Lecture 12

Biopotential Amplifiers: Problems and Solutions

References

Webster, Ch. 6 (Sec. 6.3-6.6).
- Interference: unwanted signals, present with the wanted signals, and considered as "noise".

The goal in good bioinstrumentation design is to maximize the \textit{signal-to-noise ratio (SNR)}:

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

Expressed in decibels (dB):

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

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

Example: ECG signal = 1 mV pp (peak-to-peak amplitude)  
electrode noise = 1 mV pp  
\[ \Rightarrow \text{SNR} = 20 \log_{10} 1,000 = 60 \text{ dB} \text{ (pretty good)} \]

Note: 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.

![Diagram](image)

\[ V_o = A_d \cdot V_d + A_c \cdot V_{cm} \]

where \( V_d = V_a - V_b \) : **differential** signal of interest

\[ V_{cm} = \frac{V_a + V_b}{2} \] : **common-mode** component not of interest (drift, noise)

**Common-mode rejection ratio**: \( CMRR = \frac{|A_d|}{|A_c|} \)

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

**Typically would like** \( CMRR_{(dB)} \geq 80 \text{ dB} \) (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:

\[ U_A = U_B = U_{cm} \]

All electrodes pick up the same common-mode voltage \( U_{cm} \) due to the body’s high volume conduction.

- Active heart in perfectly grounded body:

\[ U_A = \pm \frac{U_{d}}{2} \]

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

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

\[ U_A = U_{cm} - \frac{U_{d}}{2} \quad U_B = U_{cm} + \frac{U_{d}}{2} \]

Low SNR!

(Extrinsically \( \approx 1 \), or \( \text{SNR}(\text{dB}) < 0 \))

Typical 60 Hz line noise on common-mode voltage.
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 \( V_B \))

\[ SNR_{out} = \frac{|A_d| V_d}{|A_c| V_{cm}} \]

\[ CMRR \gg 2 SNR_{in} \]

\[ \Rightarrow SNR_{out} = 2 \cdot CMRR \cdot SNR_{in} \]

\[ or \quad SNR_{out (dB)} = 6 dB + CMRR (dB) + SNR_{in (dB)} \]

\[ \Rightarrow CMRR \text{ helps boost the SNR directly!} \]

Example:

\[ V_{cm} = 100 \text{ mV}_{pp} \text{ typically observed (without active grounding)} \]

\[ V_d = 1 \text{ mV}_{pp} \text{ typical lead II ECG} \]

\[ \Rightarrow SNR_{in} = -46 \text{ dB} \]

\[ \Rightarrow \text{Need CMRR} \geq 80 \text{ dB to obtain } SNR_{out} \geq 40 \text{ dB} \]

( the least useful)
How to build an amplifier with high CMRR?

**TRIAL 1:** Differential amplifier (Sec. 3.4)

![Amplifier Circuit](image)

\[ \text{KCL@} \: v_i^- : \quad \frac{v_A - v_i^-}{R_3} = \frac{v_i^- - v_O}{R_4} \]

\[ \text{KCL@} \: v_i^+ : \quad \frac{v_B - v_i^+}{R_3} = \frac{v_i^+ - 0}{R_4} \]

\[ v_O = -\frac{R_4}{R_3} (v_A - v_B), \quad \text{or} \]

\[ A_d = -\frac{R_4}{R_3}, \quad A_c = 0 \]

\[ \text{CMRR} = \infty \]  

Interpretation: Voltage drop is linear in resistance:

![Graph showing voltage and current](image)
Problems:

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

\[ V_0 = -\frac{R_4'}{R_3'} \cdot V_A + \frac{R_4''}{R_3''} \cdot \frac{R_3' + R_4'}{R_3'' + R_4''} \cdot V_B \]

\[ \Rightarrow A_d = -\frac{1}{2} \frac{R_4'}{R_3'} - \frac{1}{2} \frac{R_4''}{R_3''} \frac{R_3' + R_4'}{R_3'' + R_4''} \approx -\frac{R_4}{R_3} \]

but \[ A_c = -\frac{R_4'}{R_3'} + \frac{R_4''}{R_3''} \frac{R_3' + R_4'}{R_3'' + R_4''} \neq 0 \]

