Analog Fundamentals of the ECG Signal Chain

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Objectives

- Introduce Basic ECG Concepts
- Motivate the Need for TINA and SPICE Simulation for ECG Analysis
- Introduce Discrete Analog Functions of the ECG Signal Chain
- Motivate Need for Low Cost Integrated ECG Conditioning System
- Introduce the ADS1298 and Its Embedded ECG Circuitry and Functions

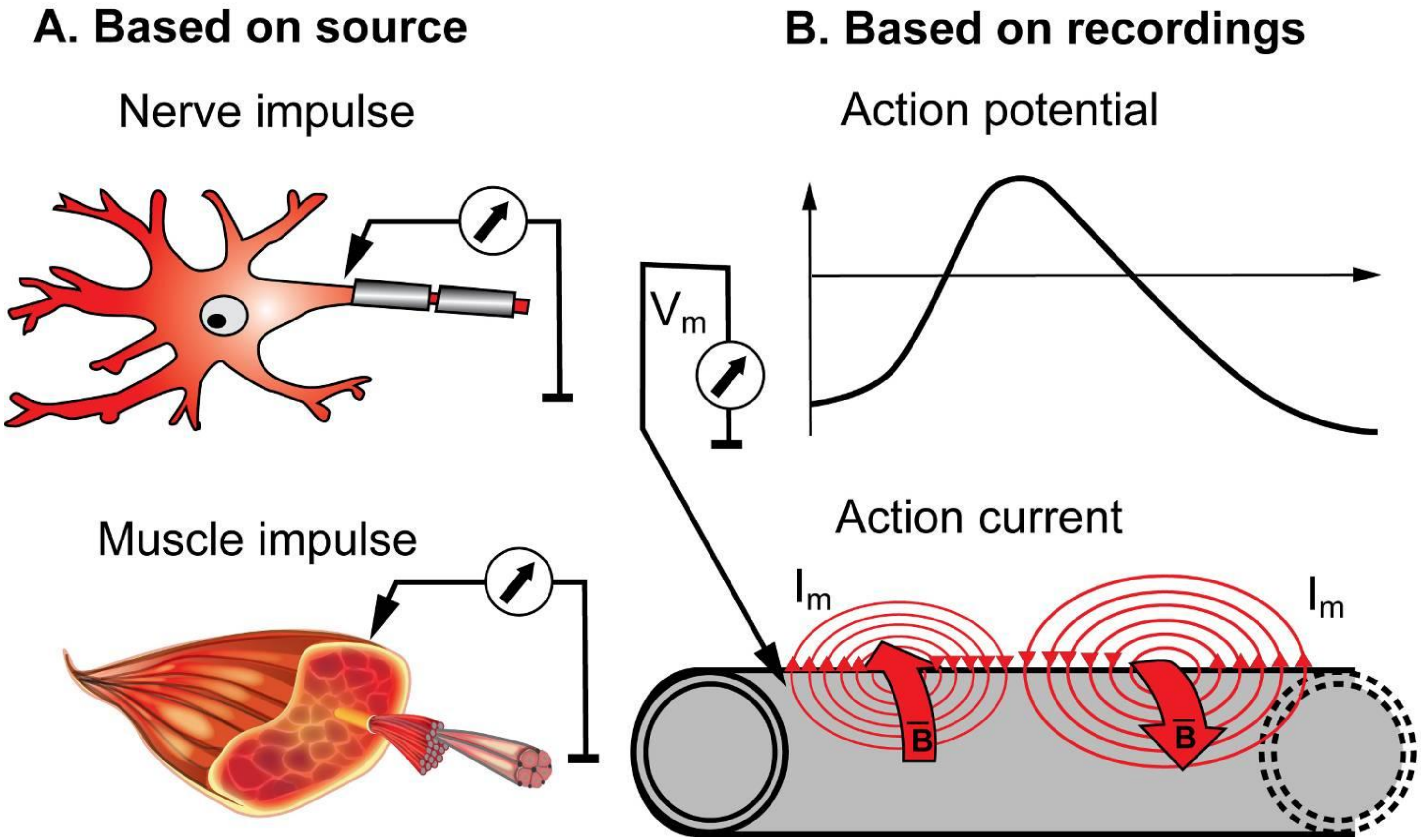
Analog Fundamentals of the ECG Signal Chain

- What is a Biopotential?
- What is ECG?
- The Einthoven Triangle
- Analog Lead Definitions, Derivations, and Purpose
- Modeling the Electrode Interface
- Input Filtering and Defibrillation Protection
- The INA front end
- AC vs. DC coupling
- Right Leg Drive (RLD) Amplifier Selection and Design
- The ECG Shield Drive
- Lead Off Detection
- PACE Detection
- INA post Gain + Analog Filtering
- A/D Conversion Options and Filtering
- **ADS129x** Introduction, Features, and Advantages

What is a Biopotential?

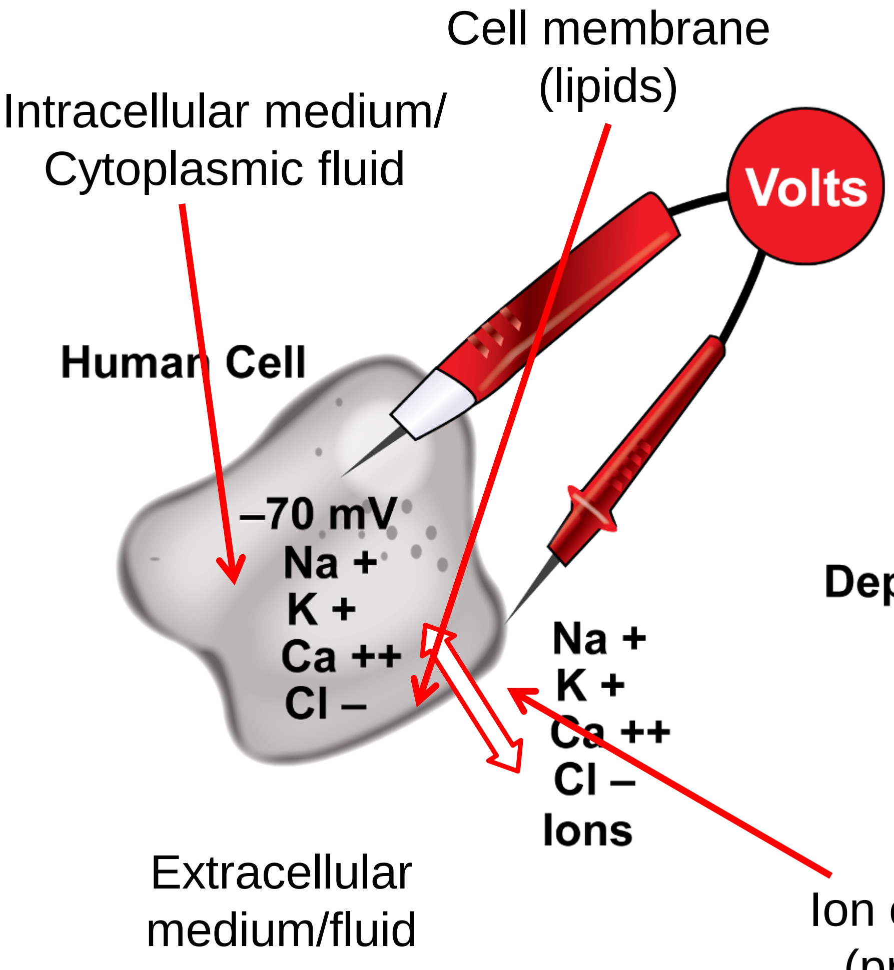
What is a Biopotential

An electric potential measured between living cells



What is a Biopotential

Every cell is like a little battery



Cell membrane
(lipids)

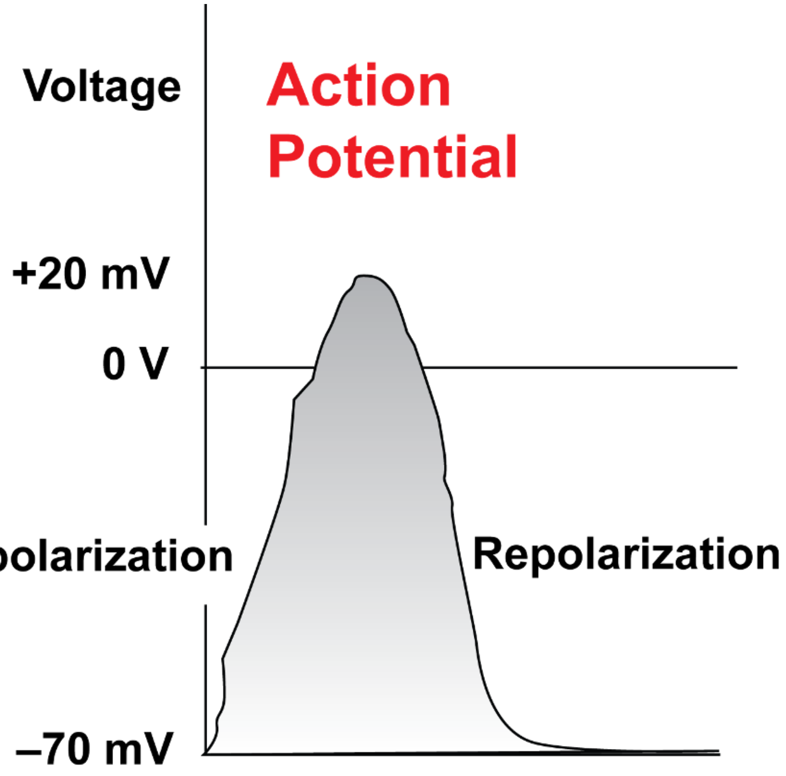
Intracellular medium/
Cytoplasmic fluid

Human Cell

–70 mV
Na +
K +
Ca ++
Cl –

Na +
K +
Ca ++
Cl –
Ions

Extracellular
medium/fluid



Depolarization

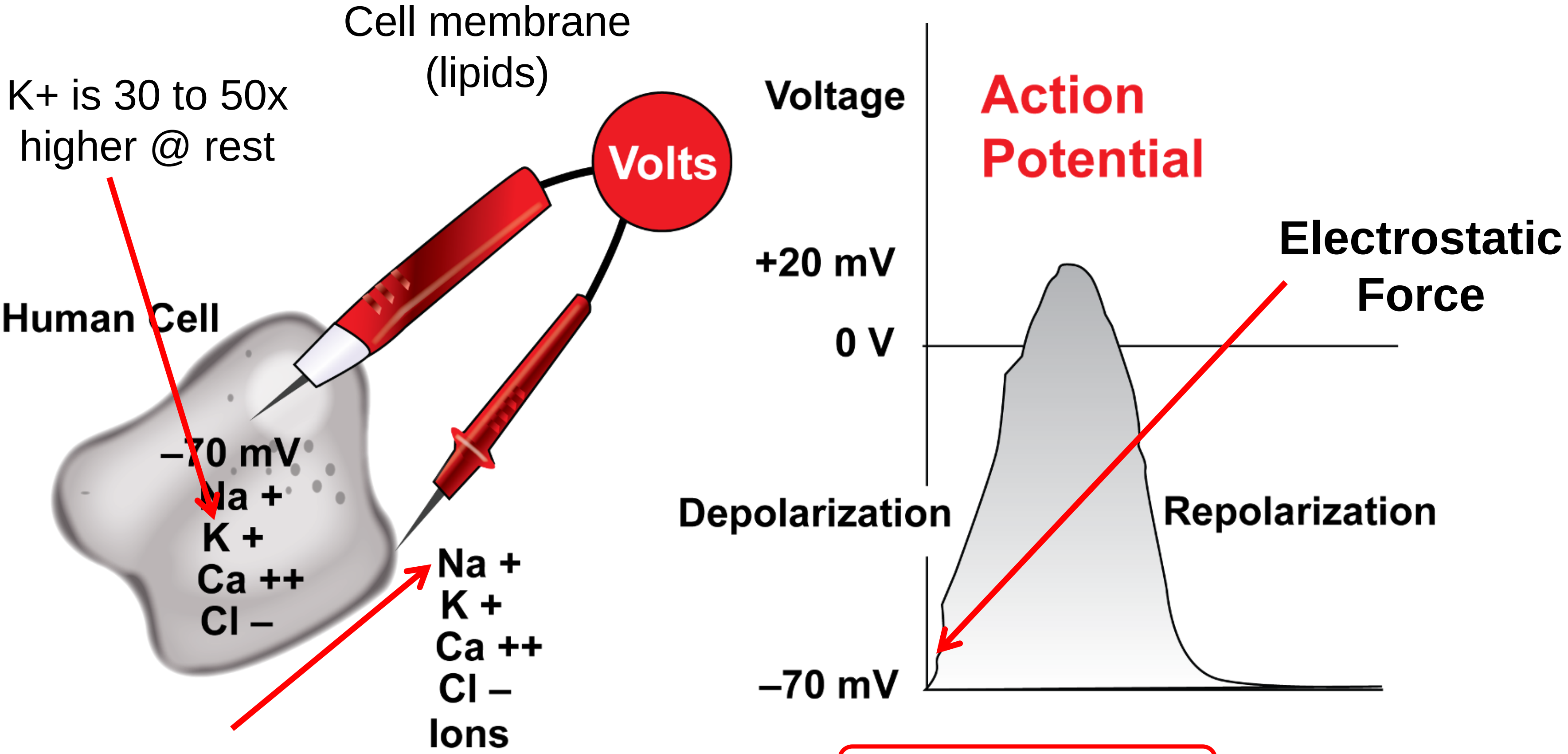
Repolarization

Ion channels
(proteins)

- Resting ion chs.
- Active ion pumps
- Gated ion chs.

What is a Biopotential

Every cell is like a little battery



Cell membrane
(lipids)

K+ is 30 to 50x
higher @ rest

Human Cell

–70 mV

Na +
K +
Ca ++
Cl –

Volts

Na +
K +
Ca ++
Cl –
Ions

Voltage

Action Potential

+20 mV

0 V

–70 mV

Depolarization

Repolarization

Electrostatic Force

Na+ conc. is 10x
higher @ rest

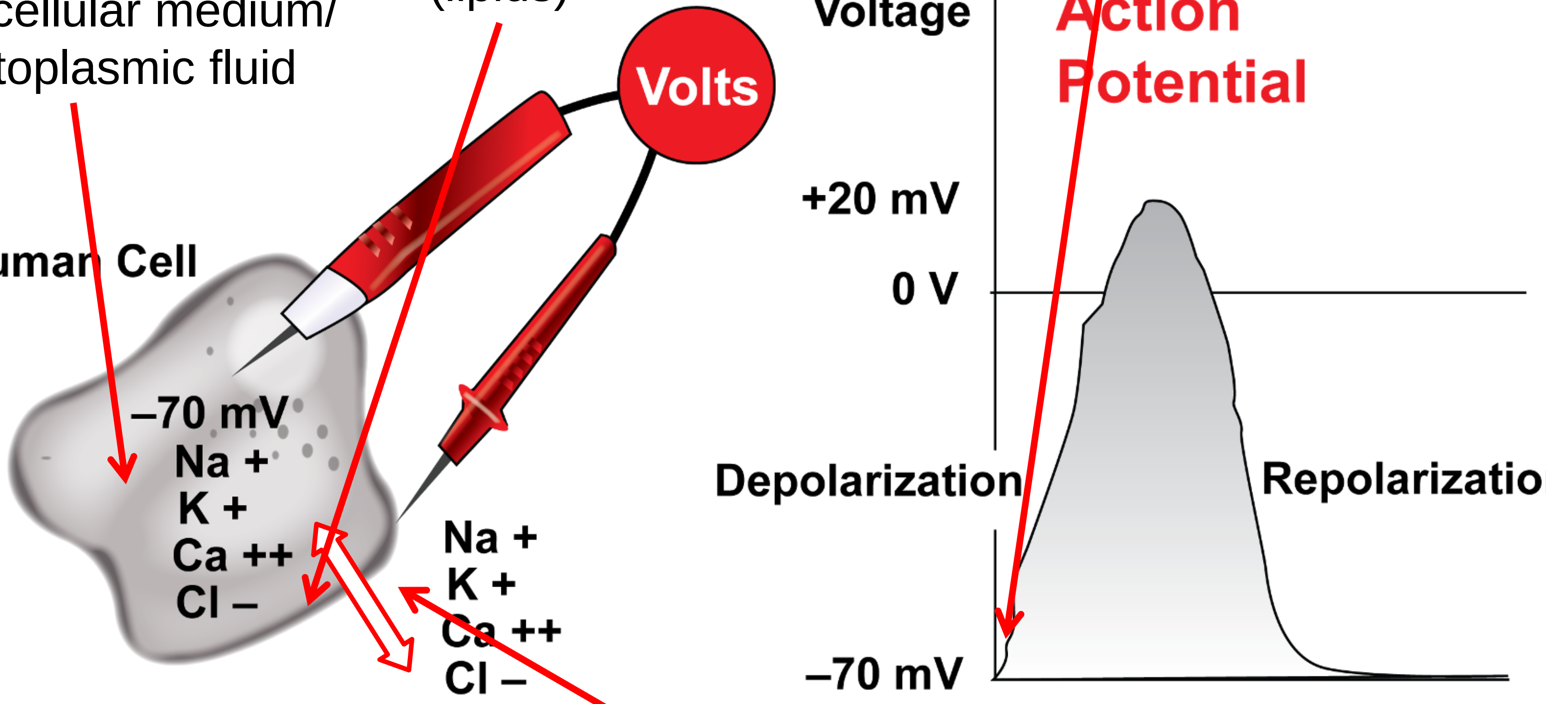
Ion channels
(proteins)

- **Resting ion chs.**
- Active ion pumps
- Gated ion chs.

