values for a nominal R_3 value

lowest corner (worst)

highest corner (worst)

All corners considered :

$\frac{R_3' - R_3}{R_3}$:	+1%	-1%			+1%	-1%	$\boxed{+1\%}$	$\boxed{-1\%}$
$\frac{R_4' - R_4}{R_4}$:	+1%	-1%			-1%	+1%	$\boxed{-1\%}$	$\boxed{+1\%}$
$\frac{R_3'' - R_3}{R_3}$:			+1%	-1%	+1%	-1%	$\boxed{-1\%}$	$\boxed{+1\%}$
$\frac{R_4'' - R_4}{R_4}$:			+1%	-1%	-1%	+1%	$\boxed{+1\%}$	$\boxed{-1\%}$

NO EFFECT ! Same voltage division as for nominal R_3 & R_4

NO EFFECT ON A_c ! Same voltage division for v_A & v_B

GREATEST EFFECT ON A_c $\approx \pm 4\%$ of A_d

Worst case : $A_c \approx \pm 4\%$ of A_d

$\Rightarrow$ $CMRR = \left| \frac{A_d}{A_c} \right| \approx 25$! clearly unacceptable

2. Differential gain A_d is limited by ratio R_4/R_3.
Can't make this ratio too large in practice !
e.g. $\begin{cases} R_3 = 1k\Omega \\ R_4 = 100k\Omega \end{cases} \Rightarrow A_d = -\frac{R_4}{R_3} = -100$

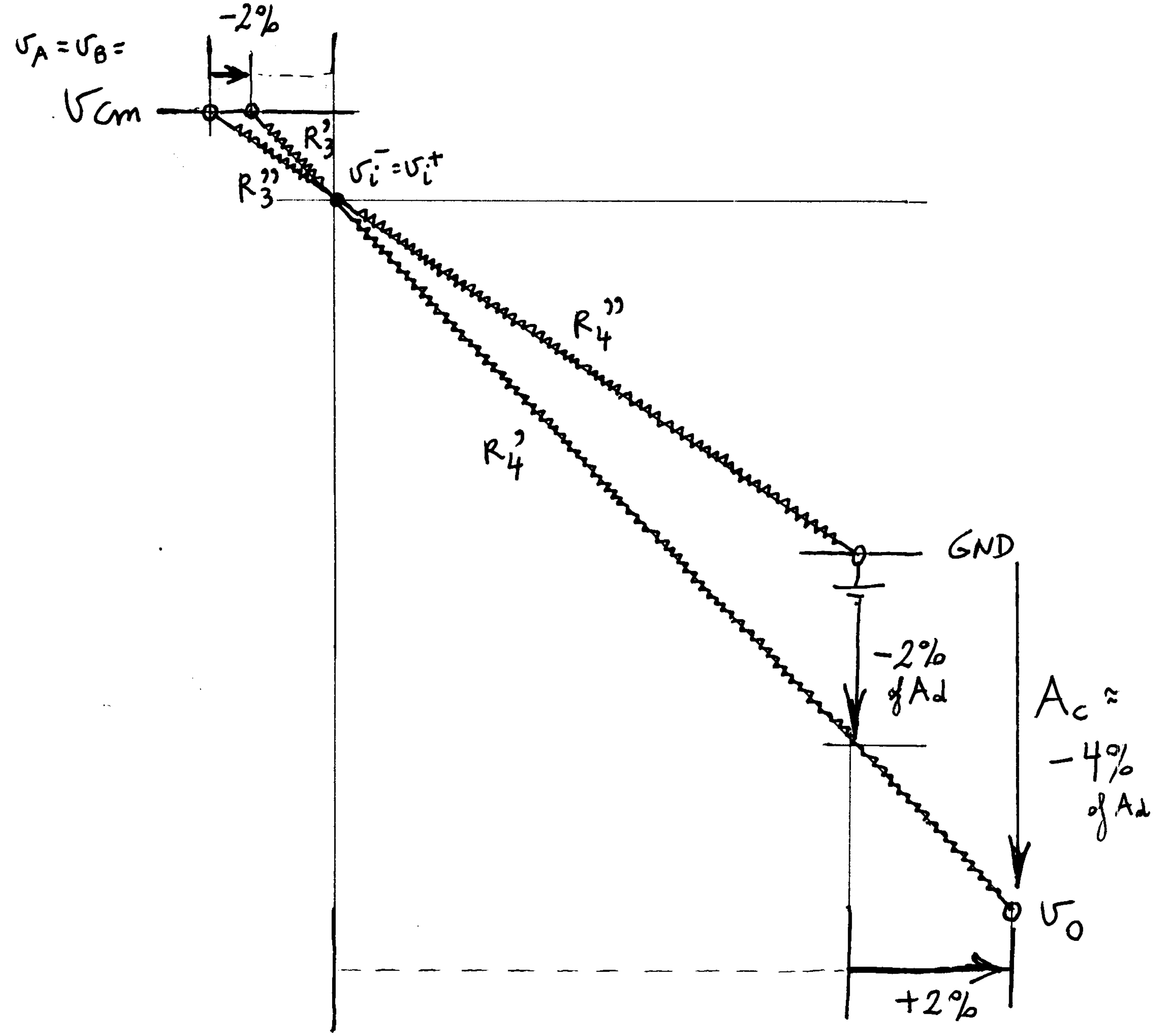
3. Input impedance is too low for use with practical electrodes
$Z_{in}: \begin{cases} \bullet\ v_A : \quad R_3 = 1k\Omega \\ \bullet\ v_B : \quad R_3 + R_4 = 101k\Omega \end{cases} < R_{electrode} \sim 150k\Omega - 2M\Omega$

$\rightarrow$ The signal is lost before it reaches the amplifier input
$\rightarrow$ Input impedance mismatch further contributes to low CMRR

Graphical interpretation of $A_c \approx \left(1 - \frac{R_3''}{R_3'} \cdot \frac{R_4'}{R_4''}\right) \cdot A_d$