Corner analysis: worst effects are expected when the resistances are at their corner tolerance values, i.e.:  
\[ \Rightarrow +1\% \text{ or } -1\% \text{ for } \pm 1\% \text{ tolerance interval:} \]

\[ \text{e.g. for } R_3': \quad -1\% \quad R_3' \quad +1\% \]

\[ \text{guaranteed range of } R_3' \text{ values for a nominal } R_3 \text{ value} \]

\[ \text{lowest corner (worst)} \quad \quad \text{highest corner (worst)} \]
All errors considered:

\[
\begin{array}{c|c|c|c|c|c|c}
\frac{R_3^2-R_3}{R_3} & +1\% & -1\% & +1\% & -1\% & +1\% & -1\% \\
\frac{R_4^2-R_4}{R_4} & +1\% & -1\% & -1\% & +1\% & -1\% & +1\% \\
\frac{R_3^3-R_3}{R_3} & +1\% & -1\% & +1\% & -1\% & -1\% & +1\% \\
\frac{R_4^3-R_4}{R_4} & +1\% & -1\% & -1\% & +1\% & +1\% & -1\%
\end{array}
\]

**NO EFFECT!**

Same voltage division as for nominal \(R_3\) & \(R_4\)

**NO EFFECT on \(Ac\)!**

Same voltage division for \(V_A\) & \(V_B\)

**GREATEST EFFECT on \(Ac\)**

\(\approx \pm 4\%\) of \(Ad\)

Worst case: \(Ac \approx \pm 4\%\) of \(Ad\)

\[\Rightarrow \text{CMRR} = \left| \frac{Ad}{Ac} \right| \approx 25! \quad \text{clearly unacceptable}\]

2. **Differential gain \(Ad\) is limited by ratio \(R_4/R_3\).**

Can't make this ratio too large in practice!

e.g. \(\{R_3 = 1k\Omega\ \Rightarrow Ad = -\frac{R_4}{R_3} = -100\}

3. **Input impedance is too low for use with practical electrodes**

\(Z_{in}: \{\)

\(V_A: R_3 = 1k\Omega \quad < \quad R_{elec} \approx 150k\Omega - 2M\Omega\)

\(V_B: R_3 + R_4 = 101k\Omega\)

\(\rightarrow\) The signal is lost before it reaches the amplifier input

\(\rightarrow\) Input impedance mismatch justin contributes to low CMRR
Graphical interpretation of $A_c \approx \left(1 - \frac{R_3'''}{R_3' \cdot \frac{R_4'''}{R_4''}}\right) \cdot A_d$

(e.g.): $\begin{align*}
R_3' & : -1\% \\
R_4' & : +1\% \\
R_3''' & : +1\% \\
R_4''' & : -1\%
\end{align*}$

$\Rightarrow A_c \approx -4\% \text{ of } A_d$
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)

![Circuit Diagram]

\[ R_1 = R_1 \pm 1\% \]
\[ R_2, R_2' = R_2 \pm 1\% \]
\[ R_3, R_3' = R_3 \pm 1\% \]
\[ R_4, R_4' = R_4 \pm 1\% \]

**Input Impedance:** \( \infty \)

**Gain Calculations:**

\[ A_{d_in} \approx 1 + 2 \frac{R_2}{R_1} \]
\[ A_{d_out} \approx -\frac{R_4}{R_3} \]
\[ A_{c_in} \approx 1 \]
\[ A_{c_out} \approx 0 \pm 0.04 \frac{R_4}{R_3} \]

**CMRR Calculations:**

\[ CMRR_{in} \approx \frac{2^{R_2}}{R_1} \]
\[ CMRR_{out} \approx 25 \]

\[ CMRR \approx 50 \frac{R_2}{R_1} \]

**Gains Multiply Across Stages!**
INPUT:

\[ V_d = V_A - V_B \]
\[ V_{cm} = \frac{V_A + V_B}{2} \]

INTERMEDIATE STAGE:

\[ V_d' = V_A - V_B \]
\[ \approx \frac{R_2 + R_1 + R_2}{R_1} (V_A - V_B) \]
\[ = \left( 1 + 2 \frac{R_2}{R_1} \right) V_d \]
\[ V_{cm}' = \frac{V_A + V_B}{2} \approx \frac{V_A + V_B}{2} = V_{cm} \]

\[ A_{d_{in}} = \frac{V_d'}{V_d} \approx 1 + 2 \frac{R_2}{R_1} \]
\[ A_{c_{in}} = \frac{V_{cm}'}{V_{cm}} \approx 1 \]

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):

\[ V^+ = 2.5V \quad (2 \text{ AAA batteries each}) \]
\[ V^- = -2.5V \]

\[ R = 1\, \text{k}\Omega \quad (\text{or higher, for lower power, at the cost of higher thermal noise}) \]

All resistances \( \pm 1\% \)

\[
\begin{align*}
A_d &= +10,100 \approx 10,000 \\
\text{CMRR} &\geq 2,525 \approx 2,500 \quad (\text{unit case}) \\
Z_{inA} &= Z_{inB} \approx \infty
\end{align*}
\]