Diffusional force

What is a Biopotential

Every cell is like a little battery



Neurotransmitter are received at dendrites

Cell membrane (lipids)

Intracellular medium/ Cytoplasmic fluid

Human Cell

–70 mV
Na +
K +
Ca ++
Cl –

Na +
K +
Ca ++
Cl –
Ions

Voltage

Action Potential

+20 mV

0 V

Depolarization

Repolarization

–70 mV

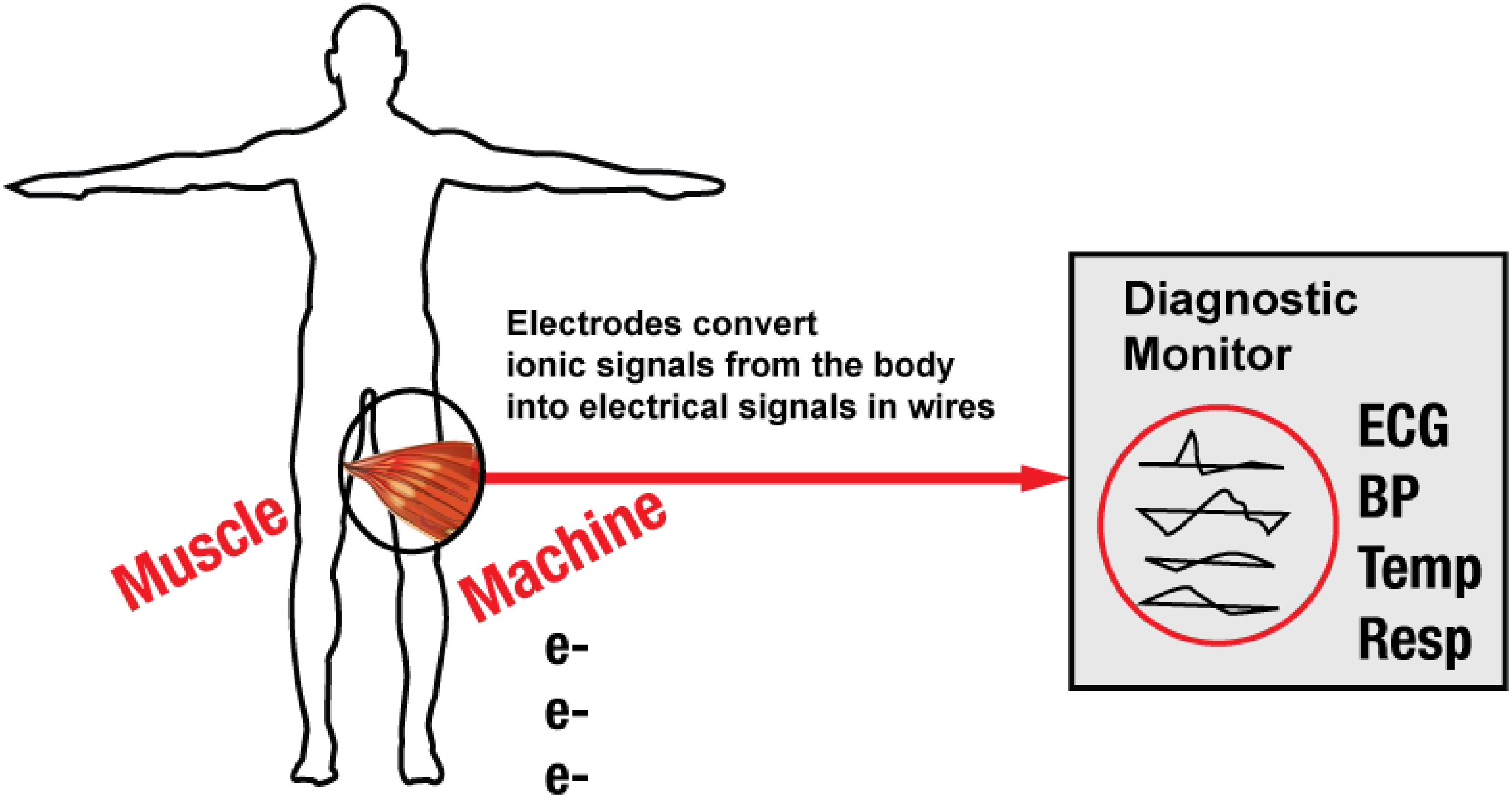
Extracellular medium/fluid

Ion channels (proteins)

- Resting ion chs.
- Active ion pumps
- **Gated ion chs.**

What is a Biopotential

Biopotentials from cells → electrodes

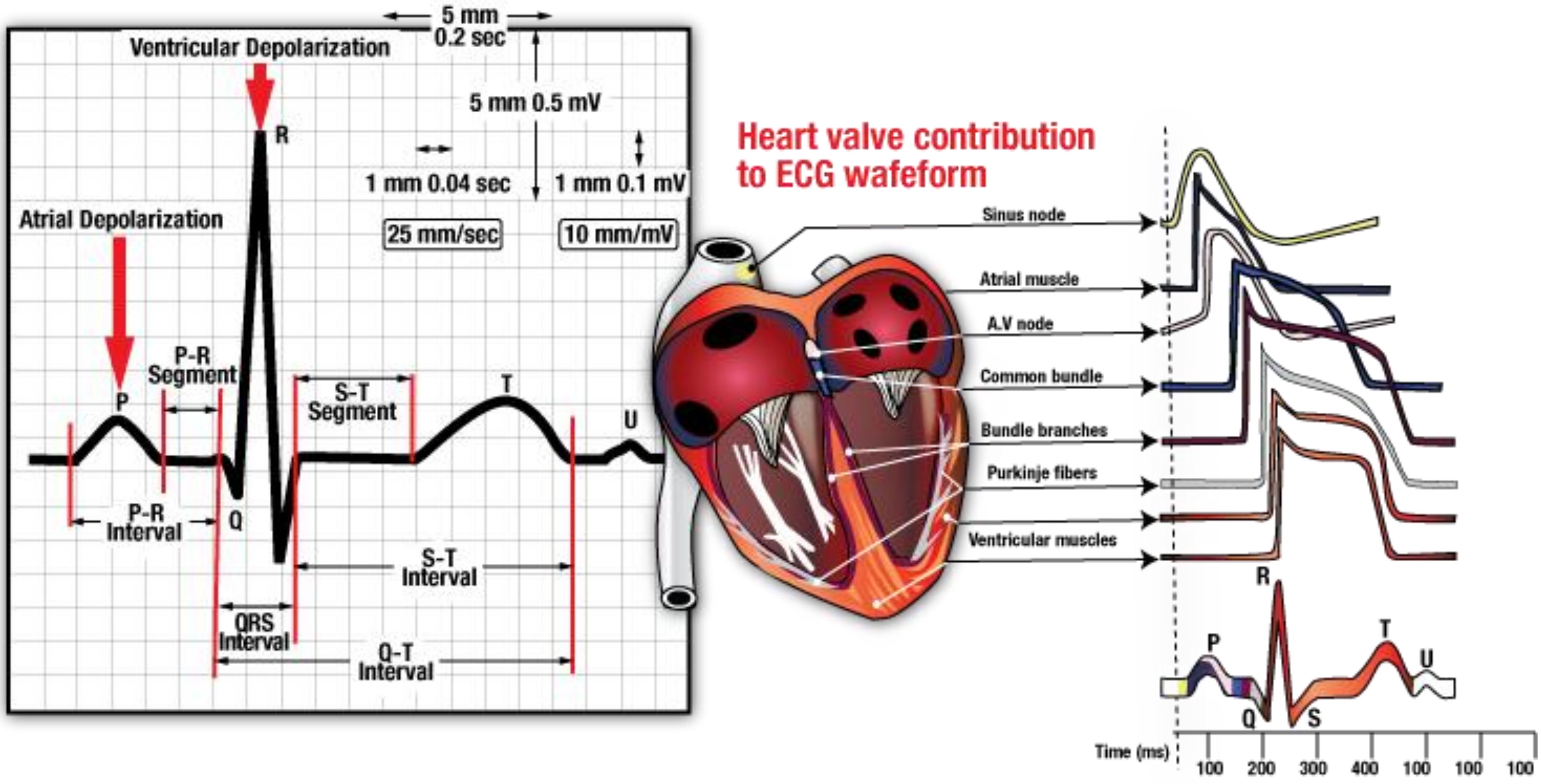


Electrodes convert ionic signals from the body into electrical signals in wires

What is ECG?

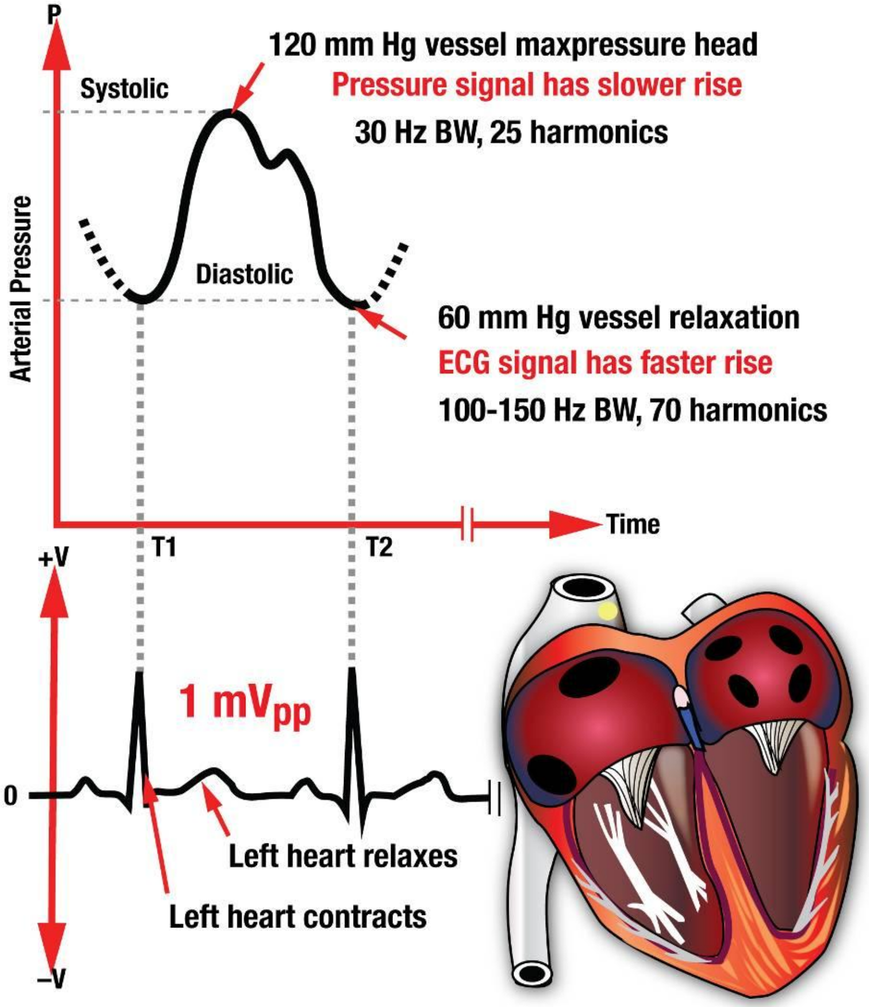
What is ECG?

A measure of the electrical activity of the heart



What is ECG?

ECG and blood pressure waves



120 mm Hg vessel maxpressure head
Pressure signal has slower rise
30 Hz BW, 25 harmonics

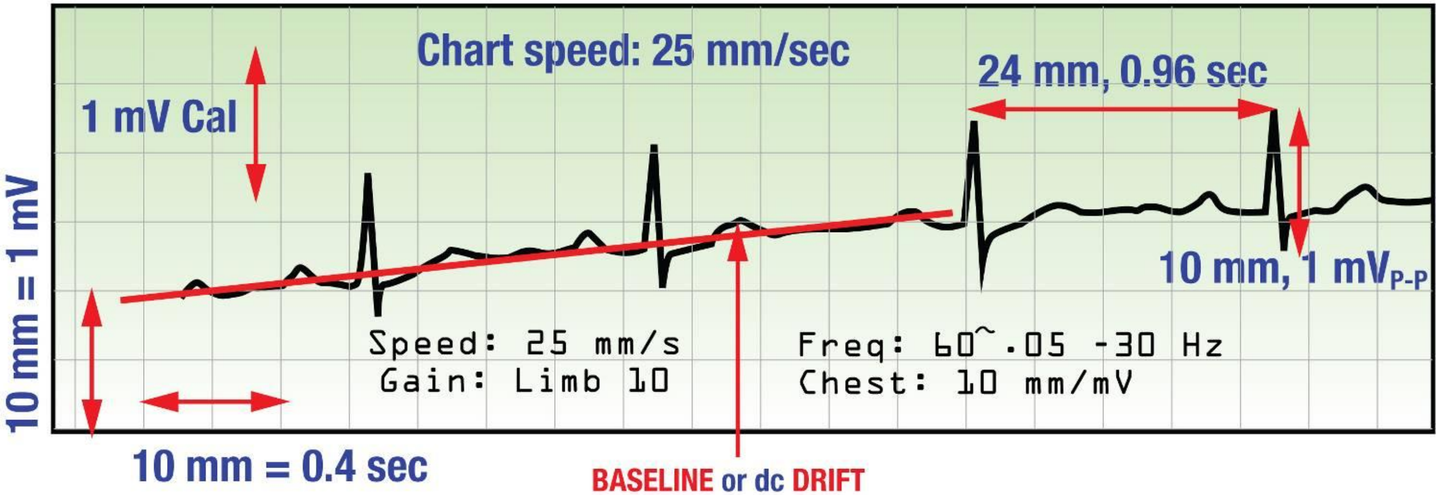
60 mm Hg vessel relaxation
ECG signal has faster rise
100-150 Hz BW, 70 harmonics

Left heart relaxes

Left heart contracts

What is ECG?

Actual ECG-normal



BASELINE or dc DRIFT
Charging time constant of high-pass filter and changing electrode half-cell potential

24 mm x 1 sec / 25mm = 0.96 sec / beat => 1 / 0.96 sec = 1.04 bps

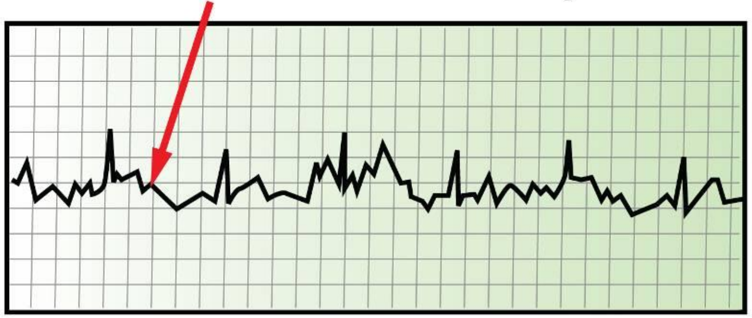
62 BPM at rest

What is ECG?

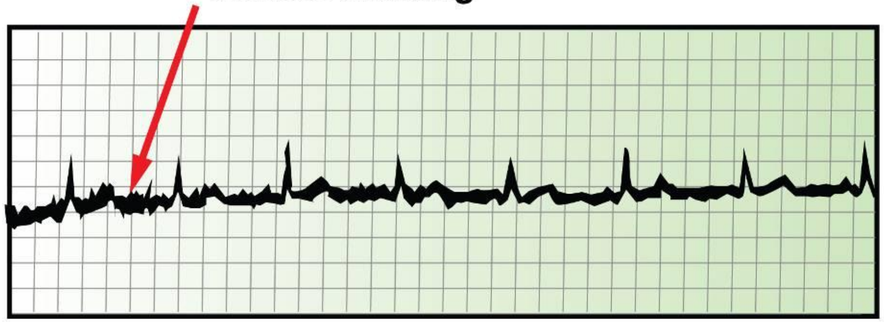
ECG irregular tracings due to external artifacts


50/60 Hz pick-up

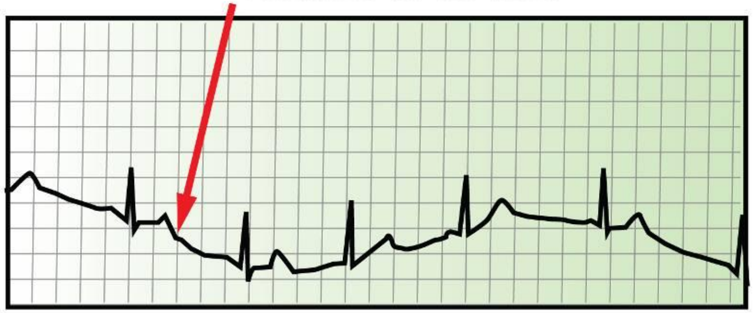
Alternating Current (AC) Interference


Baseline dc instability

Irregular Baseline


Muscle shaking

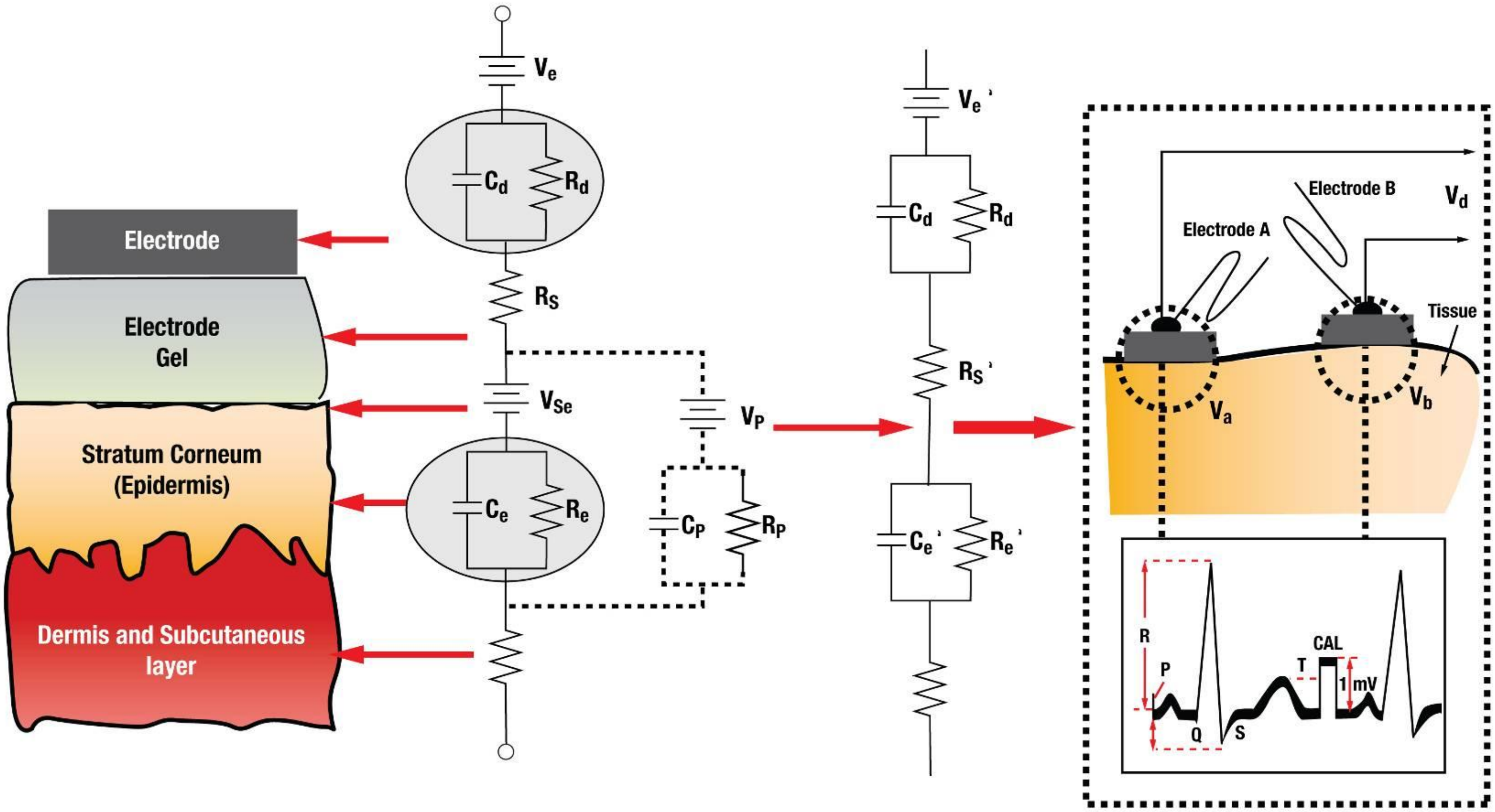
Somatic Tremor


Baseline or dc drift

Wandering Baseline

What is ECG?