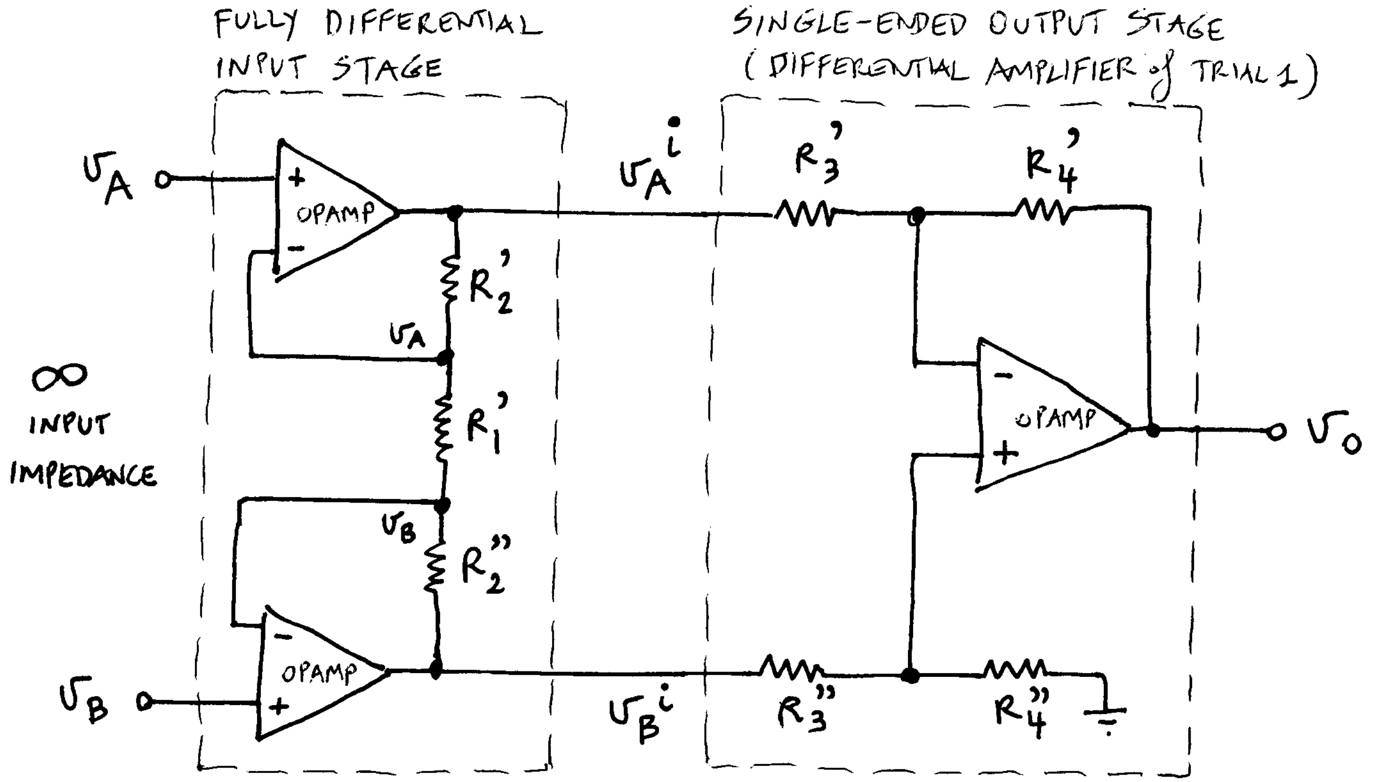
(for large $A_d = -\frac{R_4}{R_3}$)

e.g. : $\begin{cases} R_3' : -1\% \\ R_4' : +1\% \end{cases}$ $\begin{cases} R_3'' : +1\% \\ R_4'' : -1\% \end{cases}$ $\Rightarrow$ $A_c \approx -4\%$ of A_d



Let us try to fix all these problems by adding a fully differential, high-impedance gain stage in front of this differential amplifier:

TRIAL 2: Instrumentation amplifier (Sec. 3.4)



$R_1' = R_1 \pm 1\%$

$R_2', R_2'' = R_2 \pm 1\%$

$R_3', R_3'' = R_3 \pm 1\%$

$R_4', R_4'' = R_4 \pm 1\%$

$CMRR_{in} \approx 2\frac{R_2}{R_1}$

$$A_{d\,in} \approx 1 + 2\frac{R_2}{R_1}$$

$$A_{c\,in} \approx 1$$

$$A_{d\,out} \approx -\frac{R_4}{R_3}$$

$$A_{c\,out} \approx 0 \pm 0.04\frac{R_4}{R_3}$$

$CMRR_{out} \approx 25$

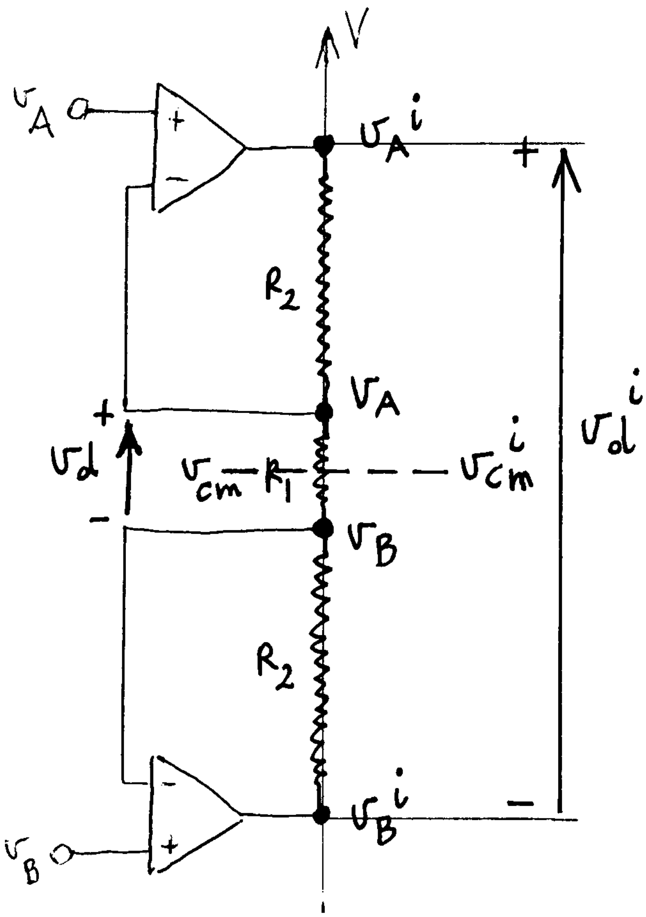
gains multiply across stages!

$$A_d \approx -\left(1 + 2\frac{R_2}{R_1}\right)\cdot\frac{R_4}{R_3}$$

$$A_c \approx 0 \pm 0.04\frac{R_4}{R_3}$$

$CMRR \approx 50\frac{R_2}{R_1}$

$= CMRR_{in} \cdot CMRR_{out}$



- INPUT:

$$v_d = v_A - v_B$$

$$v_{cm} = \frac{v_A + v_B}{2}$$

- INTERMEDIATE STAGE:

$$v_d^i = v_A^i - v_B^i$$

$$\approx \frac{R_2 + R_1 + R_2}{R_1}(v_A - v_B)$$

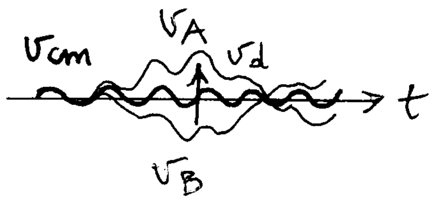
$$= \left(1 + 2\frac{R_2}{R_1}\right) v_d$$

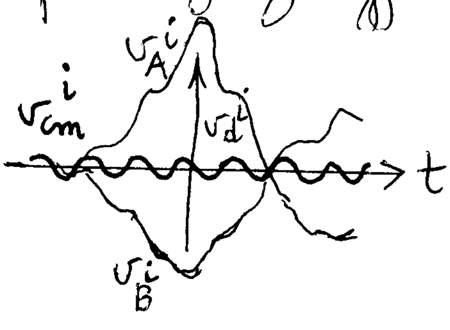
$$v_{cm}^i = \frac{v_A^i + v_B^i}{2} \approx \frac{v_A + v_B}{2} = v_{cm}$$