**Note:** 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 legs)
- Good practice for effective common-mode interference rejection:
  - Avoid any differential noise coupling
    - avoid loops in electrode wiring

- Shield whenever possible

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

\[ \frac{U_{in}^+ - U_{in}^-}{Z_{in}} \cdot \frac{Z_{in}}{Z_{in} + Z_1} \cdot \frac{Z_1}{Z_1 + Z_2} \cdot U_B \]

\[ U_{in}^+ - U_{in}^- = \frac{Z_{in}}{Z_{in} + Z_1} \cdot U_A - \frac{Z_{in}}{Z_{in} + Z_2} \cdot U_B \]

\[ |Z_2| \ll |Z_{in}| \quad \Rightarrow \quad \frac{Z_2 - Z_1}{Z_{in}} \cdot \frac{U_A + U_B}{2} + \frac{U_A - U_B}{U_{in}} \]
\[ V_{in}^+ - V_{in}^- = \frac{Z_{in}}{Z_{in} + Z_1} \cdot V_A - \frac{Z_{in}}{Z_{in} + Z_2} \cdot V_B \]

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

\[ \frac{Z_{in}}{Z_{in} + Z_2} \approx 1 - \frac{Z_2}{Z_{in}} \]

\[ \approx (1 - \frac{Z_1}{Z_{in}})(V_{cm} + \frac{V_d}{2}) - (1 - \frac{Z_2}{Z_{in}})(V_{cm} - \frac{V_d}{2}) \]

\[ = \frac{Z_2 - Z_1}{Z_{in}} \cdot V_{cm} + (1 + \frac{Z_1 + Z_2}{2Z_{in}}) \cdot V_d \]

\[ \approx \frac{Z_2 - Z_1}{Z_{in}} \cdot V_{cm} + V_d \]

\[ \Rightarrow V_0 \approx A_d \cdot \frac{Z_2 - Z_1}{Z_{in}} \cdot V_{cm} + A_d \cdot V_d \]

\[ A_{eff} \quad \text{effective common-mode gain due to impedance mismatch} \]

\[ \Rightarrow CMRR = \frac{|A_d|}{|A_{eff}|} = \frac{|Z_{in}|}{|Z_2 - Z_1|} \]
\[ CMRR = \frac{|Z_{in}|}{|Z_2 - Z_1|} = \frac{\text{input impedance}}{\text{electrode impedance mismatch}} \]

Mismatch in electrode impedance degrades the CMRR of even a perfect biopreamplifier.

**Example:**
\[
\begin{align*}
Z_{in} &= 100 \text{ M} \Omega \quad \text{(large!)} \\
Z_1 &= 100 \text{ k} \Omega \\
Z_2 &= 110 \text{ k} \Omega
\end{align*}
\]
\[ \Rightarrow \quad CMRR = 10,000 \quad \text{(80 dB)} \]

\[
\begin{align*}
Z_{in} &= 1 \text{ M} \Omega \\
Z_1 &= 200 \text{ k} \Omega \\
Z_2 &= 100 \text{ k} \Omega
\end{align*}
\]
\[ \Rightarrow \quad CMRR = 10 \quad \text{(20 dB)} \quad \text{BAD!} \]

**Note:** \( Z_1, Z_2, \text{ and } Z_{in} \) are affected by capacitance, and depend on frequency.

- \( Z_1, Z_2 : \quad R_s + \frac{R_d}{1 + j\omega R_d C_d} \)

- \( Z_{in} : \quad C_{in} \approx 10 \mu F \)

**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):

\[ V_{cm} = R_{RL} \cdot i_{d} \]

Common-mode noise in the body
Right leg noise
Electrode impedance injected into the body by coupling

$\Rightarrow V_{cm} \approx 100$ 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:

\[ V_o = -\frac{R_f}{R_a} \cdot V_cm \]  (KCL)

\[ V_cm = R_{RL} \cdot i_D + V_o \]  (KVL)

\[ V_cm = \frac{R_{RL} \cdot i_D}{1 + \frac{R_f}{R_a}} = R_{RL\text{eff}} \cdot i_D \] with \[ R_{RL\text{eff}} = \frac{R_{RL}}{1 + \frac{R_f}{R_a}} \]

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
\[ R_{in} = 600 \Omega \]
\[ R_a = 1 \Omega \] \[ 1 + \frac{R_f}{R_a} = 101 \] (\( \approx 100 \))
\[ R_{in} \approx 1 \Omega \] and \( V_{cm} \approx 1 \text{ mV pp} \)