Modeling the electrode interface

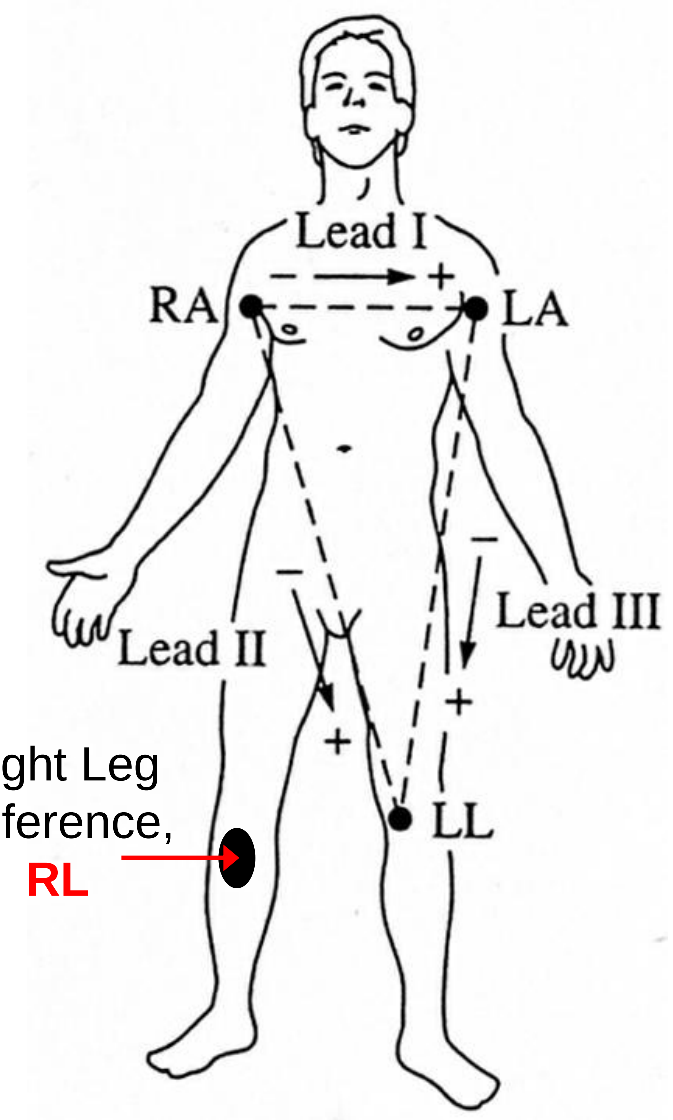


Electrical characteristics include a **DYNAMIC** resistance, capacitance, and offset voltage

Analog Lead Derivation

Analog Lead Derivation

ECG Einthoven Triangle, 1907



Right Leg
Reference,
RL

3 Body **Electrodes**,
3 Derived **Leads** = I, II, III

$$\textbf{LEAD I} = V_{LA\text{-}RL} - V_{RA\text{-}RL}$$

$$\textbf{LEAD II} = V_{LL\text{-}RL} - V_{RA\text{-}RL}$$

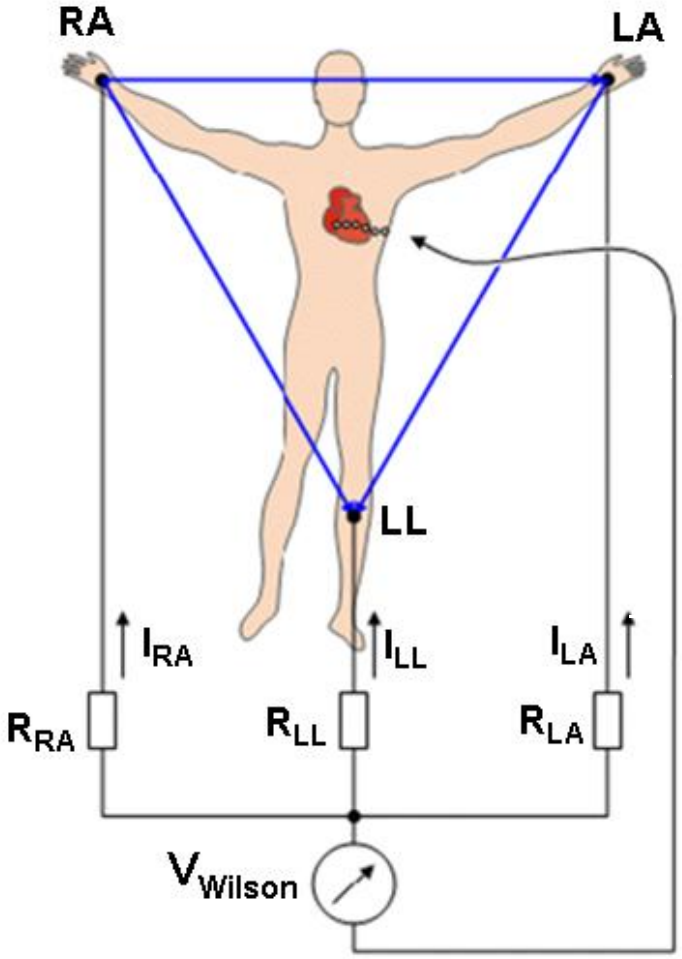
$$\textbf{LEAD III} = V_{LL\text{-}RL} - V_{LA\text{-}RL}$$

Einthoven's Law

In electrocardiogram at any given instant the potential of any wave in Lead II is equal to the ***sum*** of the potentials in Lead I and III.

Analog Lead Derivation

The **Wilson Central** (WCT) Provides Chest Lead Reference at Center of Einthoven Triangle



Assuming:

$$R_{RA} = R_{LA} = R_{LL}$$

Then:

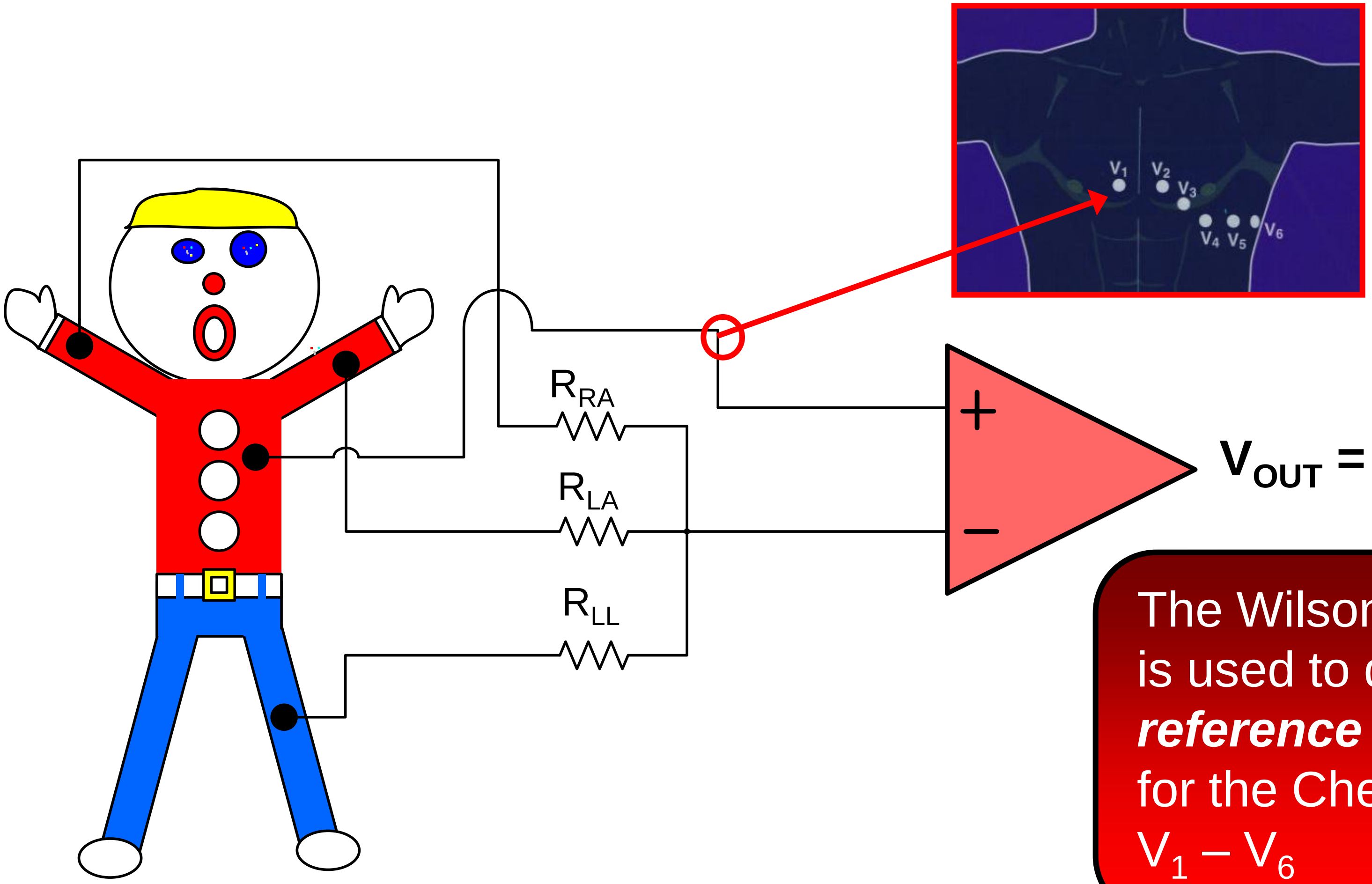
$$3 \cdot \frac{\Phi_{WCT}}{R_{RA}} = \frac{\Phi_{RA} + \Phi_{LA} + \Phi_{LL}}{R_{RA}}$$

$$\Phi_{WCT} = \frac{\Phi_{RA} + \Phi_{LA} + \Phi_{LL}}{3}$$

*Drawing Taken From Bioelectromagnetism, Jaako Malmivuo and Robert Plonsey

Analog Lead Derivation

The Wilson Central is the AVERAGE potential between RA, LA, and LL

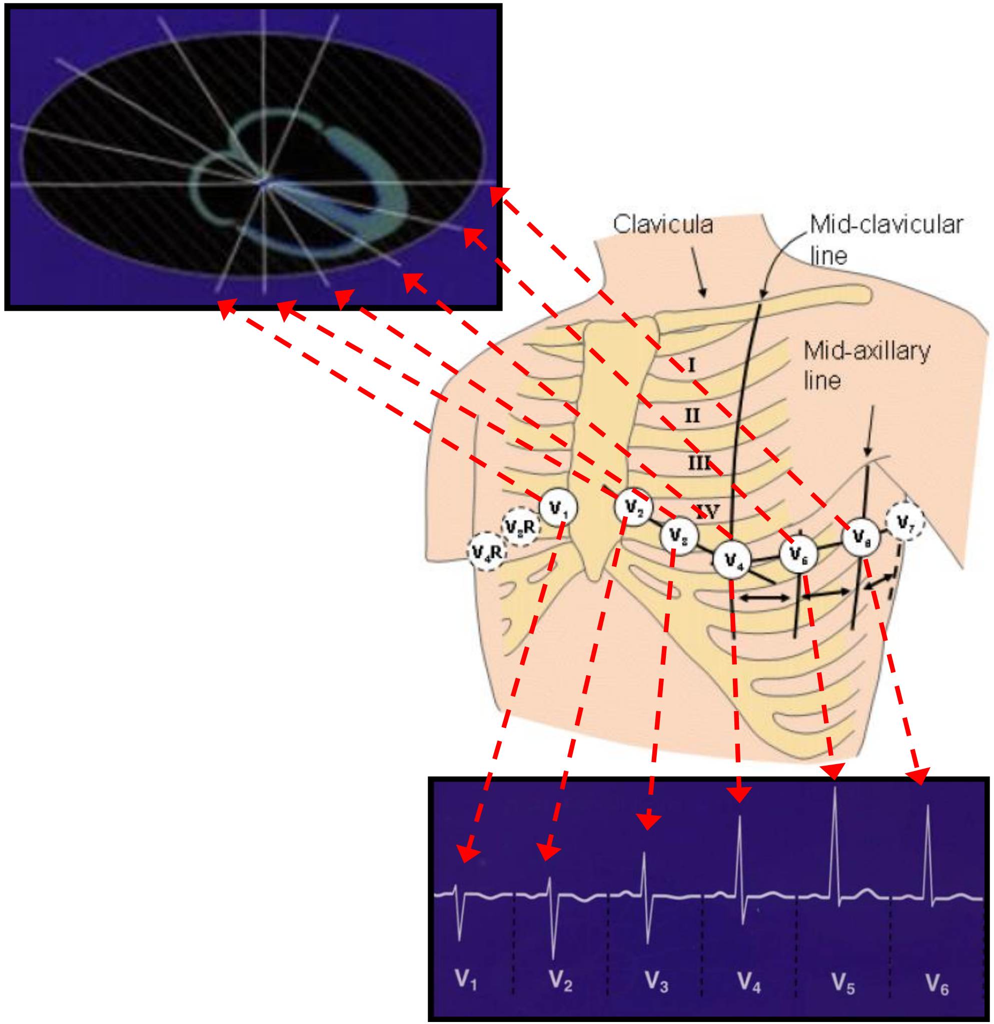


$V_{OUT} = V_{(1\text{-}6)} - V_{WCT}$

The Wilson Central is used to derive a ***reference*** potential for the Chest Leads, $V_1 – V_6$

Analog Lead Derivation

Chest Lead Signals Provide Different Information at Different Cross-Sectional Angles



Different Chest Leads Provide:

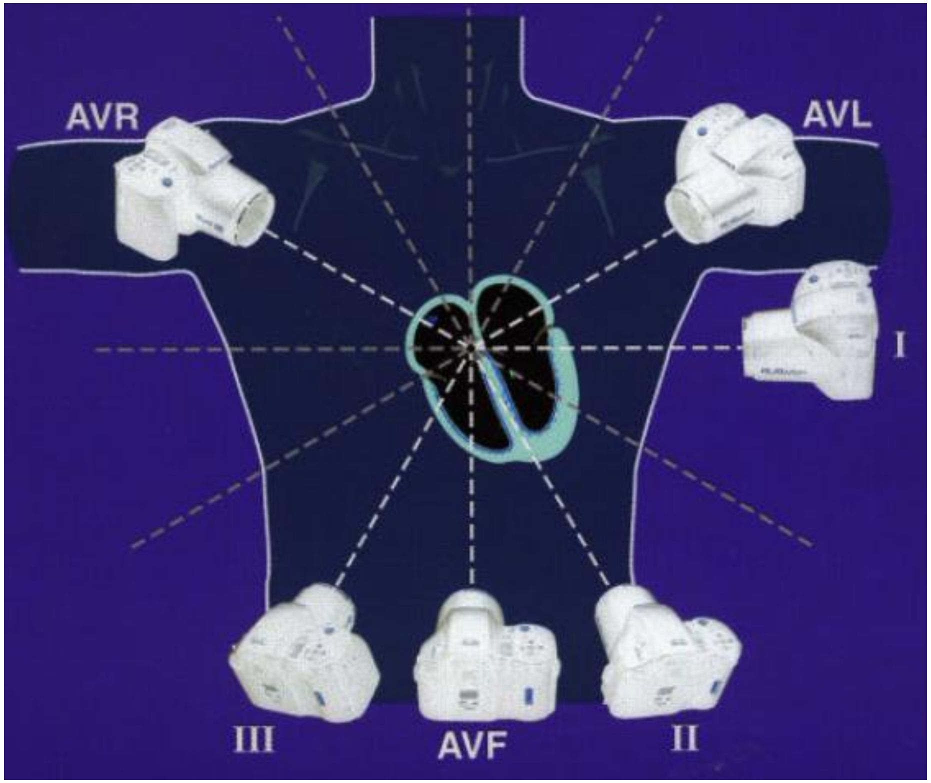
*Unique ECG Signature

*Enhanced Pattern Recognition

*Isoelectric Point @ V_3-V_4

Analog Lead Derivation

Augmented Leads Derived via WCT to Provide Enhanced Vector Information



✓Each lead provides ***unique*** information about the ECG Output Signal

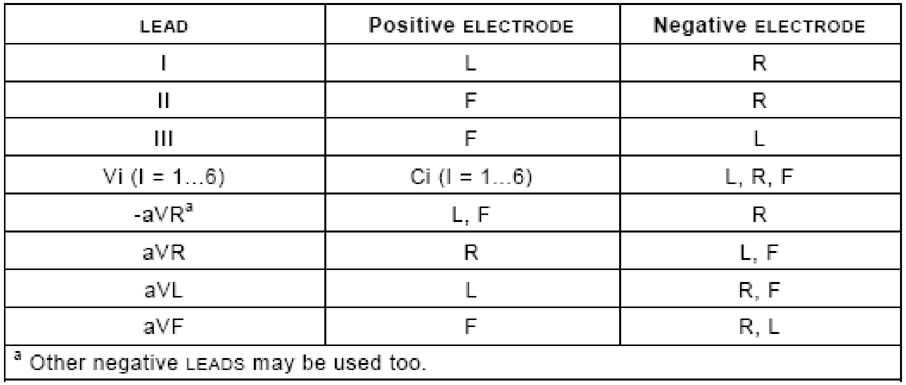
✓Multiple Angles Give a Better Than 2-D ***Picture*** of the ECG Output

✓AVR, AVL, AVF derived via midpoint of 2 limbs (resistor divider) with Respect to 3rd limb

Analog Lead Derivation

IEC60601-2-51—Diagnostic

Table 109 – Connection of ELECTRODES for a particular LEAD

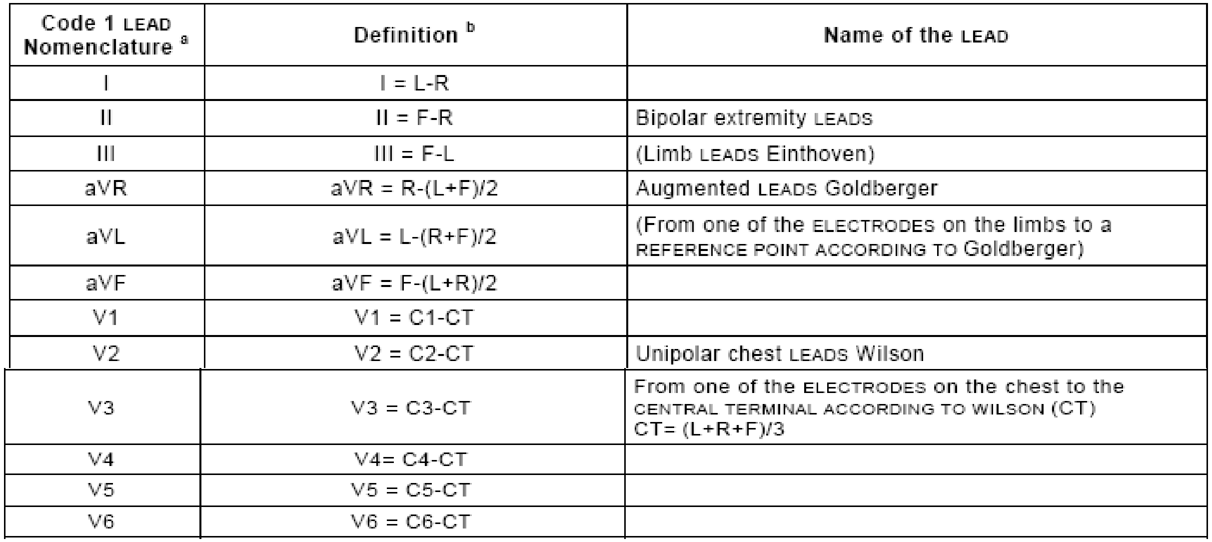


LEAD	Positive ELECTRODE	Negative ELECTRODE
I	L	R
II	F	R
III	F	L
Vi (I = 1…6)	Ci (I = 1…6)	L, R, F
-aVR[a]	L, F	R
aVR	R	L, F
aVL	L	R, F
aVF	F	R, L

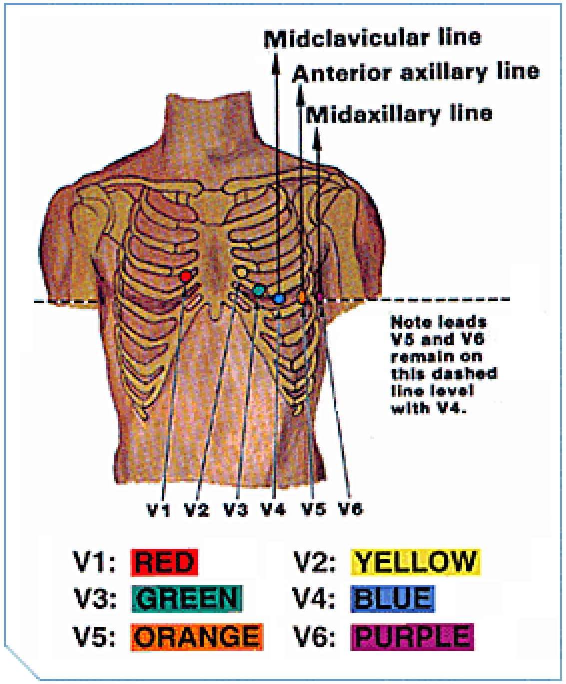
[a] Other negative LEADS may be used too.