$$\Rightarrow \quad A_{d\,in} = \frac{v_d^i}{v_d} \approx 1 + 2\frac{R_2}{R_1}$$

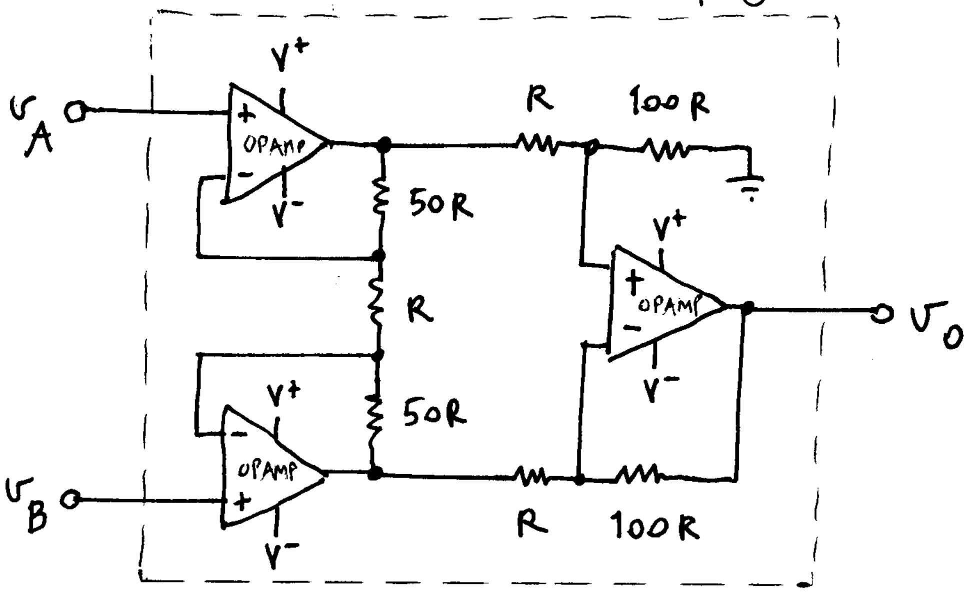
$$A_{c\,in} = \frac{v_{cm}^i}{v_{cm}} \approx 1$$

So the input stage boosts the overall CMRR by $1 + 2\frac{R_2}{R_1} \approx 2\frac{R_2}{R_1} \gg 1$, even though it does not reject common-mode inputs by itself! (purely by high differential gain)





Practical instrumentation amplifier (non-inverting):



OPAMP: $\frac{1}{4}$ TLC084 (quad low-power opamp)

$V^+ = 2.5V$ (2 AAA batteries each)
$V^- = -2.5V$

$R = 1k\Omega$ (or higher, for lower power, at the cost of higher thermal noise)

All resistances $\pm 1\%$

$$\Rightarrow \begin{cases} A_d = +10{,}100 \approx 10{,}000 \\ CMRR \geq 2{,}525 \approx 2{,}500 \quad \text{(worst case)} \\ Z_{in_A} = Z_{in_B} \approx \infty \end{cases}$$

<u>NOTE</u>: Designing/using an I.A. with very high CMRR is NECESSARY but NOT SUFFICIENT for effective common-mode rejection:

→ reduce sources of common-mode noise/interference

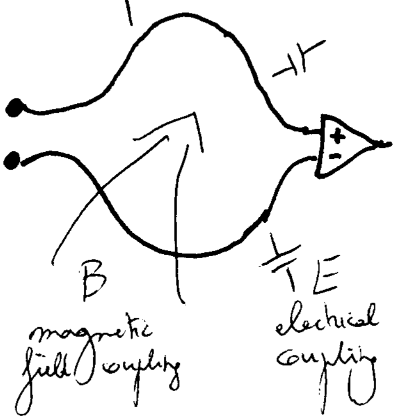
→ reduce sources of CMRR degradation

→ perform ACTIVE GROUNDING (DRIVEN RIGHT LEG)

– Good practice for effective common-mode interference rejection:

- Avoid any differential noise coupling
  - avoid loops in electrode wiring

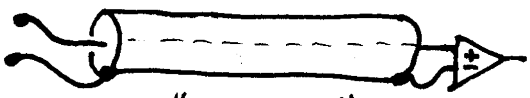
BAD:



GOOD:

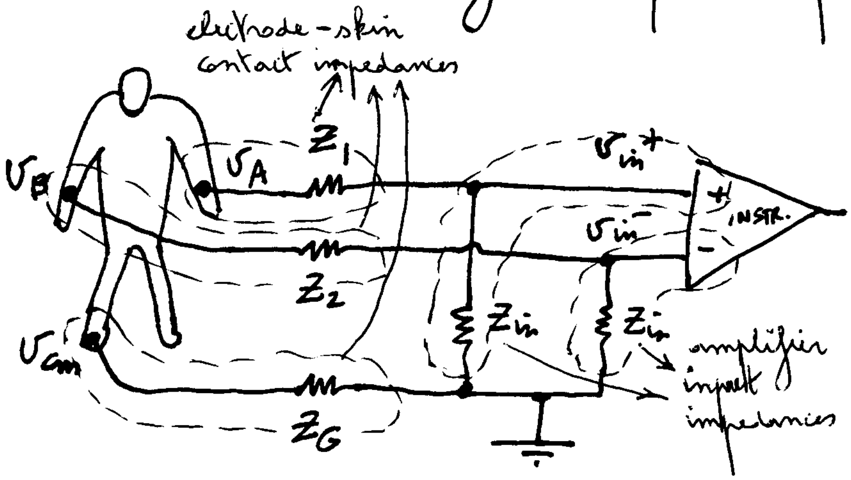
"TWISTED PAIR" cable

  - shield wherever possible



"CO-AXIAL" cable

- Avoid mismatch in electrode impedance, where the instrument has finite input impedance.



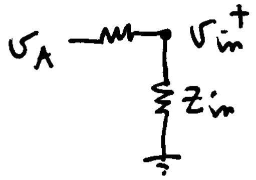
$v_o = A_d \cdot (v_{in}^+ - v_{in}^-)$

"IDEAL" instrumentation amplifier (CMRR $= \infty$)

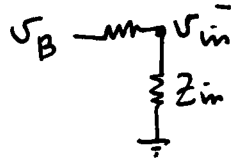
$$v_{in}^+ - v_{in}^- = \frac{Z_{in}}{Z_{in}+Z_1} \cdot v_A - \frac{Z_{in}}{Z_{in}+Z_2} \cdot v_B$$

$|Z_1| \ll |Z_{in}|$
$|Z_2| \ll |Z_{in}|$

$$\approx \frac{Z_2 - Z_1}{Z_{in}} \cdot \underbrace{\frac{v_A + v_B}{2}}_{v_{cm}} + \underbrace{v_A - v_B}_{v_d}$$



and



$$v_{in}^+ - v_{in}^- = \underbrace{\frac{Z_{in}}{Z_{in}+Z_1}} \cdot \overset{v_{cm}+\frac{v_d}{2}}{v_A} - \frac{Z_{in}}{Z_{in}+Z_2} \cdot \overset{v_{cm}-\frac{v_d}{2}}{v_B}$$

$$\frac{Z_{in}}{Z_{in}+Z_1} = \frac{1}{1+\frac{Z_1}{Z_{in}}} \approx 1 - \frac{Z_1}{Z_{in}} \quad \text{for } |Z_1| \ll |Z_{in}|$$

and similarly $\frac{Z_{in}}{Z_{in}+Z_2} \approx 1 - \frac{Z_2}{Z_{in}}$

$$\approx \left(1 - \frac{Z_1}{Z_{in}}\right)\left(v_{cm} + \frac{v_d}{2}\right) - \left(1 - \frac{Z_2}{Z_{in}}\right)\left(v_{cm} - \frac{v_d}{2}\right)$$

$$= \frac{Z_2 - Z_1}{Z_{in}} \cdot v_{cm} + \left(1 + \underbrace{\frac{Z_1+Z_2}{2\,Z_{in}}}_{\ll 1}\right) \cdot v_d$$

$$\approx \frac{Z_2 - Z_1}{Z_{in}} \cdot v_{cm} + v_d$$

$$\Rightarrow \quad v_o \approx \underbrace{A_d \cdot \frac{Z_2 - Z_1}{Z_{in}}}_{A_{c\,eff}} \cdot v_{cm} + A_d \cdot v_d$$

effective common-mode gain due to impedance mismatch

$$\Rightarrow \quad CMRR = \frac{|A_d|}{|A_{c\,eff}|} = \frac{|Z_{in}|}{|Z_2 - Z_1|}$$

$$\Rightarrow \quad CMRR = \frac{|Z_{in}|}{|Z_2 - Z_1|} = \frac{\text{input impedance}}{\text{electrode impedance MISMATCH}}$$