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

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

- Normal operation: opamp in linear region
  \[ \rightarrow \] zero output impedance for effective driving of \( R_L \)
- Current limiting operation during short circuit: opamp saturated
  \[ \rightarrow \] maximum current:

\[ i_{shunt} = \frac{V^+}{R_o} \]

\[ \text{and} \ \ \ \ \ \ \end{aligned} \]

\[ V^+ = 2.5V \] \[ i_{shunt} = 2.5 \text{ mA (safe)} \]

\[ R_o = 1 \text{ M\Omega} \]
Practical one-lead ECG circuit with DRL active grounding:

(Sec. 6.5; Fig. 6.15)

e.g., lead I:

\[
\begin{aligned}
&V_A^+ \quad V_A^- \\
&50R \\
&V_B^+ \quad V_B^- \\
&50R \\
&Ra \quad (*) \\
&V_I^+ \quad V_I^- \\
&100R \\
&V_0^+ \quad V_0^- \\
&2.5V \\
&-2.5V \\
&R = 1 k\Omega \\
&Ra = 20 k\Omega \quad \text{all ±1%} \\
&R_f = 1 M\Omega \\
&R_o = 1 M\Omega
\end{aligned}
\]

All opamps \( \frac{1}{4} \) TLC084

\( V = 2.5V \)

\( V = -2.5V \)

DRL equivalent circuit:

\( R_{RL} \text{eff} = \frac{R_{RL}}{1 + 2 \frac{R_f}{R_a}} : \text{factor 101 reduction in } V_{cm} \)

and \( I_{shunt} = 2.5 mA \)
— 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:

![Circuit Diagram]

$V_r \approx 0.7V$

for a typical diode

Almost zero current for $V_{in} < V_r$

Almost infinite current for $V_{in} > V_r$

$V_{in}$ input signal passes through unattenuated

$V_r$ amplifies input is clamped at $V_r$, clipping the signal

$V_{in}$ positively clipped input to the amplifier
- Single diode, negative clipping:

\[ V_{in} \]

\[ R \]

\[ V_{in} \]

In: \[ V^+ \]

\[ V^- \]

\[ -V_b \]

\[ V_{in} \]

- Parallel double diode, bidirectional clipping:

\[ V_{in} \]

\[ R \]

\[ V_{in} \]

In: \[ V^+ \]

\[ V^- \]

\[ +V_b \]

\[ -V_b \]

\[ V_{in} \]

- Double-rail diode protection:

\[ V^+ \]

\[ V^- \]

\[ V_{in} \]

\[ V^+ - V_b \]

\[ V_{in} < V^- - V_b \]

\[ V^- - V_b < V_{in} < V^- + V_b \]

\[ V_{in} > V_b \]
- Parallel double diode, capacitive AC signal coupling:

\[ V_{in} \]
\[ C \]
\[ V_{in} \]
\[ I \]
\[ R_{eff} \]

\[ C \approx \text{Loop F} - \text{Lo F} \]

\[ V_{in} \]
\[ V_{in} \]

\[ V_{in} \approx \text{Small} \]
\[ V_{in} \approx \text{Large} \]
\[ \sim \text{Op-amp} \sim \text{F} \]

\[ f_c \ll 1 \text{ Hz} \]

\[ A(f) \]
\[ f \sim \text{f}_{\text{max}} \]

\[ f_c \ll 1 \text{ Hz} \]

\[ f \sim \text{f}_{\text{max}} \]

\[ f_c \ll 1 \text{ Hz} \]

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\[ f \sim \text{f}_{\text{max}} \]

\[ f_c \ll 1 \text{ Hz} \]

\[ f \sim \text{f}_{\text{max}} \]
Impedance cancellation: necessary to compensate for large RC delays due to high electrode impedance and high line capacitance.

(See. 6.6)

\[ \frac{R_s}{C_s} \]  

Positive feedback! (OK as long as \( C_s \) is sufficiently large)

Amplifier with fixed gain \( A_v \)

Convex cable for effective shielding

electrode

\[ \text{to amplifier} \]

\[ \forall \quad (E_s - v_i)/R_s + jwC_s v_i = jwC_f (A_v v_i - v_i) \]

\[ \text{(KCL)} \]

or \[ v_i = \frac{1}{1 + jwR_sC_{eff}} E_s \]

\[ v_o = \frac{A_v}{1 + jwR_sC_{eff}} E_s \]

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

\[ \Rightarrow \quad Z = 0 \text{ when } C_{eff} = 0 \quad \forall \quad C_s > (A_v - 1)C_f \]

be careful to avoid instability!