Standards	Electrodes Needed
1 Lead	LA, RA
3 Lead	LA, RA, LL
6 Leads	LA, RA, LL
12 Leads	LA, RA, LL, V1-6

Table 110 – LEADS and their identification (nomenclature and definition)

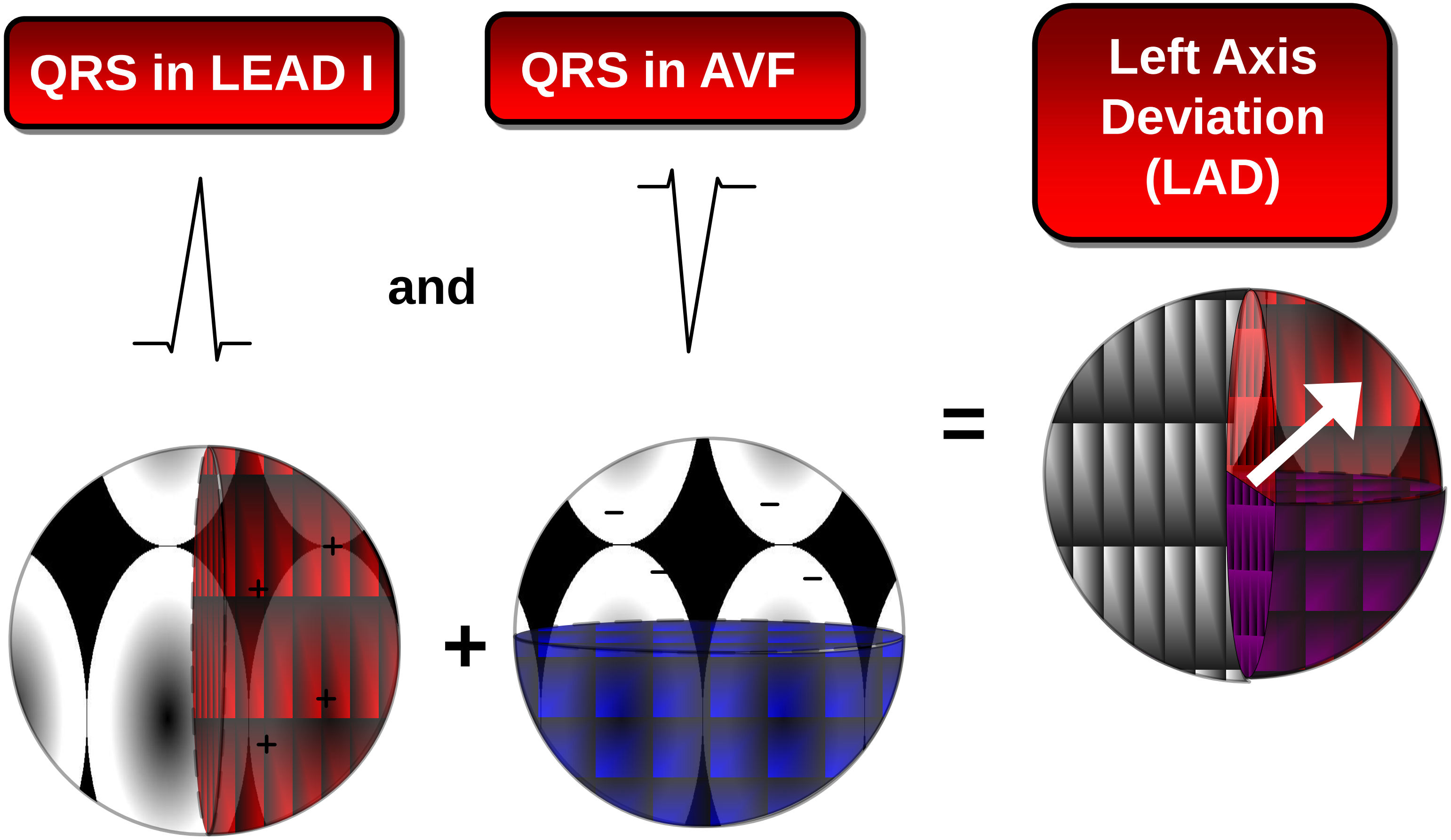


Code 1 LEAD Nomenclature [a]	Definition [b]	Name of the LEAD
I	I = L-R	
II	II = F-R	Bipolar extremity LEADS
III	III = F-L	(Limb LEADS Einthoven)
aVR	aVR = R-(L+F)/2	Augmented LEADS Goldberger
aVL	aVL = L-(R+F)/2	(From one of the ELECTRODES on the limbs to a REFERENCE POINT ACCORDING TO Goldberger)
aVF	aVF = F-(L+R)/2	
V1	V1 = C1-CT	
V2	V2 = C2-CT	Unipolar chest LEADS Wilson
V3	V3 = C3-CT	From one of the ELECTRODES on the chest to the CENTRAL TERMINAL ACCORDING TO WILSON (CT) CT= (L+R+F)/3
V4	V4= C4-CT	
V5	V5 = C5-CT	
V6	V6 = C6-CT	



Analog Lead Derivation

Different Lead Combinations Reveal Axis Deviation

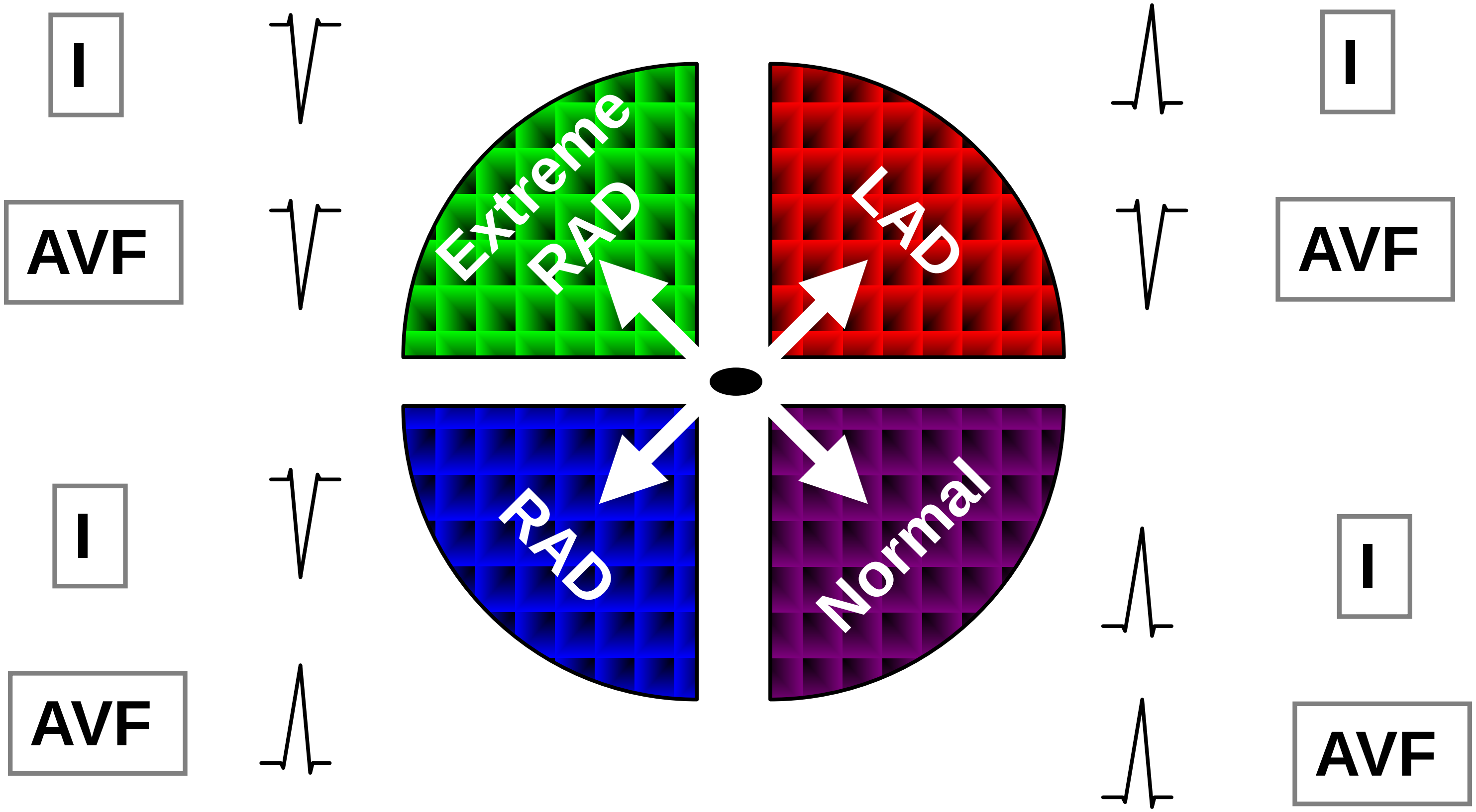


QRS in LEAD I and **QRS in AVF** = **Left Axis Deviation (LAD)**

Mean QRS Vector Points Toward Area of Infarction (Damage)

Analog Lead Derivation

Different Lead Combinations Reveal Axis Deviation



ECG Input Filtering, Defibrillation Protection, and Isolation

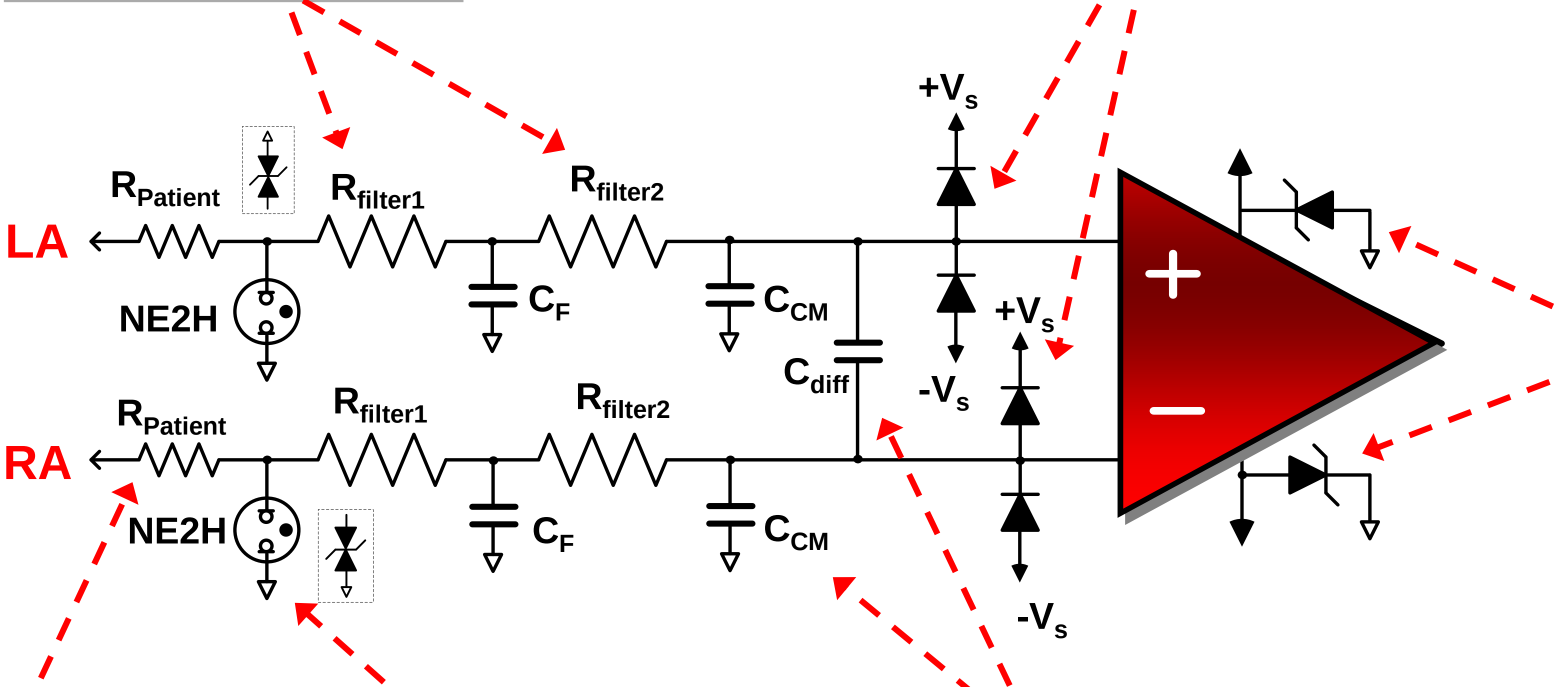
ECG Input Filtering and Protection

Example: LEAD I Protection with Input Filtering

Series Resistance Limits Input Current

C_F, C_{cm}, and C_{diff} + R_{filter} form LPF

Protection Diodes



Zener Diode

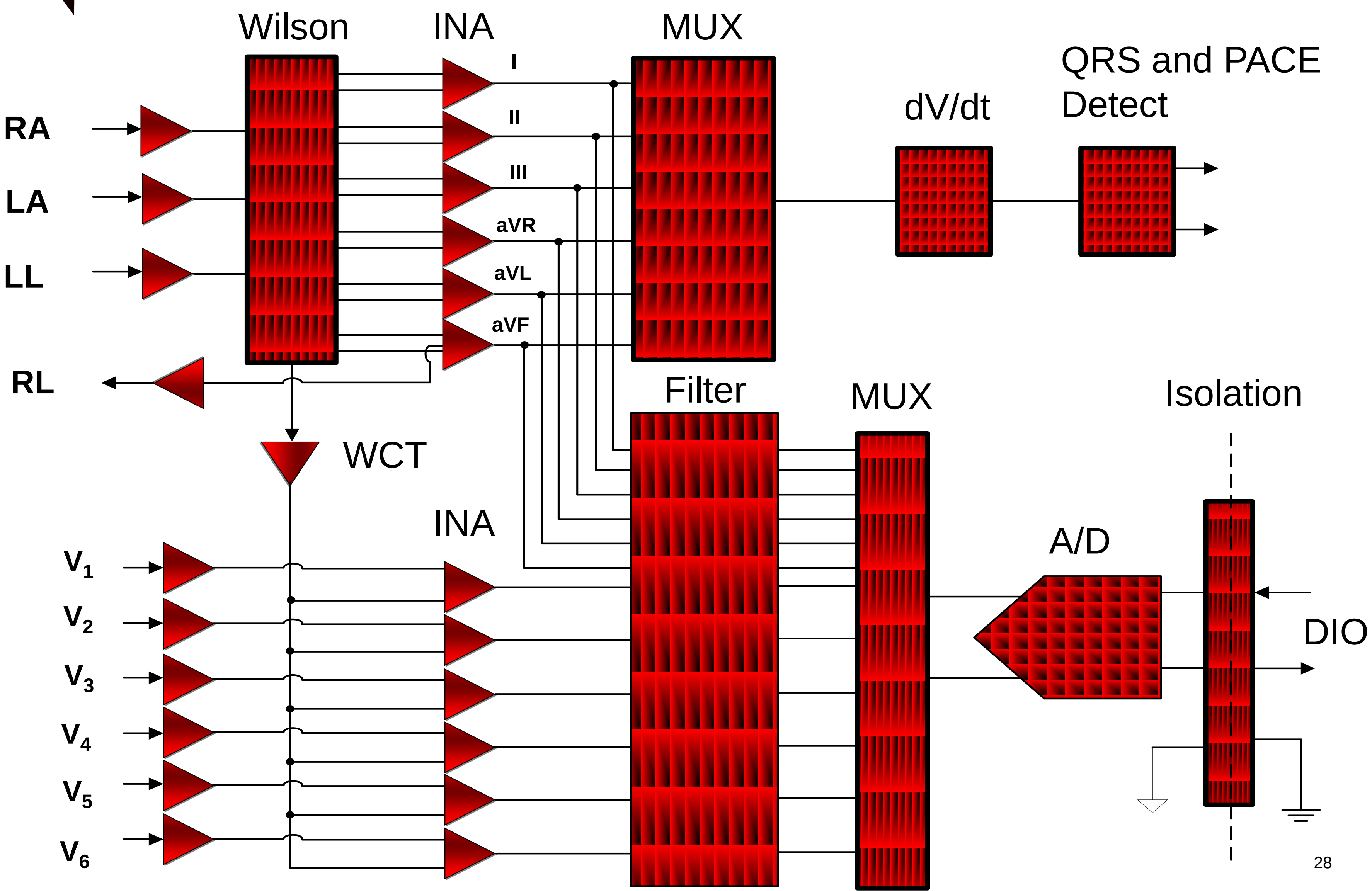
Patient Protection 10-20k ohms

Ne2H Lamps/TVS Protect Against Defibrillation Voltages

$C_{diff} = 10 \times C_{cm}$

System Block Diagram

System Block Diagram



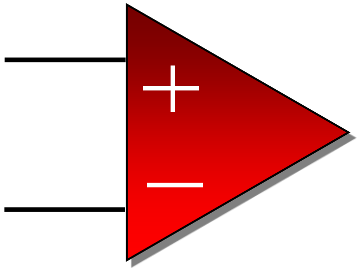
The INA Front End

The INA Front End

Key Features of the INA Front End

Important

- ✓Input Bias Current
- ✓Input Impedance
- ✓Input Current Noise
- ✓Input Voltage Noise
- ✓Power Consumption
- ✓DC/AC CMRR



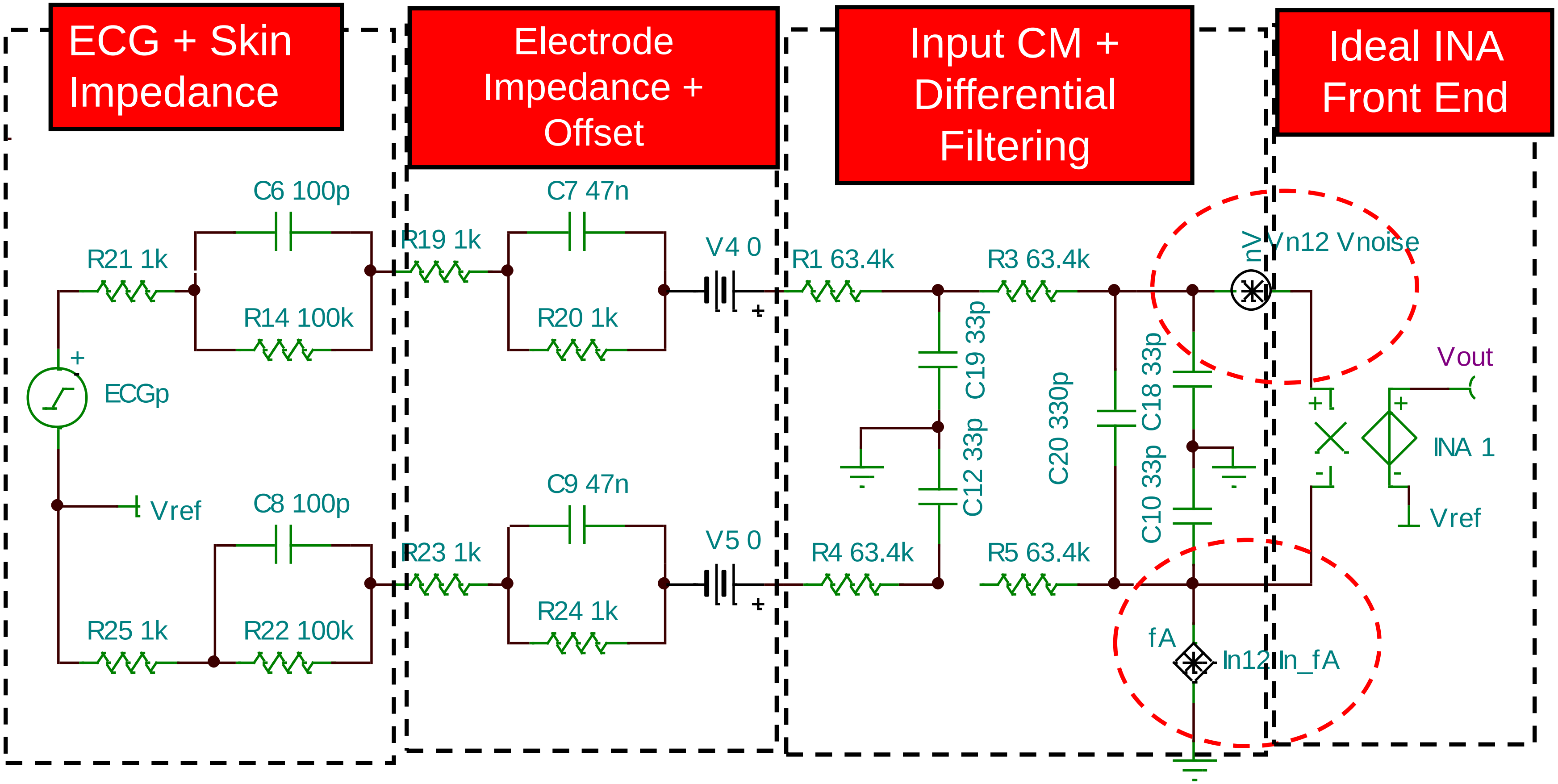
Less Important

- ✓Input Offset Voltage
- ✓Input Offset Voltage Drift
- ✓Gain Error
- ✓Nonlinearity
- ✓PSRR