$\Rightarrow$ Mismatch in electrode impedance degrades the CMRR of even a perfect bioamplifier

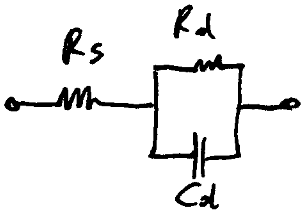
Example:

$$\begin{cases} Z_{in} = 100\,M\Omega & \text{(large!)} \\ Z_1 = 100\,k\Omega \\ Z_2 = 110\,k\Omega \end{cases} \Rightarrow \quad CMRR = 10{,}000 \quad (80\,dB)$$

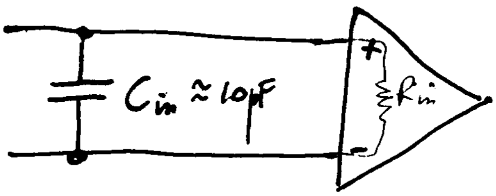
$$\begin{cases} Z_{in} = 1\,M\Omega \\ Z_1 = 200\,k\Omega \\ Z_2 = 100\,k\Omega \end{cases} \Rightarrow \quad CMRR = 10 \quad (20\,dB) \quad \text{BAD!}$$

Note: Z_1, Z_2 and Z_{in} are affected by capacitance, and depend on frequency

- Z_1, Z_2: $\quad R_S + \dfrac{R_d}{1 + j\omega R_d C_d}$



- Z_{in}:



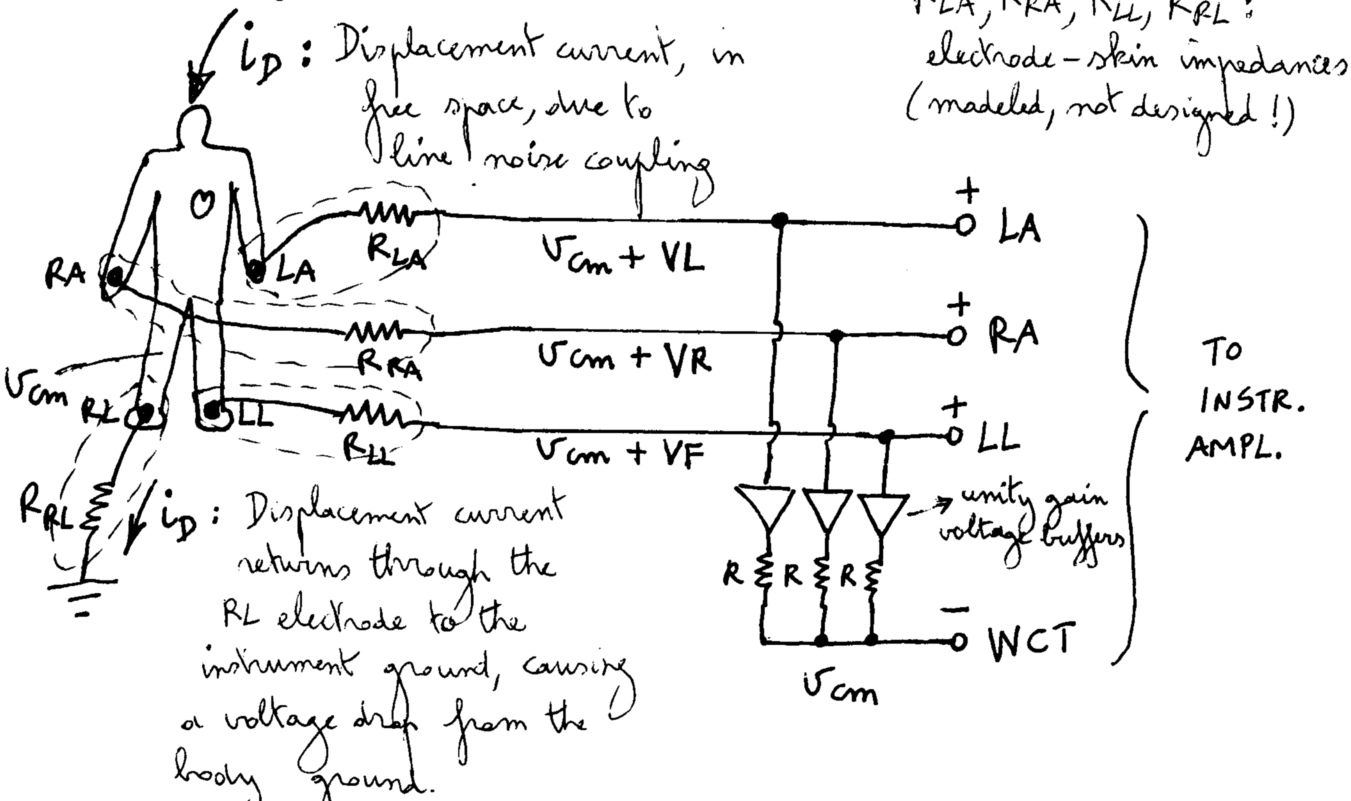
: $\dfrac{1}{j\omega C_{in}}$ is significant at higher frequencies

Remedy: ACTIVE grounding of the body by COMMON-MODE FEEDBACK : "DRIVEN RIGHT LEG" (DRL).

- Active grounding: "DRIVEN RIGHT LEG" (DRL)

  Solution to the CMRR degradation due to impedance mismatch, by reducing v_{cm} directly.

  - Passive grounding (not recommended):