*DC Errors such as VOS are swamped out by the Offsets Introduced by the **Skin-Electrode** Contacts

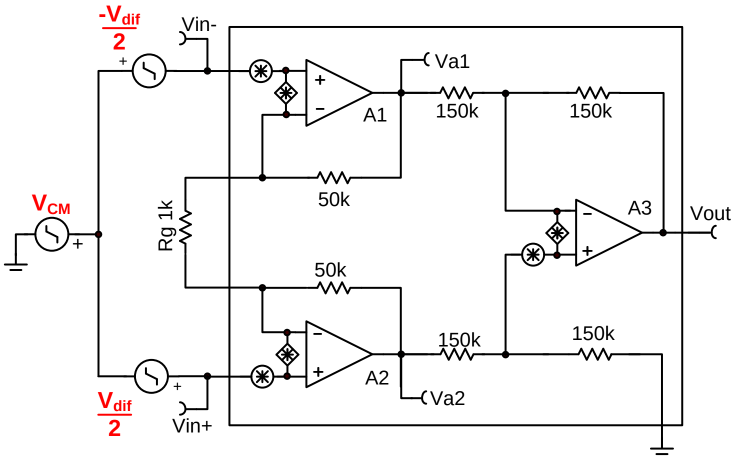
The INA Front End

Ideal Simulation Circuit with Current and Voltage Noise Sources

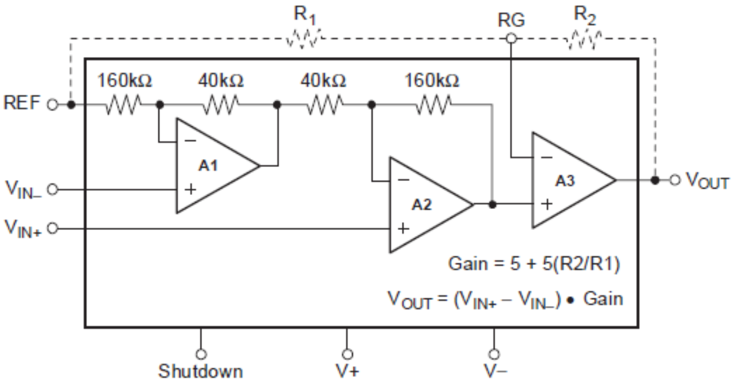


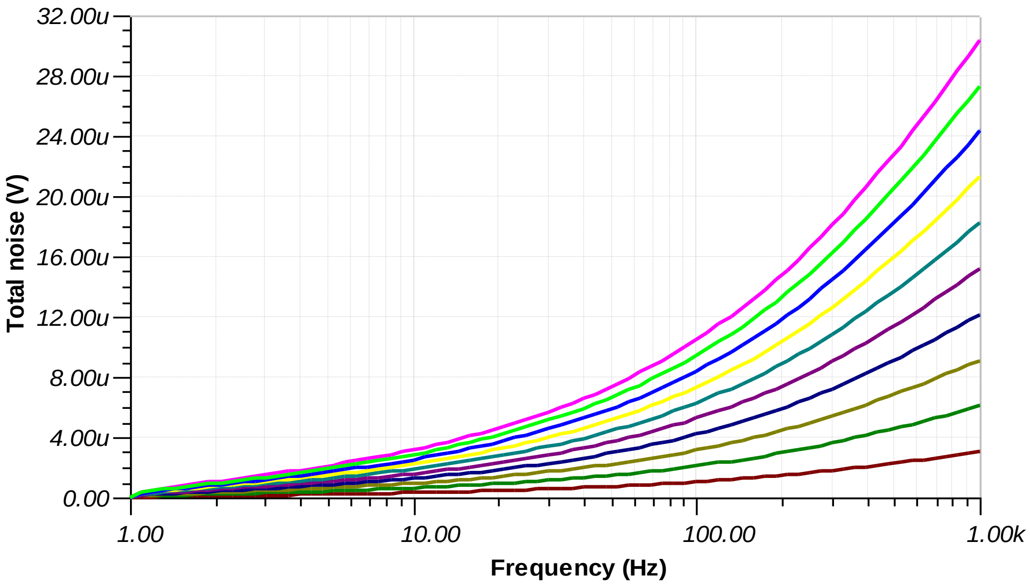
The INA Front End

Simulation Showing Output-Referred Total RMS Noise vs. Bandwidth (G = 1-10)



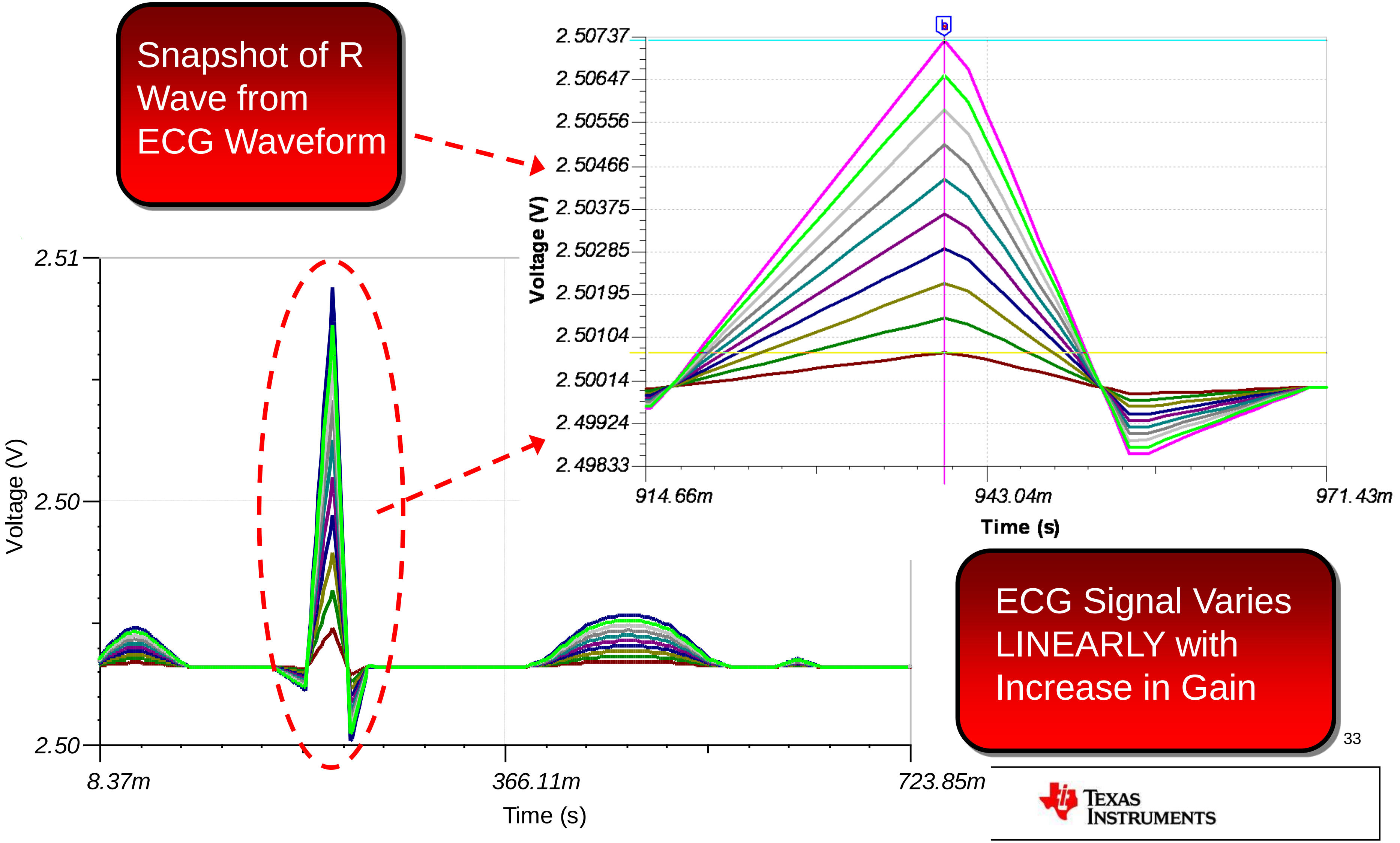
Noise may **NOT** necessarily increase linearly with gain (INA or PGA topology dependent)





The INA Front End

TINA Simulations Showing Output-Referred ECG Signal (G = 1-10)

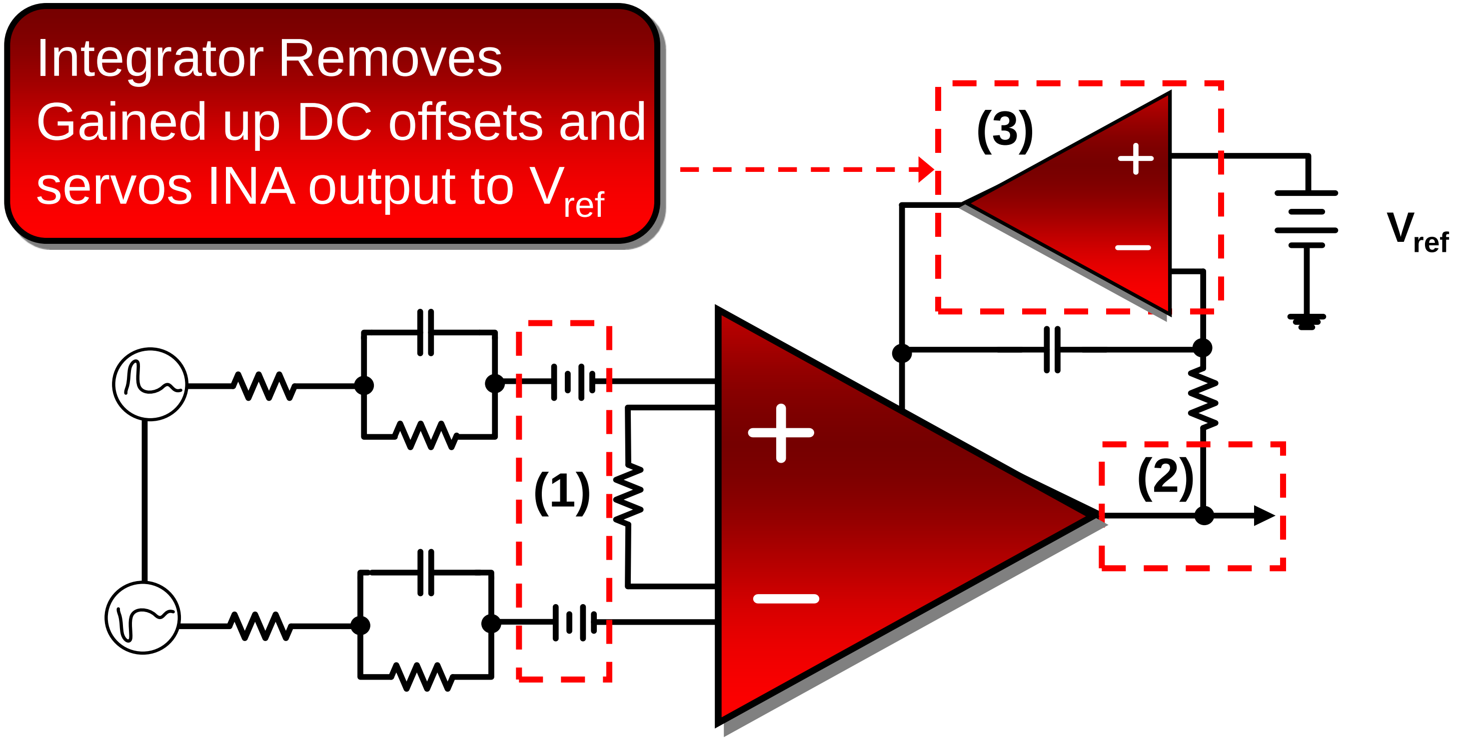


Snapshot of R Wave from ECG Waveform

ECG Signal Varies LINEARLY with Increase in Gain

The INA Front End

What is the MAX gain on the INA When Using a DC Removal Circuit?



Integrator Removes Gained up DC offsets and servos INA output to V_{ref}

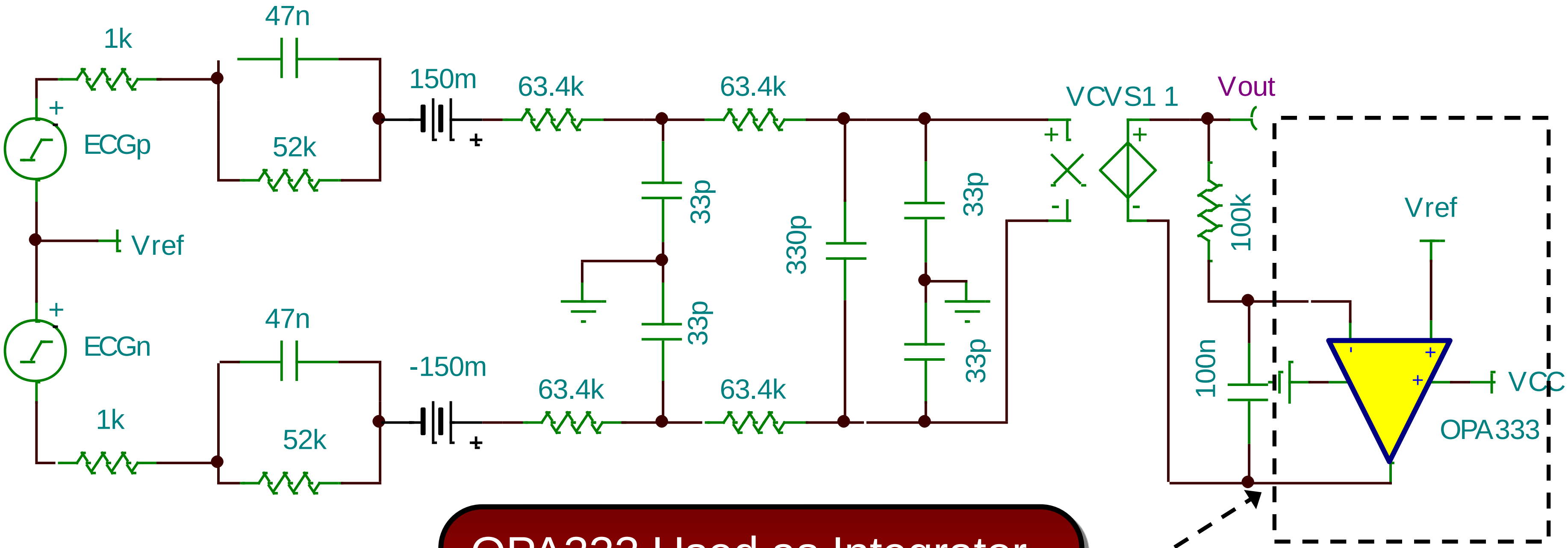
(1) Electrode Offset MAX = **+/- 300mV**

(2) Swing of INA = **V(+) – 50mV**

(3) Integrator Compliance = **(ECGp + ECGn +VOS + $\mathbf{VOS_{electrode}}$)* Gain < $\mathbf{V_{CC}}$ - $\mathbf{V_{ref}}$**

The INA Front End

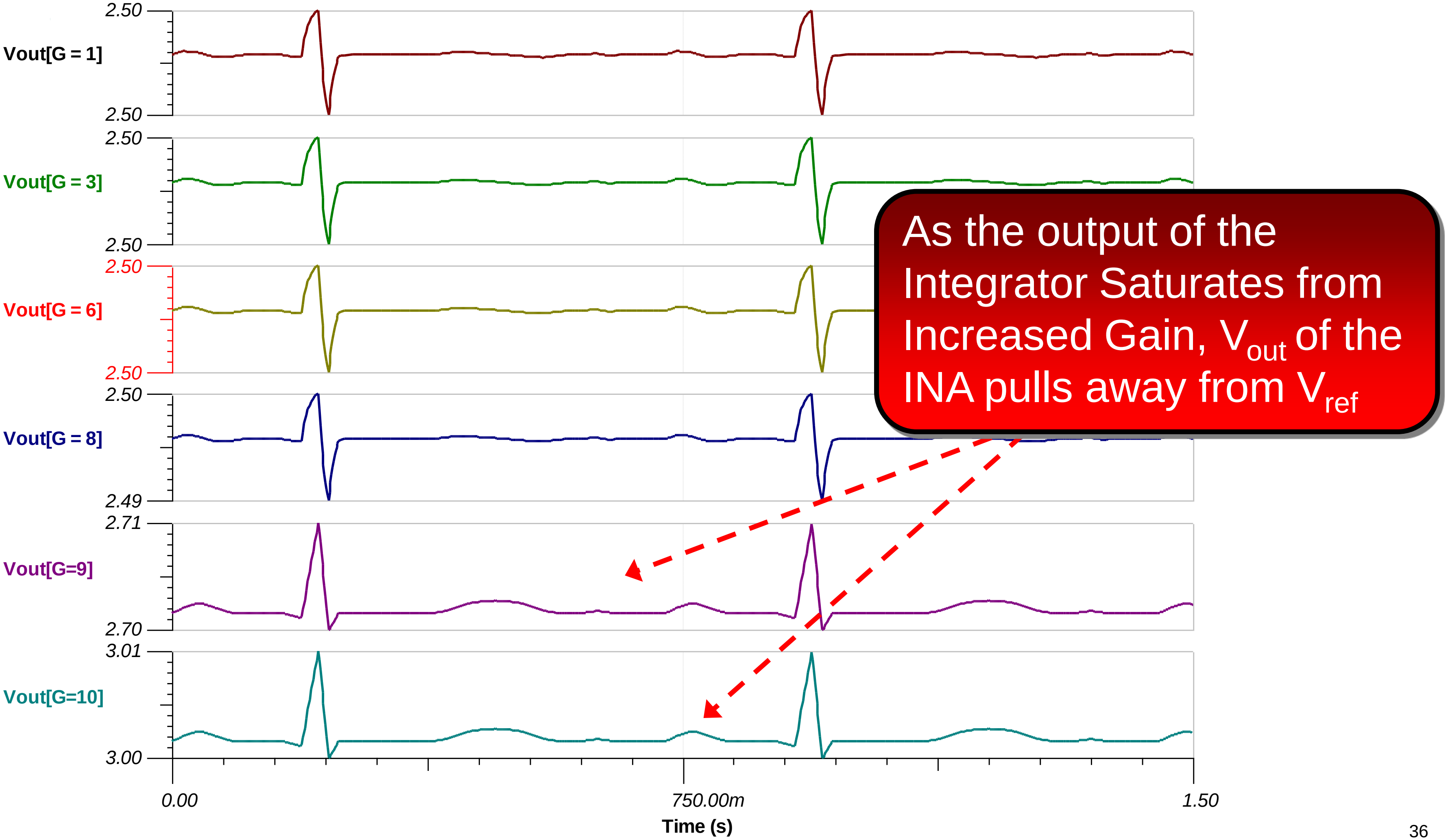
Simulation Circuit with Ideal INA and V_{ref} = 2.5V as Integrator Input



OPA333 Used as Integrator to Remove DC and Simulate Real Response During Saturation

The INA Front End

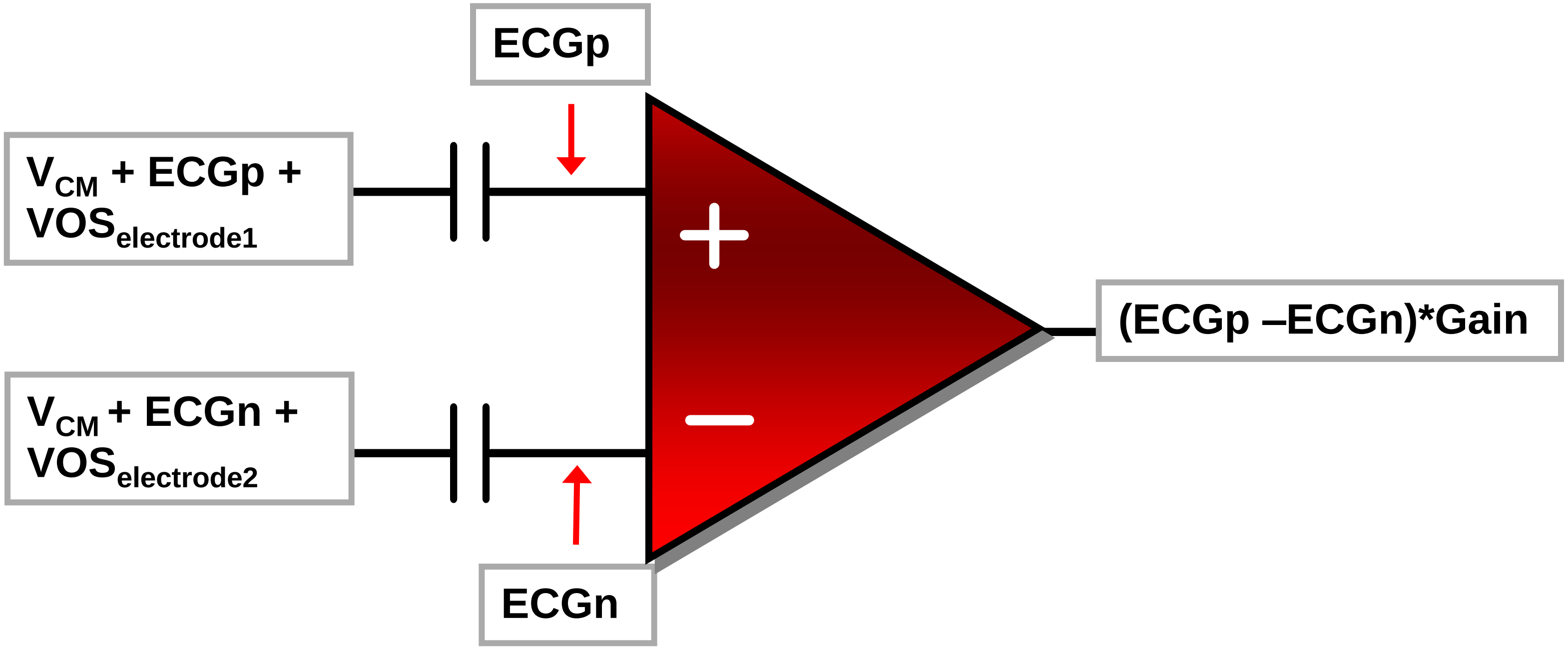
ECG + Integrator Output of INA vs. Gain for V_{ref} = 2.5V



As the output of the Integrator Saturates from Increased Gain, V_{out} of the INA pulls away from V_{ref}

The INA Front End

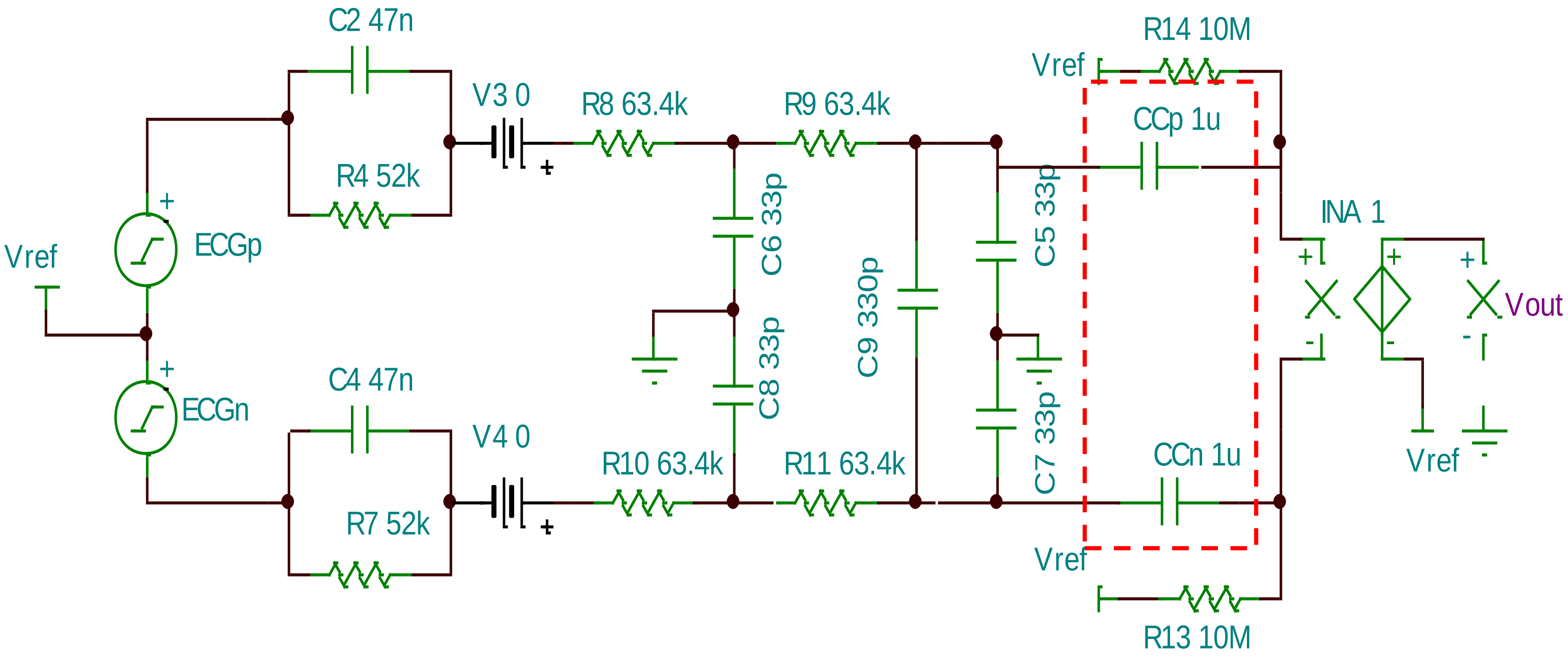
If it is Advantageous to Maximize Gain with a Low Noise INA up Front, Why not AC Couple?



*AC Coupling Removes Electrode Offsets so that Higher Gain Can Be Used for Potentially Better SNR in the Signal Chain Path

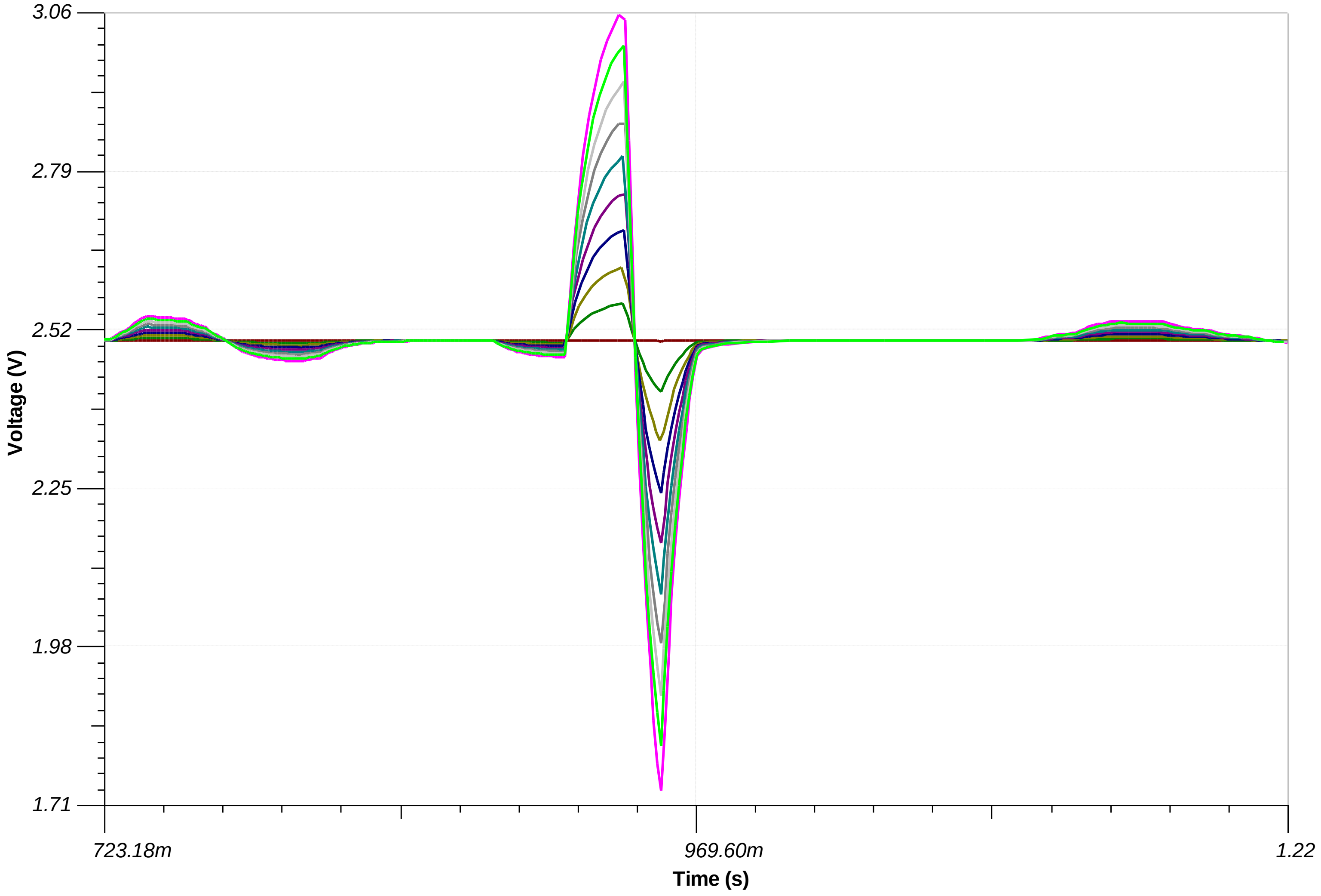
The INA Front End

TINA Simulation Circuit to Show AC-Coupled INA Gain Sweep



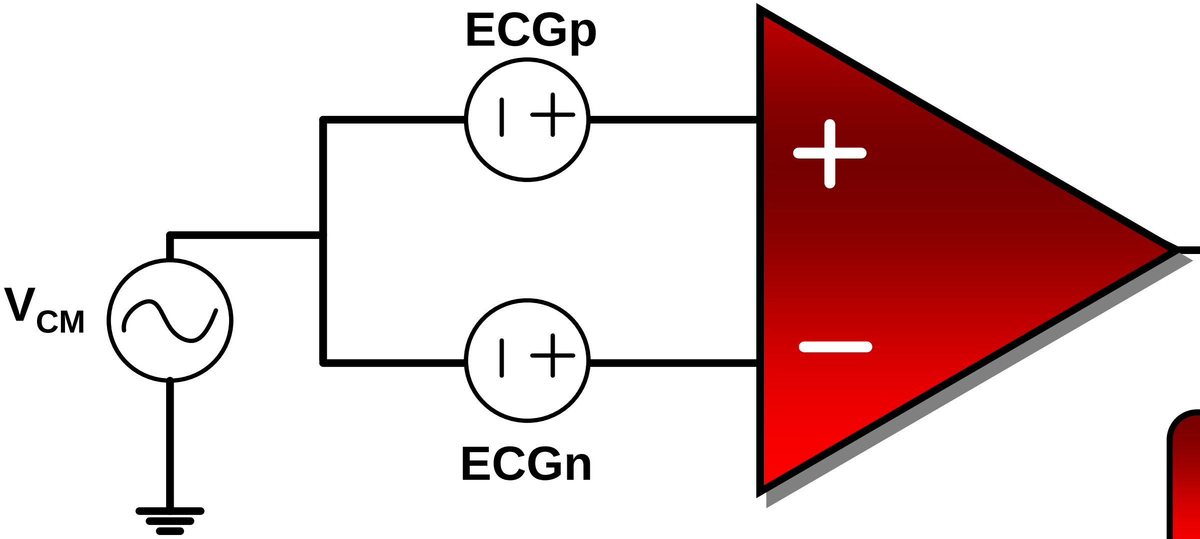
The INA Front End

INA Gain = 1-1000 with V_{REF} = 2.5V AC Coupled



The INA Front End

What is CMRR? Why is it Important in ECG?



$$\Delta V_{out} = \Delta V_{OS} \cdot Gain$$

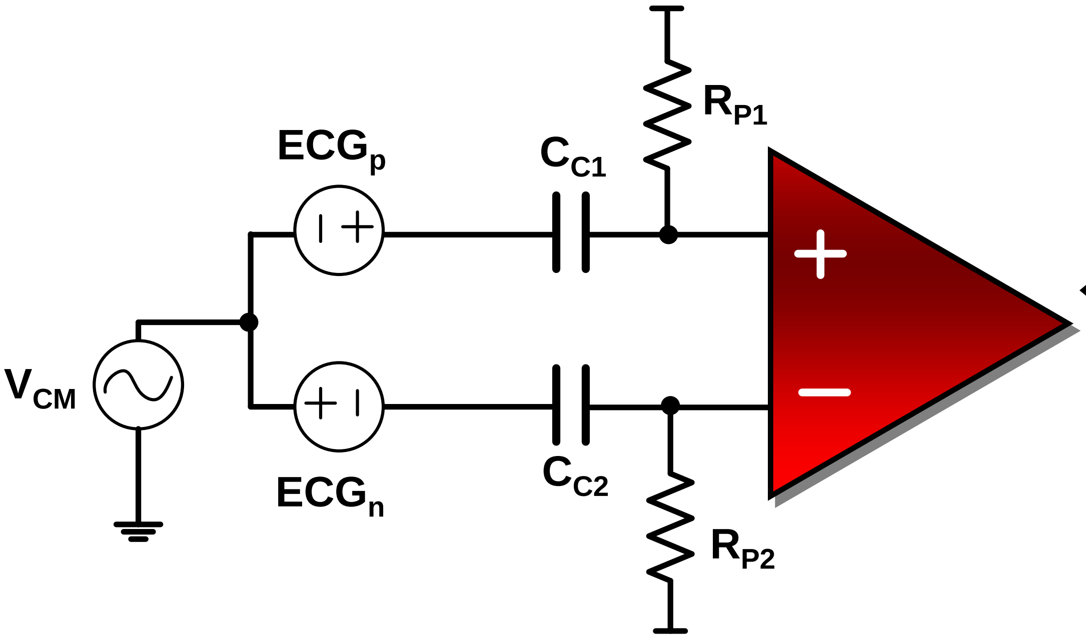
Lower **CMRR** = More Unwanted Output Signal

$$CMRR = -20\log_{10}\left(\frac{\Delta V_{CM}}{\Delta V_{OS}}\right) = -20\left(\frac{\Delta V_{CM} \cdot Gain}{\Delta V_{out}}\right)$$

$$V_{out} = \left(ECGp - ECGn + VOS_{electrode} + VOS_{OPA} + \frac{\Delta V_{CM}}{10^{\frac{-CMRR}{20}}}\right) \cdot Gain$$

The INA Front End

What is CMRR? Why is it Important in ECG?



The INA **includes** the R and C and must be considered in the overall CMRR Analysis

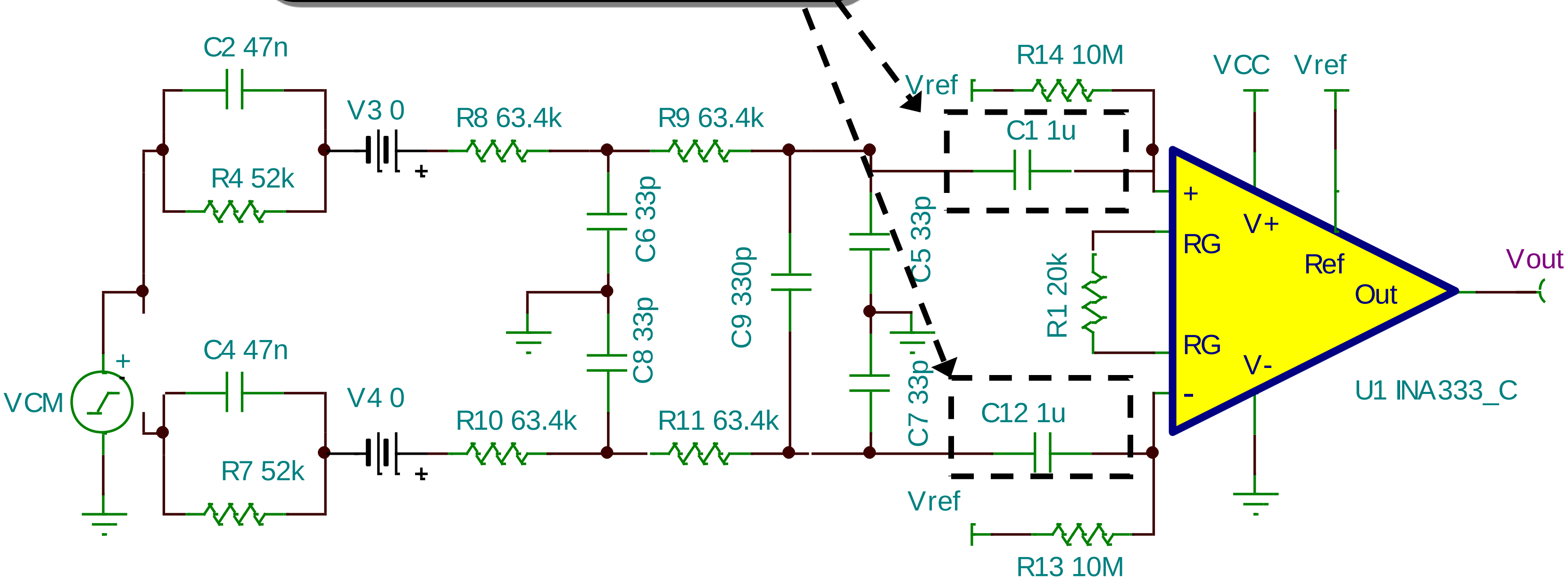
- Mismatch in R_p and C_c causes a **differential** signal error
- Even a 1% tolerance on Cc cause a **20dB** attenuation in CMRR

$$V_{diff\_actual} = V_{inp} \cdot \sqrt{\frac{R_{p1}^2}{R_{p1}^2 + \left(\frac{j}{\boxed{\omega C_{c1}}}\right)^2}} - V_{inn} \cdot \sqrt{\frac{R_{p2}^2}{R_{p2}^2 + \left(\frac{j}{\boxed{\omega C_{c2}}}\right)^2}}$$