$$v_{cm} \approx R_{RL} \cdot i_D$$

v_{cm} ↓ Common-mode noise in the body

R_{RL} ↓ right leg electrode impedance

i_D ↓ line noise displacement current injected into the body by coupling

$i_D \approx 1\,\mu A_{pp}$ typical

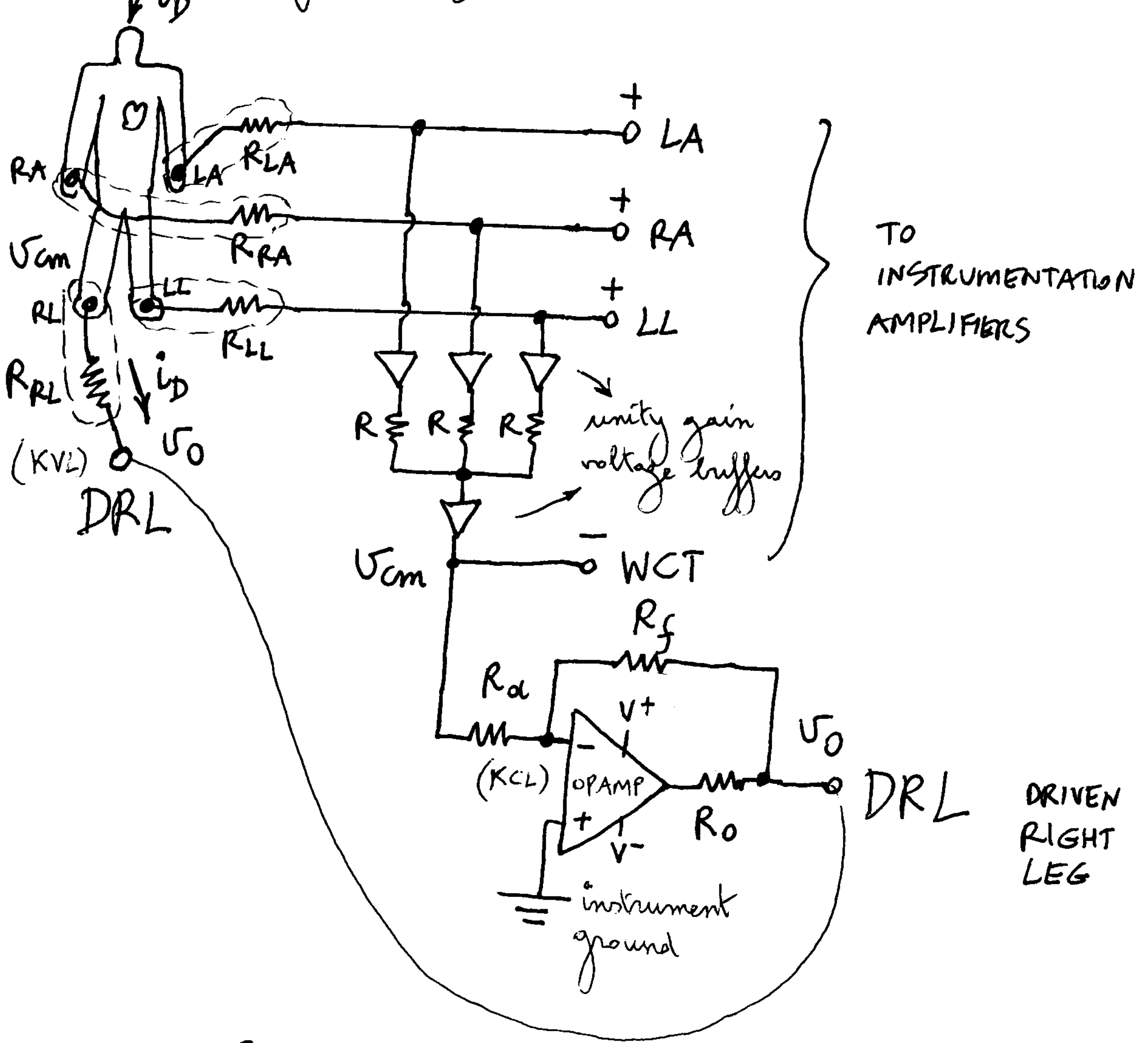
$R_{RL} \approx 100\,k\Omega$ typical

$\Rightarrow v_{cm} \approx 100\,mV_{pp}$

but can be much larger!

Large v_{cm} may leak through at the differential outputs and may saturate the instrumentation amplifiers.

— DRL active grounding :



$$v_0 = -\frac{R_f}{R_a} \cdot v_{cm} \quad (\text{KCL})$$

$$v_{cm} = R_{RL} \cdot i_D + v_0 \quad (\text{KVL})$$

$$v_{cm} = R_{RL}\, i_D - \frac{R_f}{R_a} v_{cm}$$

$$\Rightarrow v_{cm} = \frac{R_{RL}\, i_D}{1 + \frac{R_f}{R_a}} = R_{RL\,eff} \cdot i_D \quad \text{with } R_{RL\,eff} = \frac{R_{RL}}{1 + \frac{R_f}{R_a}}$$

effective electrode resistance equivalent to passive grounding

$\Rightarrow$ DRL active grounding reduces R_{RL} (and v_{cm}) by feedback gain $1 + \frac{R_f}{R_a}$.

Example : $i_D = 1\,\mu A_{pp}$

$R_{RL} = 100\,k\Omega$

$R_a = 1\,k\Omega$

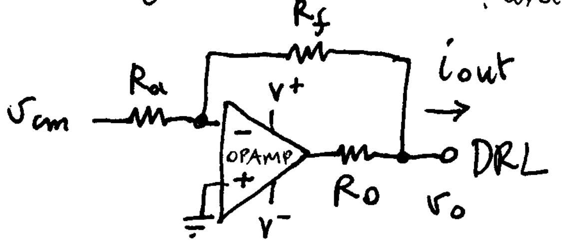
$R_f = 100\,k\Omega$

$$\left.\begin{matrix} R_a = 1\,k\Omega \\ R_f = 100\,k\Omega \end{matrix}\right\} \quad 1 + \frac{R_f}{R_a} = 101 \quad (\cong 100)$$

$\Rightarrow$ $R_{RL\,\text{effective}} \cong 1\,k\Omega$ and $v_{cm} \cong 1\,mV_{pp}$

hundred-fold improvement in SNR
(for common mode noise)

Function of R_o : patient and instrument protection against short circuit
(and caregiver)

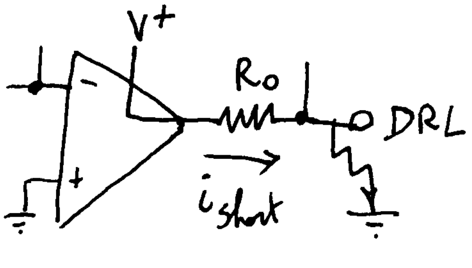


– normal operation : OPAMP in linear region

$\rightarrow$ zero output impedance for effective driving of RL

– current limiting operation during short circuit: OPAMP saturated

$\rightarrow$ maximum current:



$$i_{short} = \frac{V^+}{R_o}$$

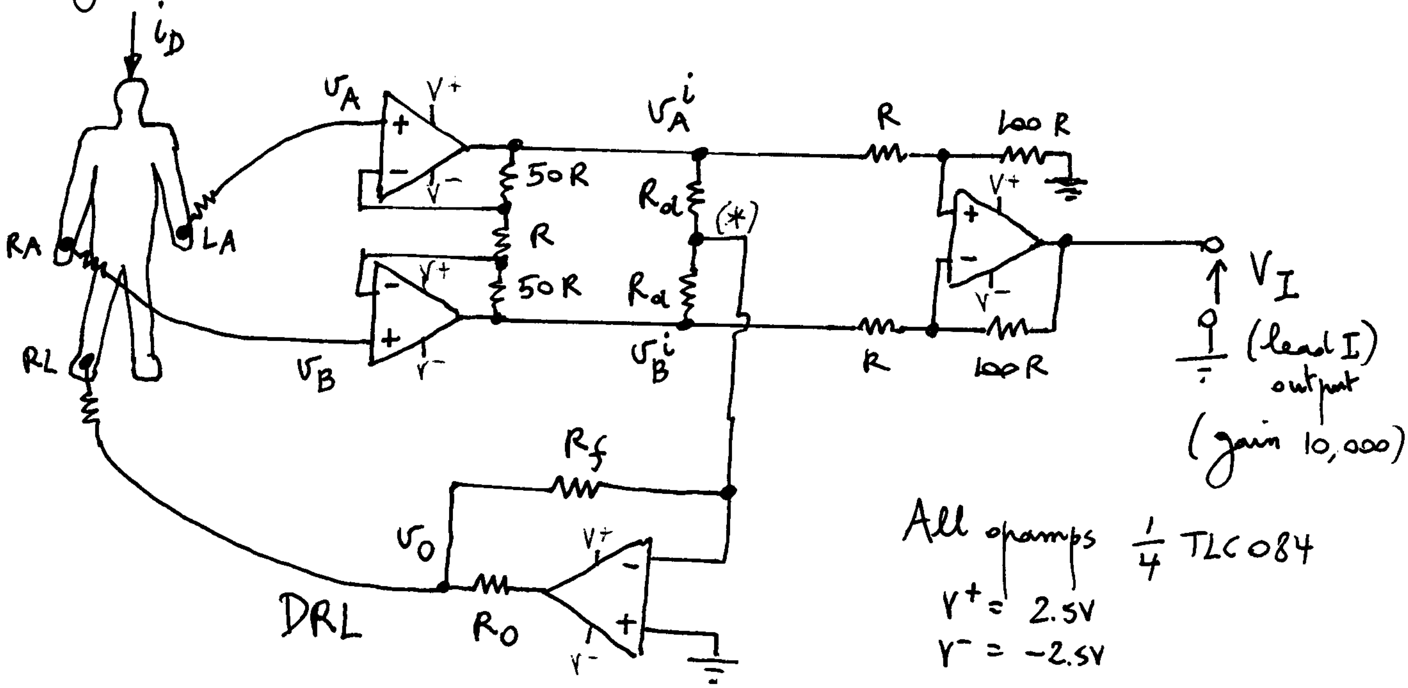
and $\frac{V^-}{R_o}$ in other polarity

e.g.: $\left.\begin{matrix} V^+ = 2.5V \\ R_o = 1\,M\Omega \end{matrix}\right\}$ $i_{short} = 2.5\,\mu A$ (SAFE)

Practical one-lead ECG circuit with DRL active grounding:

(Sec. 6.5 ; Fig. 6.15)

e.g., lead I:



All opamps $\frac{1}{4}$ TLC084

$V^+ = 2.5V$

$V^- = -2.5V$

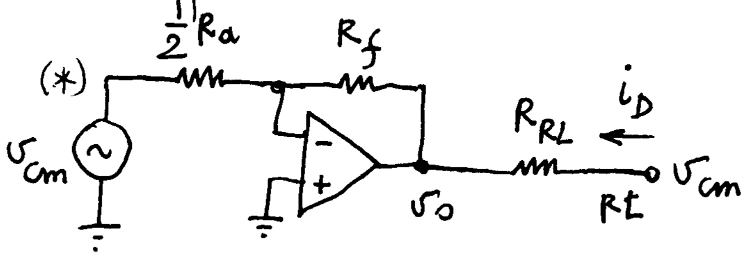
$R = 1\,k\Omega$

$R_a = 20\,k\Omega$ all $\pm 1\%$

$R_f = 1\,M\Omega$

$R_0 = 1\,M\Omega$

DRL equivalent circuit:



$$\Rightarrow \quad R_{RL_{eff}} = \frac{R_{RL}}{1 + 2\frac{R_f}{R_a}}$$

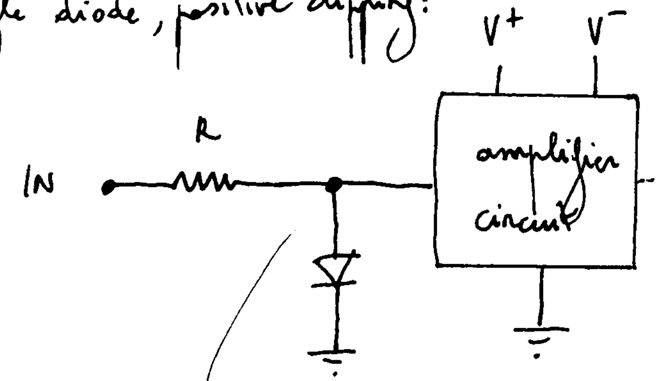
: factor 101 reduction in v_{cm}

and $i_{short} = 2.5\,\mu A$

— Transient protection and AC signal coupling (Sec. 6.4)

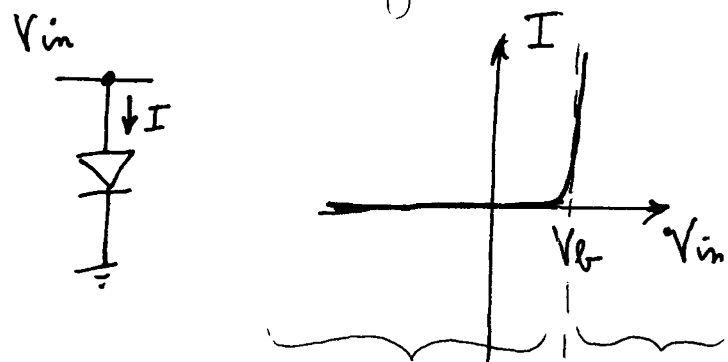
Diodes can be used to clamp over-voltage for protection of voltage-sensitive inputs:

- Single diode, positive clipping:



$R \sim 1-10\,k\Omega$

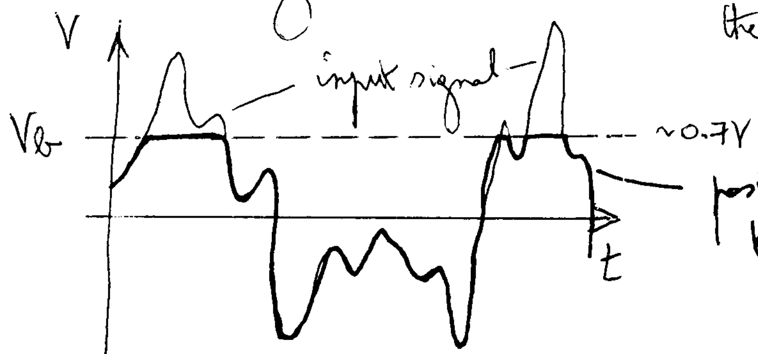
protects the amplifier input from voltages coming in that are greater than $+V_b$:



$V_b \approx 0.7V$ for a typical diode

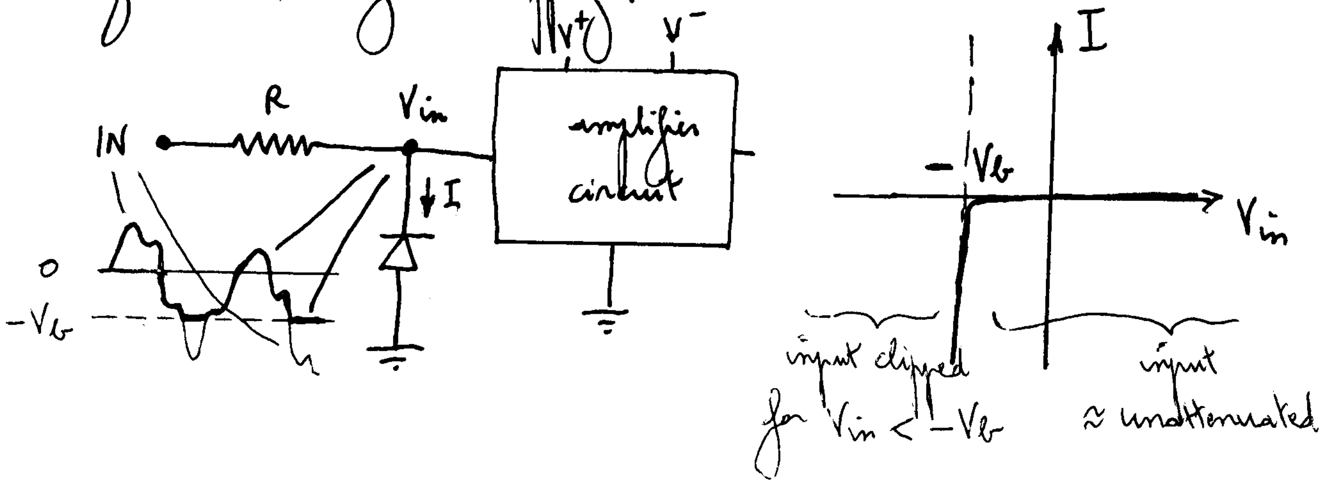
Almost zero current for $V_{in} < V_b$
$\Rightarrow$ input signal passes through unattenuated

Almost infinite current for $V_{in} > V_b$
$\Rightarrow$ amplifier input is clamped at V_b, clipping the signal