The INA Front End

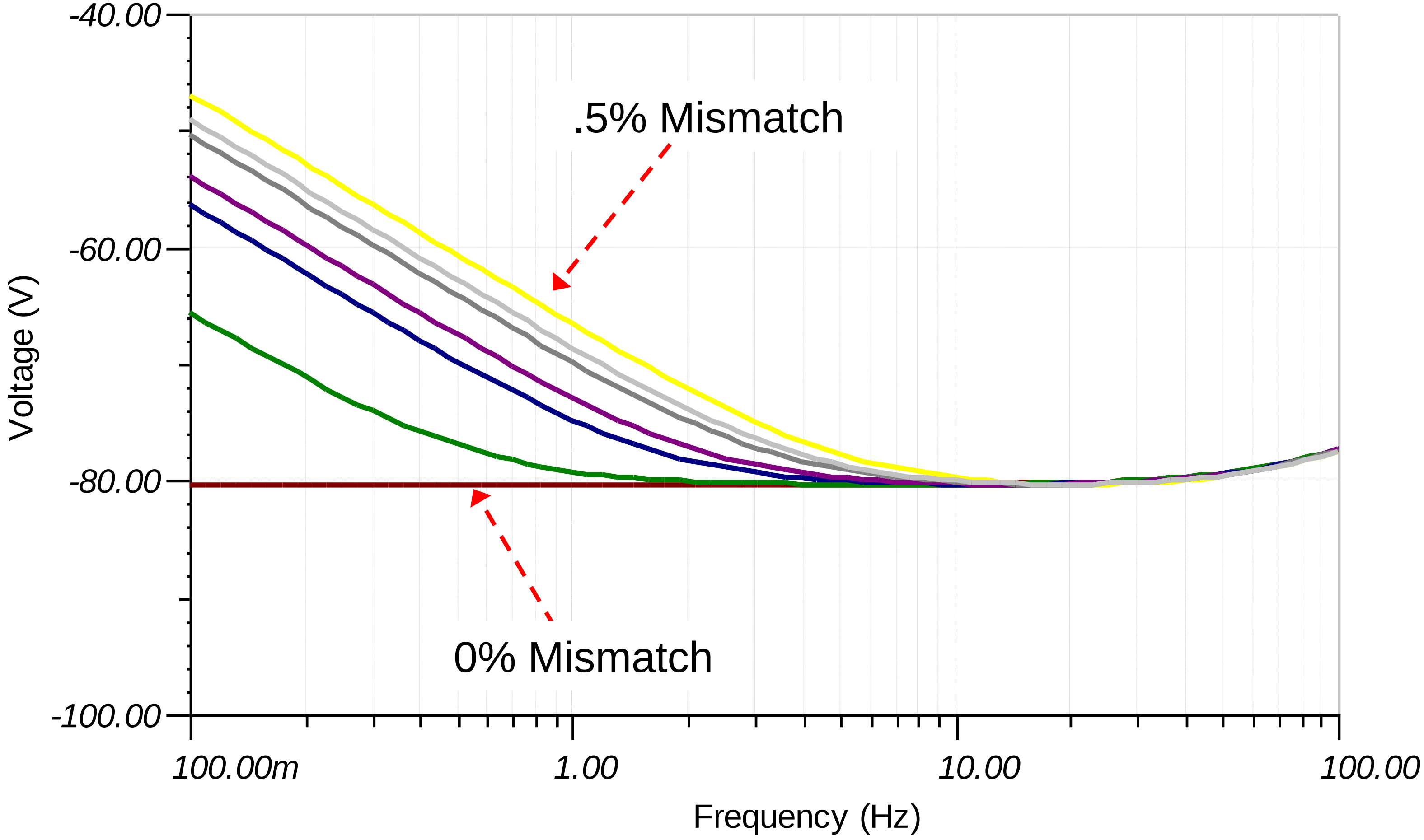
50/60Hz Common Mode Simulation Circuit with 1µF Coupling Capacitors Mismatched

CMR TINA Circuit Test By Sweep of ***Mismatch*** of Input Coupling Capacitors



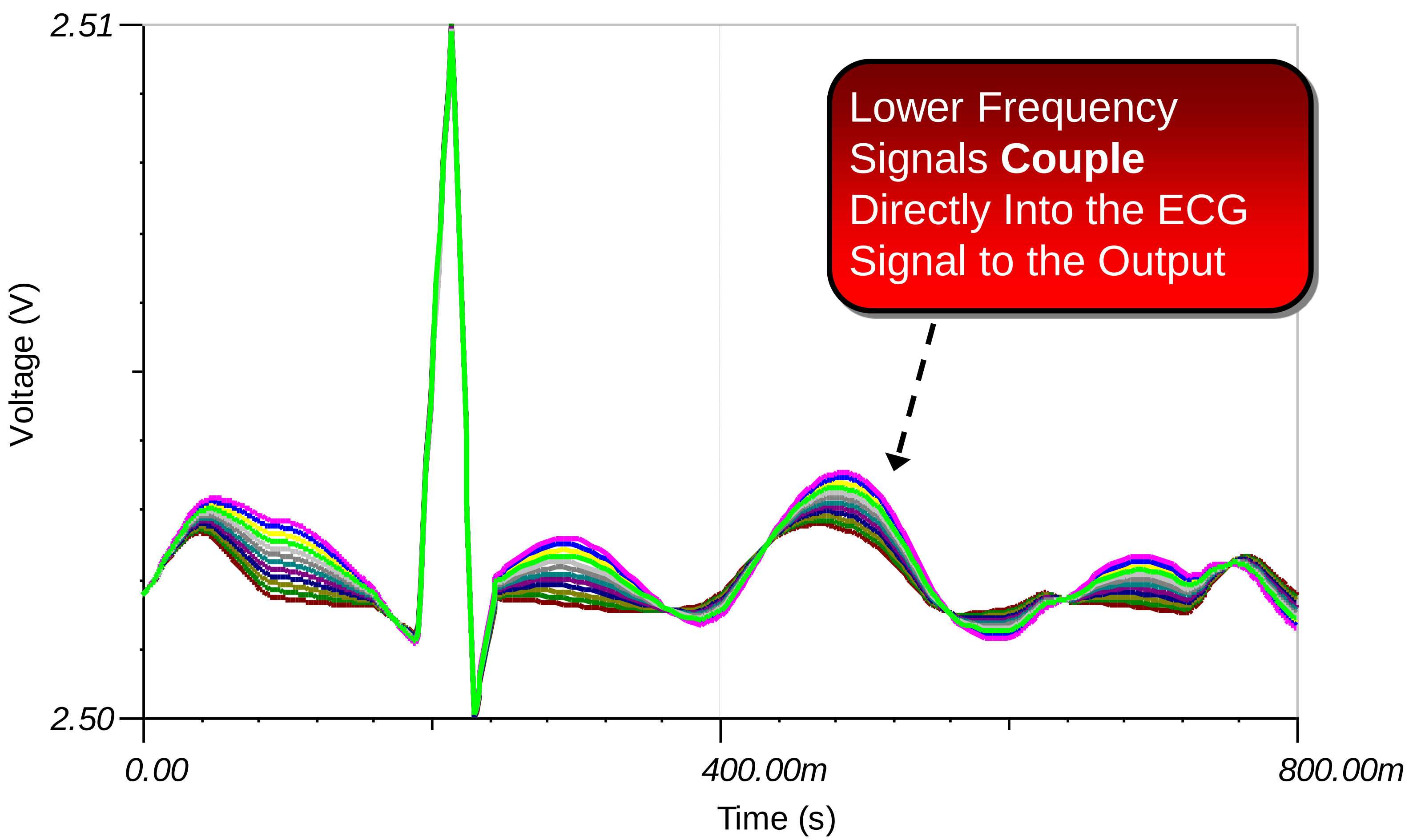
The INA Front End

Plot of CMRR vs. Frequency for .01 - .5% Coupling Capacitor Mismatch



The INA Front End

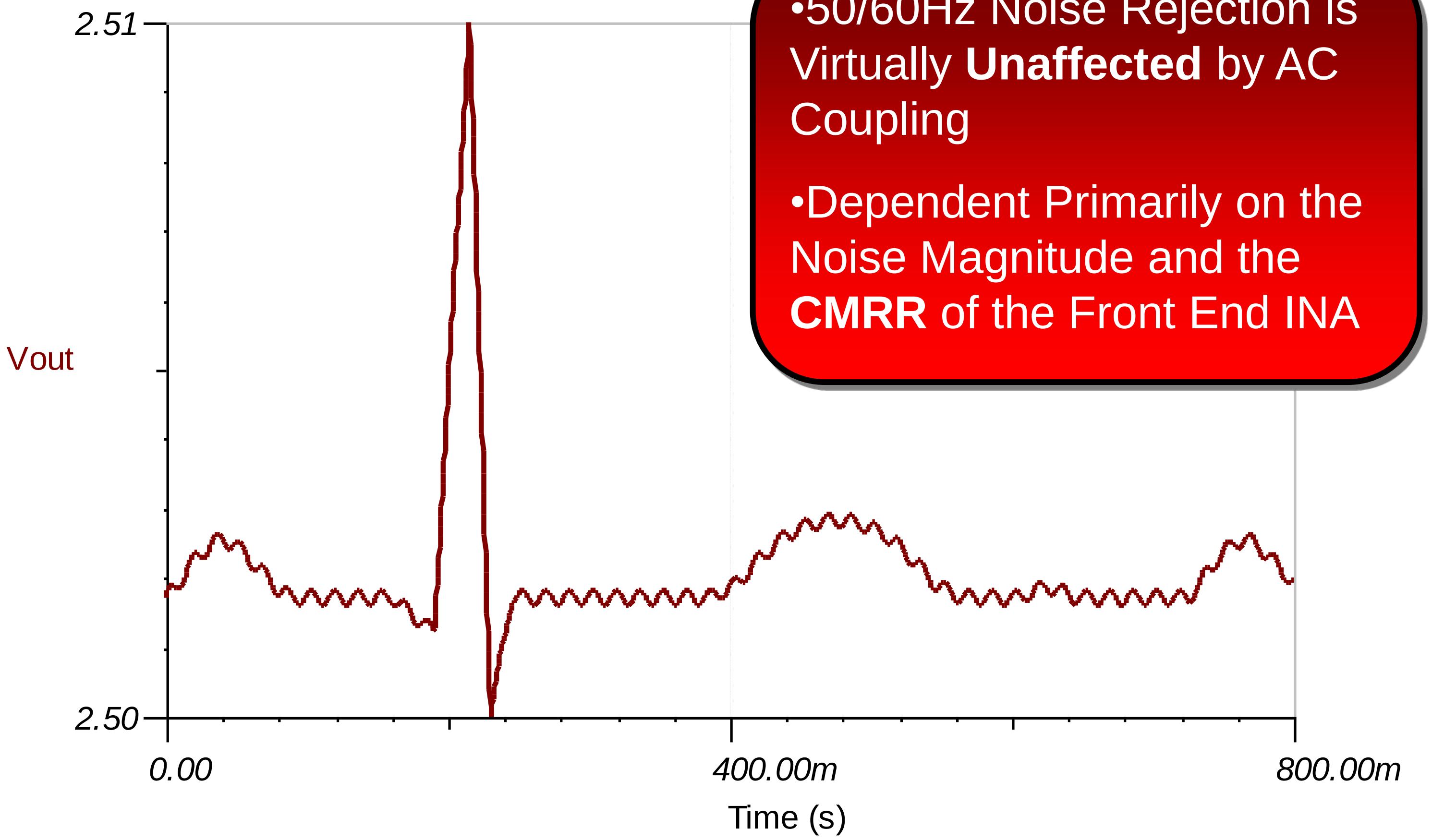
Plot of ECG Response to 5Hz CM Input Signal (0%-.5%) CC Mismatch



Lower Frequency Signals **Couple** Directly Into the ECG Signal to the Output

The INA Front End

Plot of ECG Response to 50/60 Hz CM Input Signal

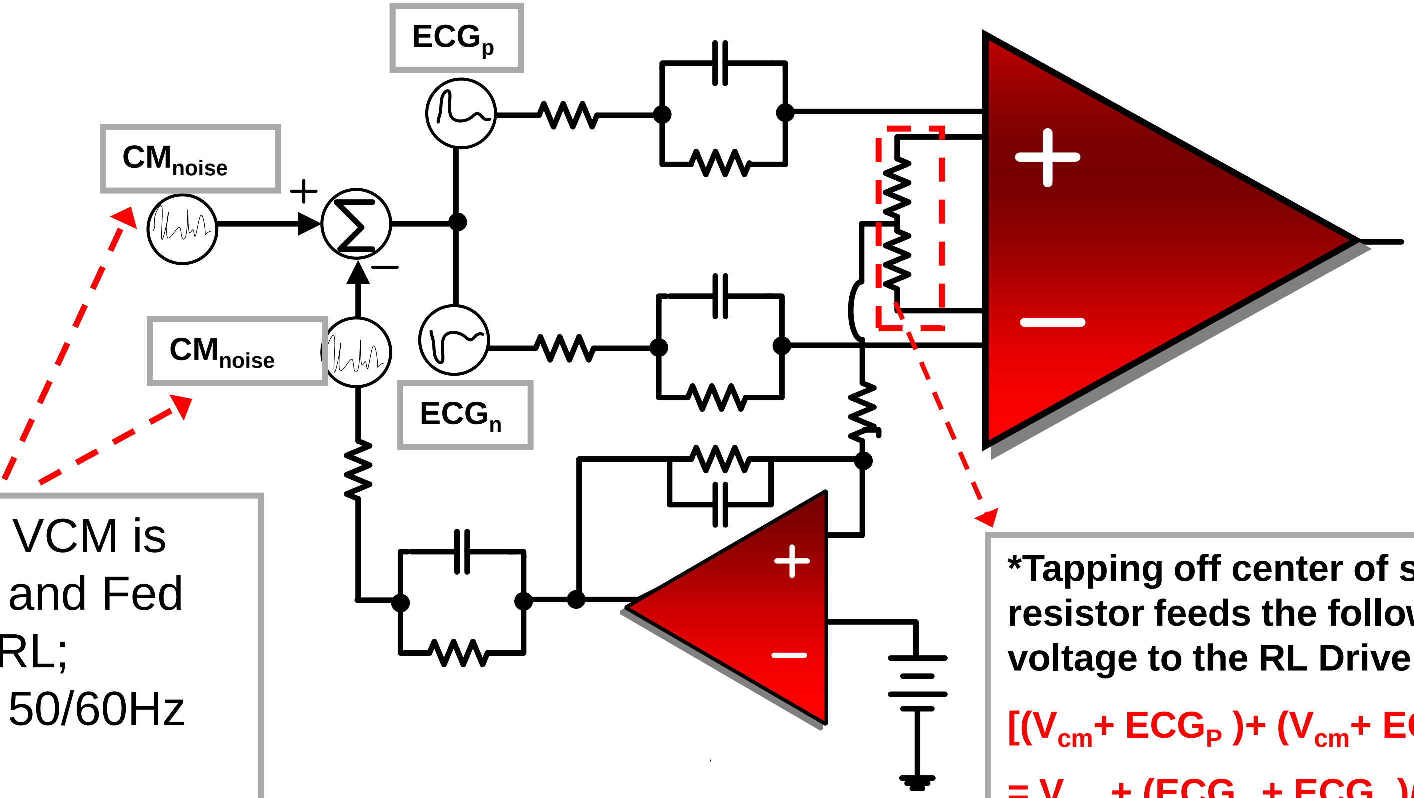


- •50/60Hz Noise Rejection is Virtually **Unaffected** by AC Coupling
- •Dependent Primarily on the Noise Magnitude and the **CMRR** of the Front End INA

The Right Leg Drive Amplifier

The RL Drive Amplifier

The RL Drive Amplifier Serves 2 Purposes: (1) Common Mode Bias (2) Noise Cancellation



Average VCM is Inverted and Fed Back to RL; Cancels 50/60Hz noise

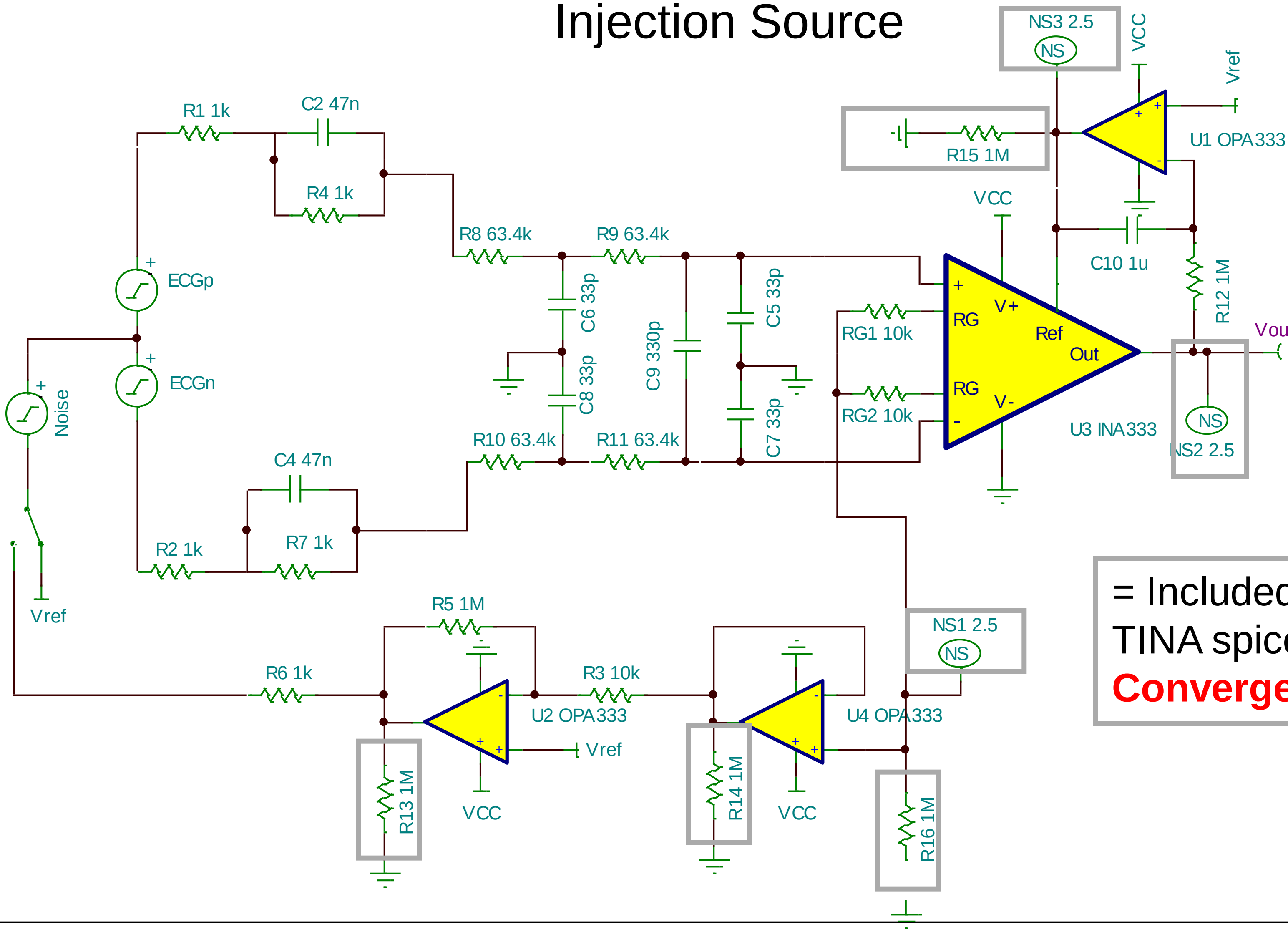
***Tapping off center of split gain resistor feeds the following voltage to the RL Drive Circuit**