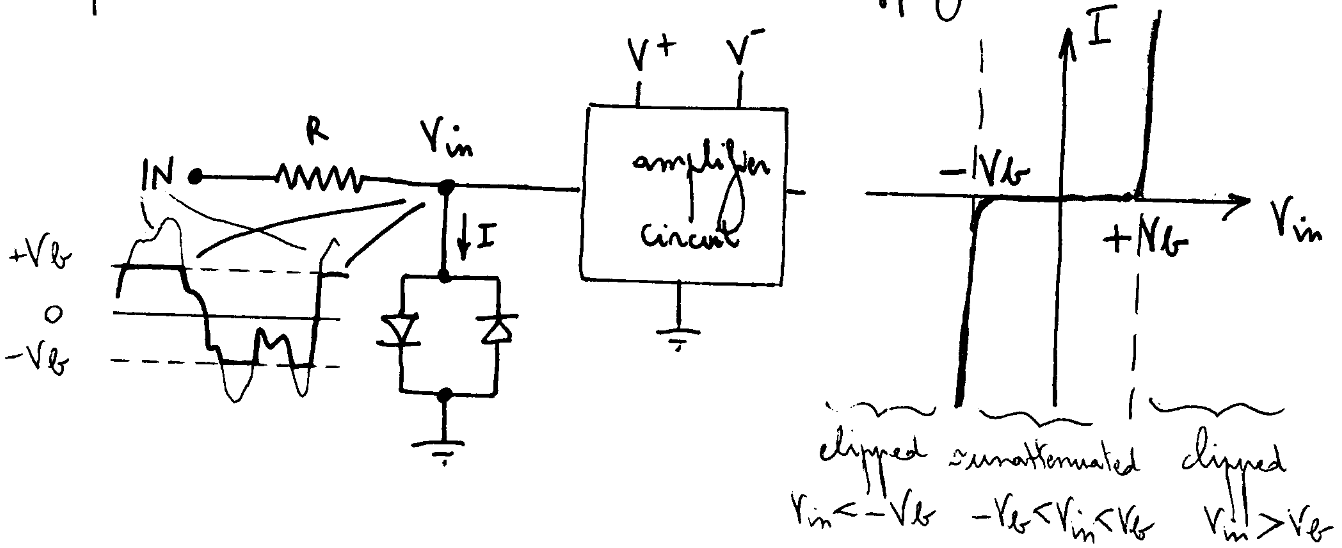


positively clipped input to the amplifier

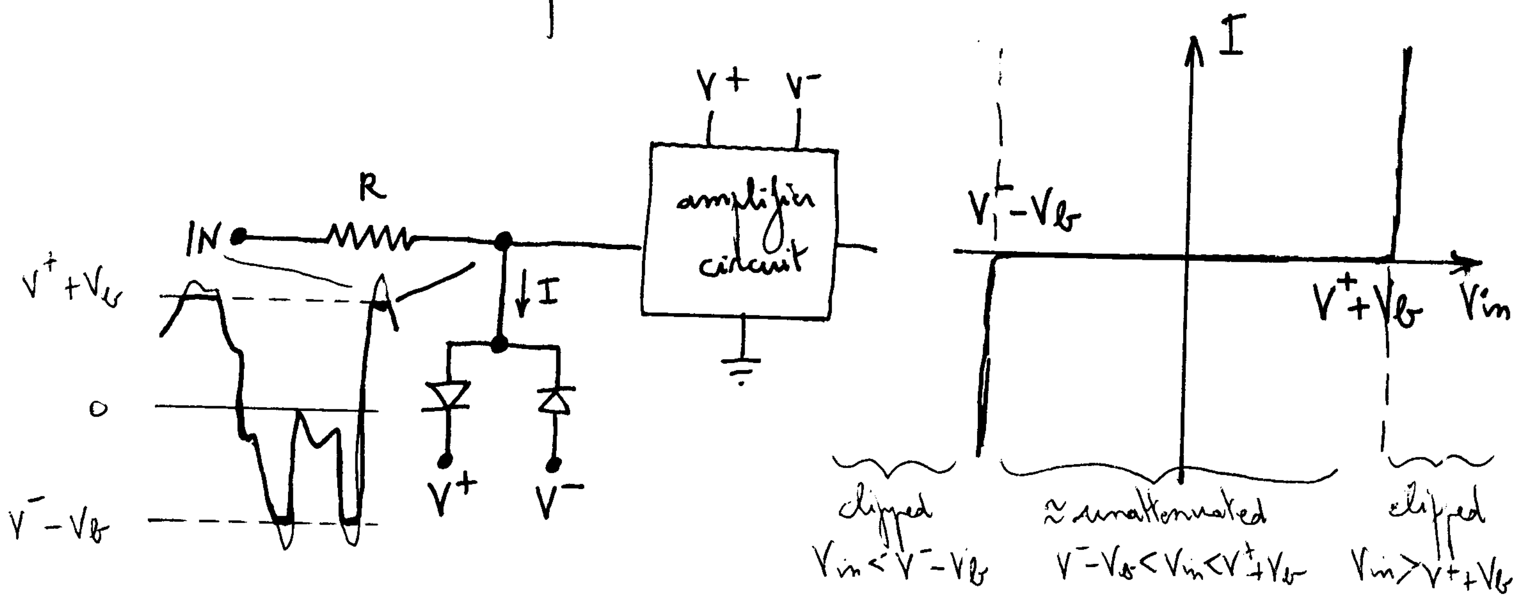
- single diode, negative clipping:



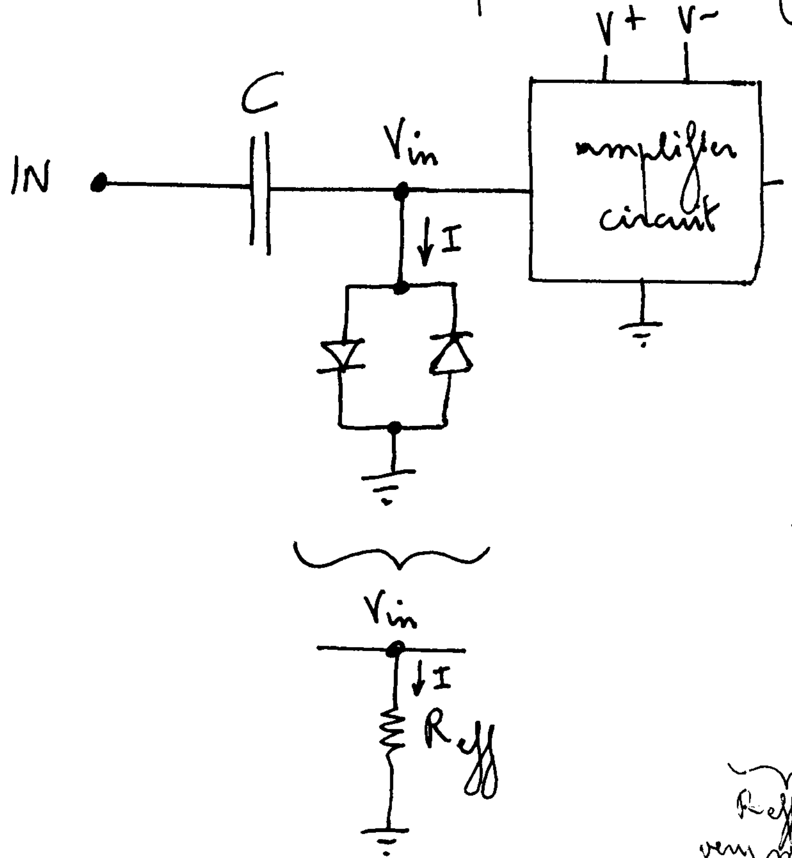
- parallel double diode, bidirectional clipping:



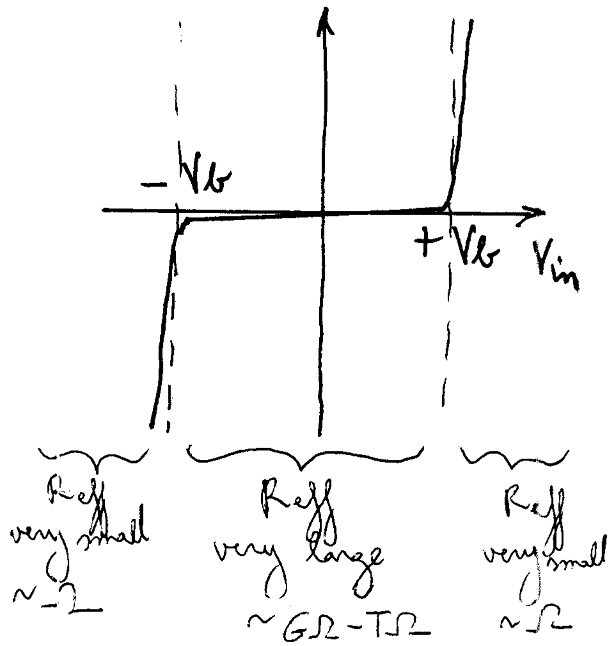
- double-rail diode protection:



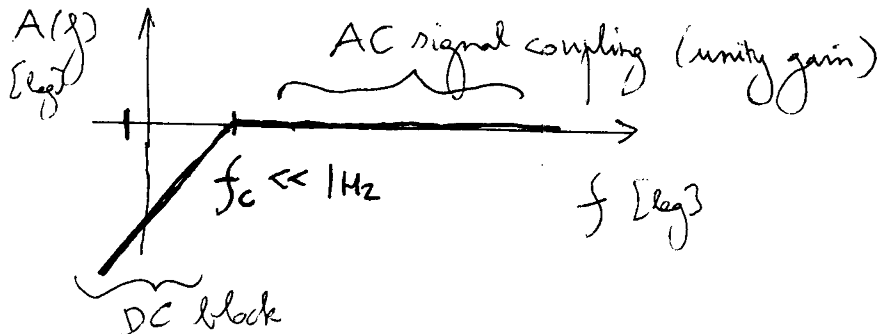
• parallel double diode, capacitive AC signal coupling:



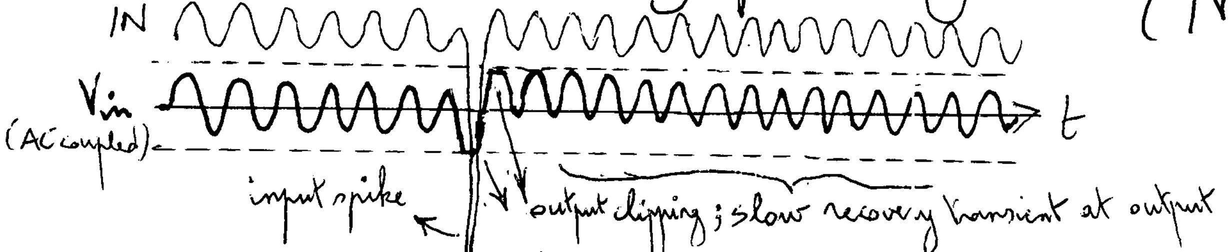
$C \sim 100\,pF - 10\,nF$



⇒ HIGHPASS at very low cut-off frequency $f_c \ll 1\,Hz$
for SMALL SIGNALS

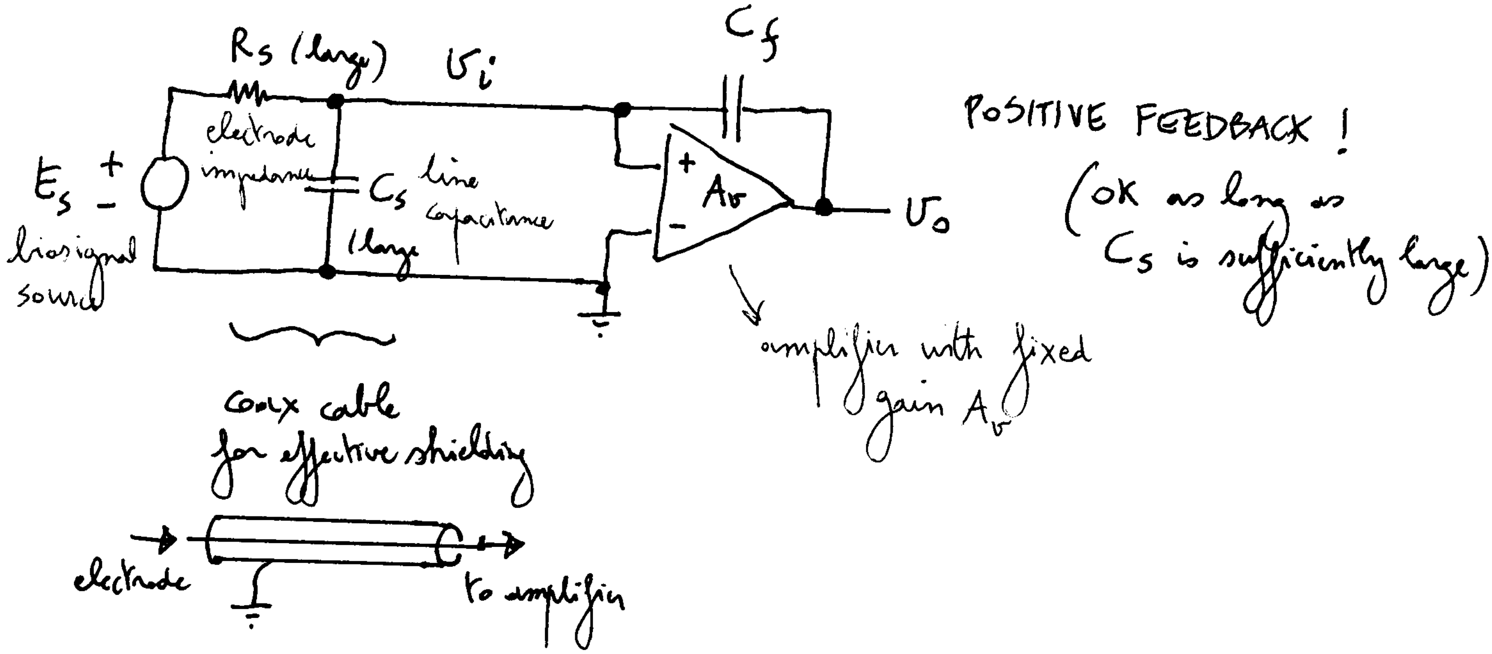


⇒ CLIPPING and over-voltage protection for LARGE SIGNALS
($|V_{in}| > V_b$)



— Impedance cancellation: necessary to compensate for large RC delays due to high electrode impedance and high line capacitance:

(Sec. 6.6)



$\Rightarrow$ (KCL @ v_i)

$$-(E_S - v_i)/R_S + j\omega C_s v_i = j\omega C_f (A_v v_i - v_i)$$

or
$$v_i = \frac{1}{1 + j\omega R_S C_{eff}} E_s$$

$$v_o = \frac{A_v}{1 + j\omega R_S C_{eff}} E_s$$

where $C_{eff} = C_s - (A_v - 1) C_f$ effective capacitance due to feedback

$\Rightarrow$ $Z = 0$ when $C_{eff} = 0$ for $C_S = (A_v - 1) C_f$

but careful to avoid instability! $C_s > (A_v - 1) C_f$