$[(V_{cm}+ ECG_P)+ (V_{cm}+ ECG_n)]/ 2$

$= V_{cm} + (ECG_p + ECG_n)/2$

The RL Drive Amplifier

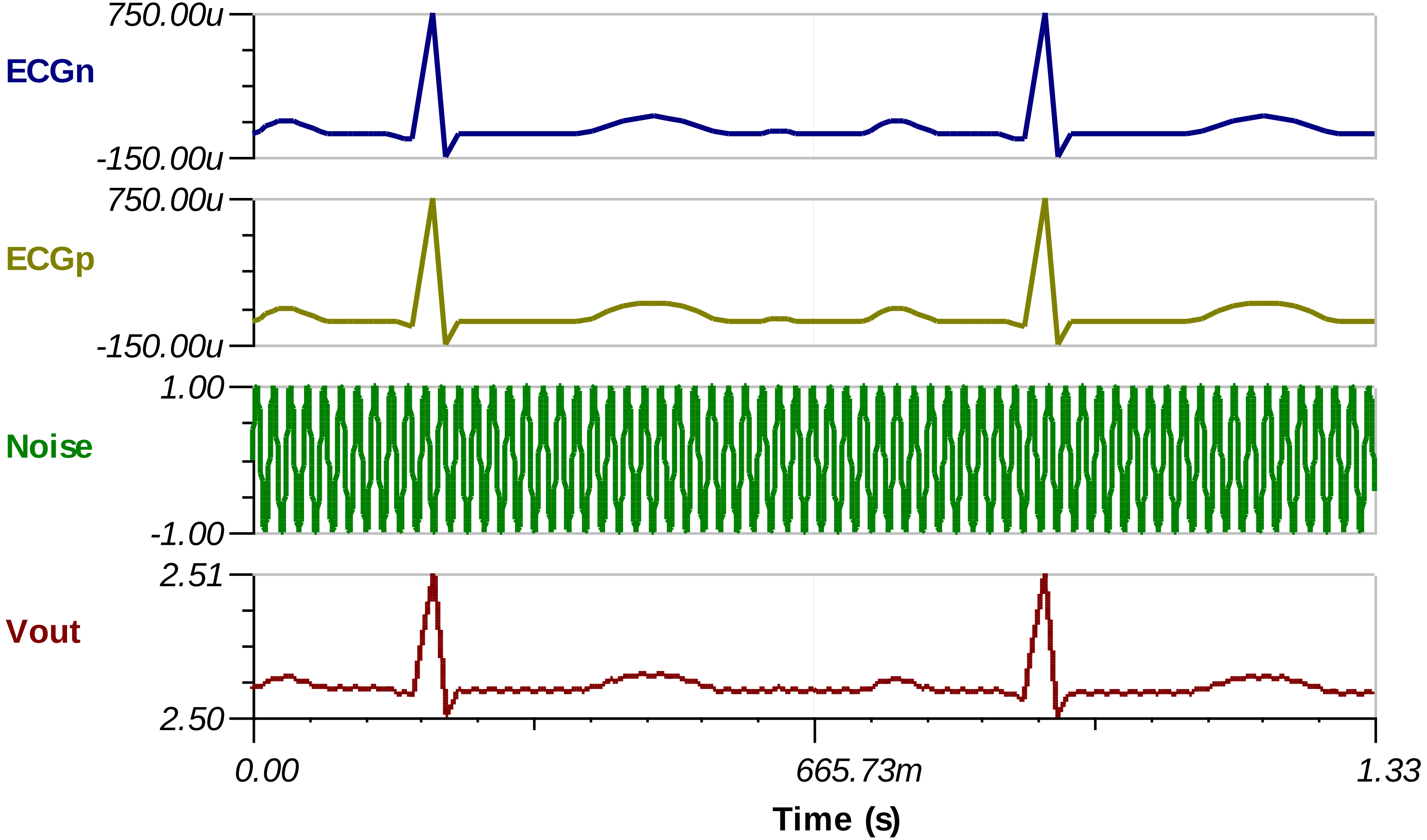
Simulation Circuit for Response to 50/60 Hz CM Noise Injection Source



= Included for TINA spice **Convergence**

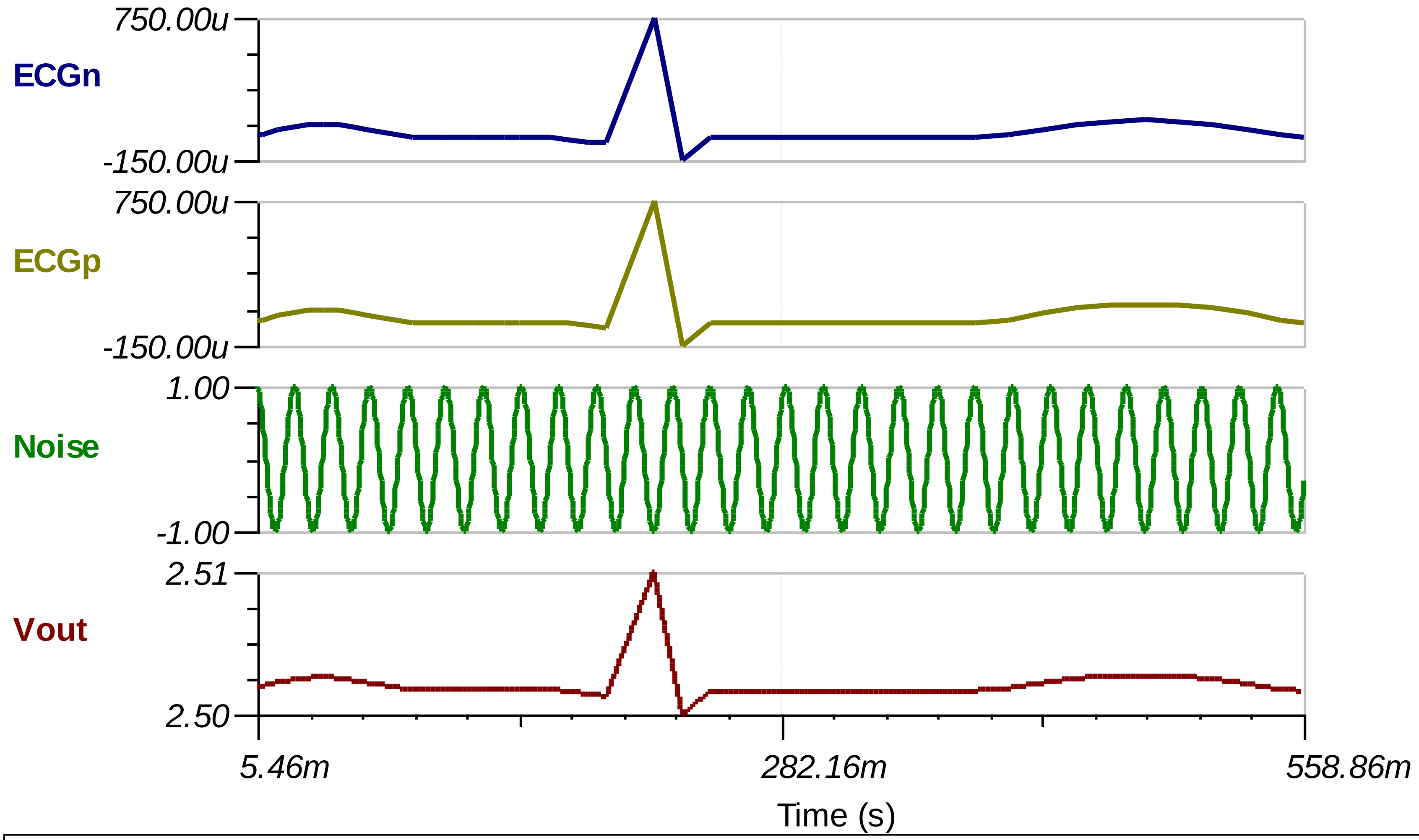
The RL Drive Amplifier TINA

Simulation with NO RL Drive; CM Noise is Coupled to Output



The RL Drive Amplifier

TINA Simulation with RL Drive; Output Noise is Reduced

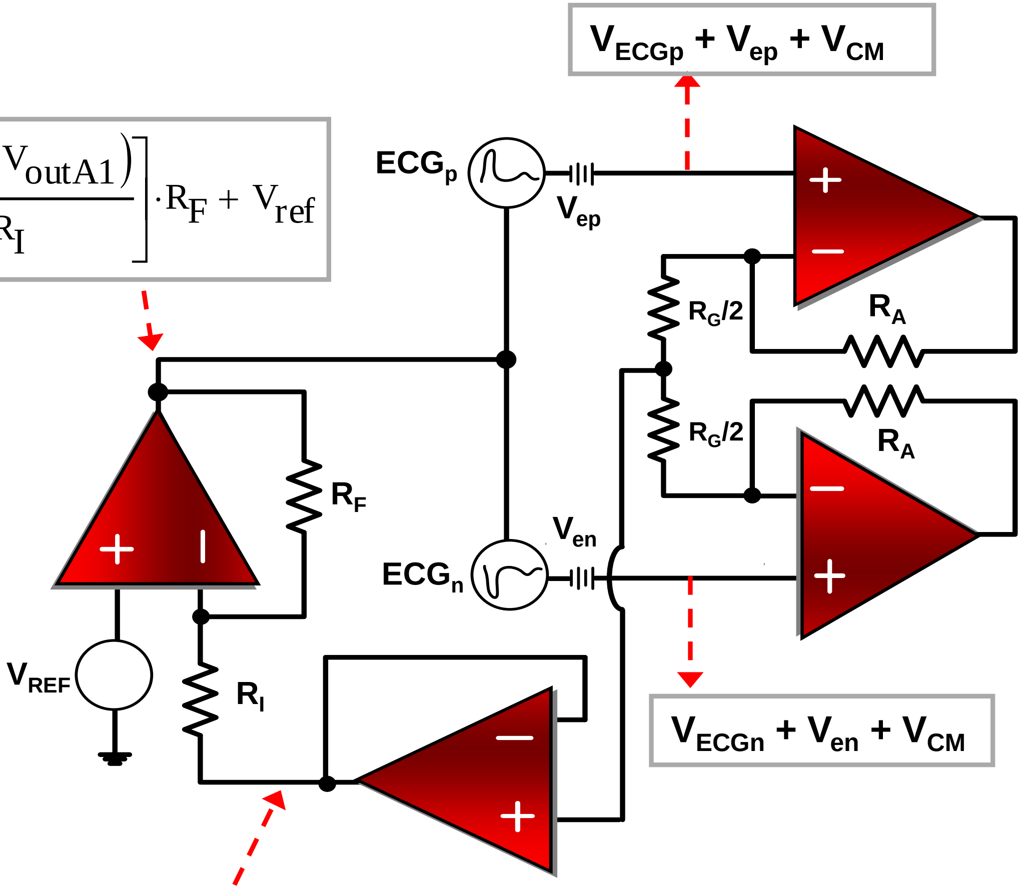


The RL Drive Amplifier

Analyzing the RLD Amplifier Loop

$V_{ECGp} + V_{ep} + V_{CM}$

$$V_{RL} = \left[\frac{\left(V_{ref} - V_{outA1}\right)}{R_I}\right] \cdot R_F + V_{ref}$$



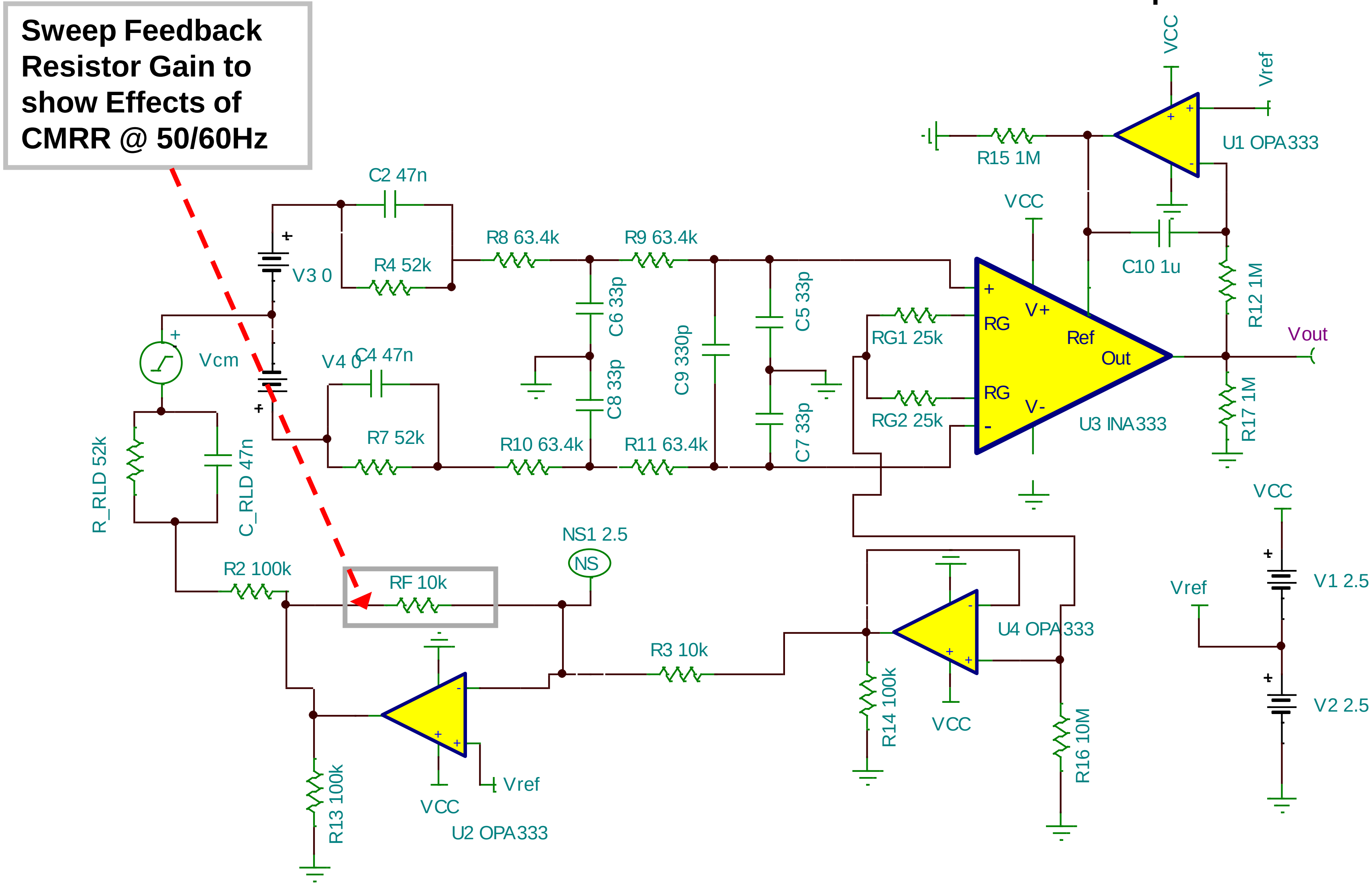
$V_{ECGn} + V_{en} + V_{CM}$

- More Gain = Better CMRR
- Loop Corrects for Electrode Offset, VOS$_{A1}$, and VOS$_{RLD}$

$$V_{outA1} = \left[\frac{\left(V_{cm} + V_{ECGp} + V_{ep}\right) - \left(V_{cm} + V_{ECGn} + V_{en}\right)}{R_G}\right] \cdot \left(\frac{R_G}{2}\right) + VOS_{A1}$$

The RL Drive Amplifier

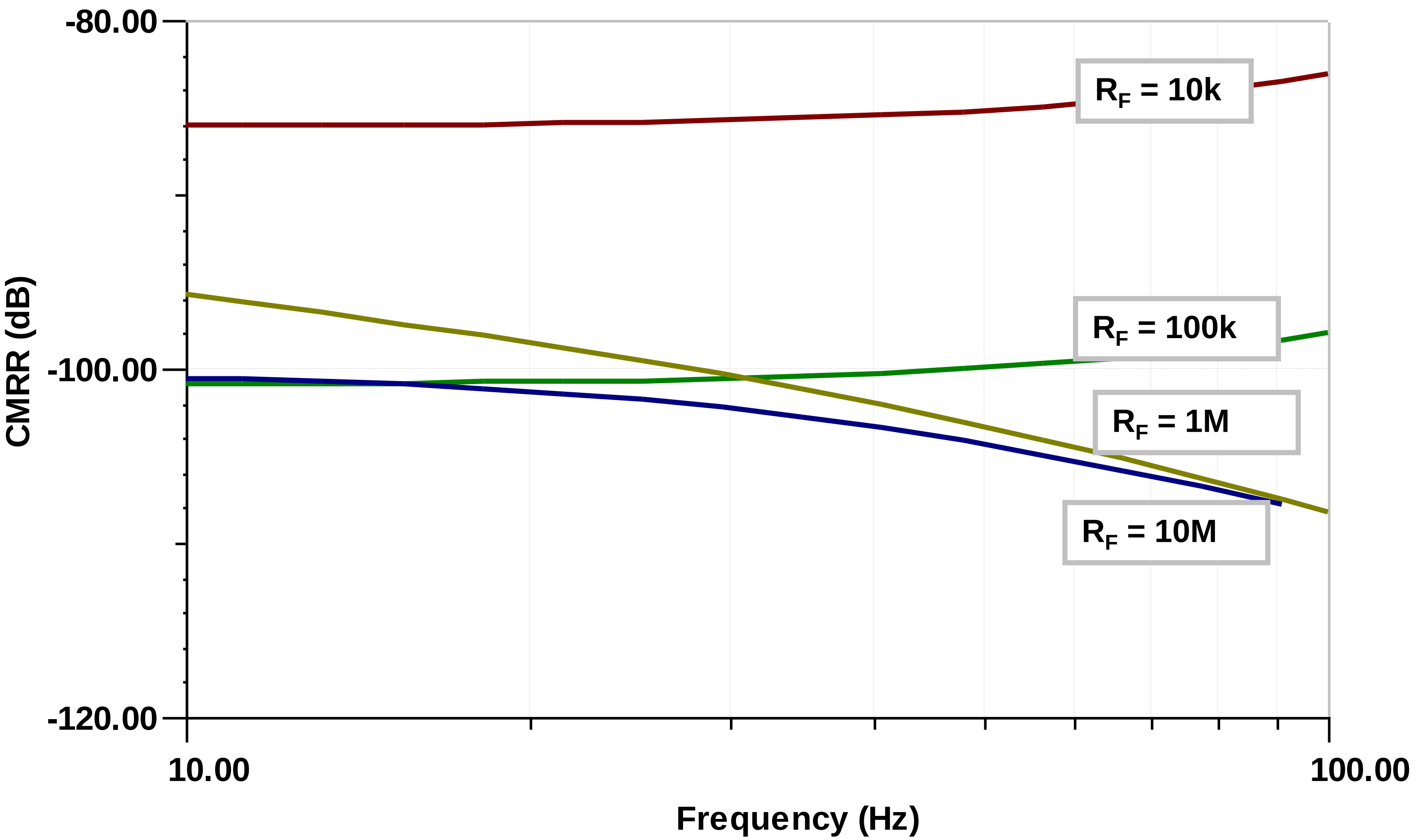
Simulation Circuit for CMRR of RLD Loop



Sweep Feedback Resistor Gain to show Effects of CMRR @ 50/60Hz

The RL Drive Amplifier

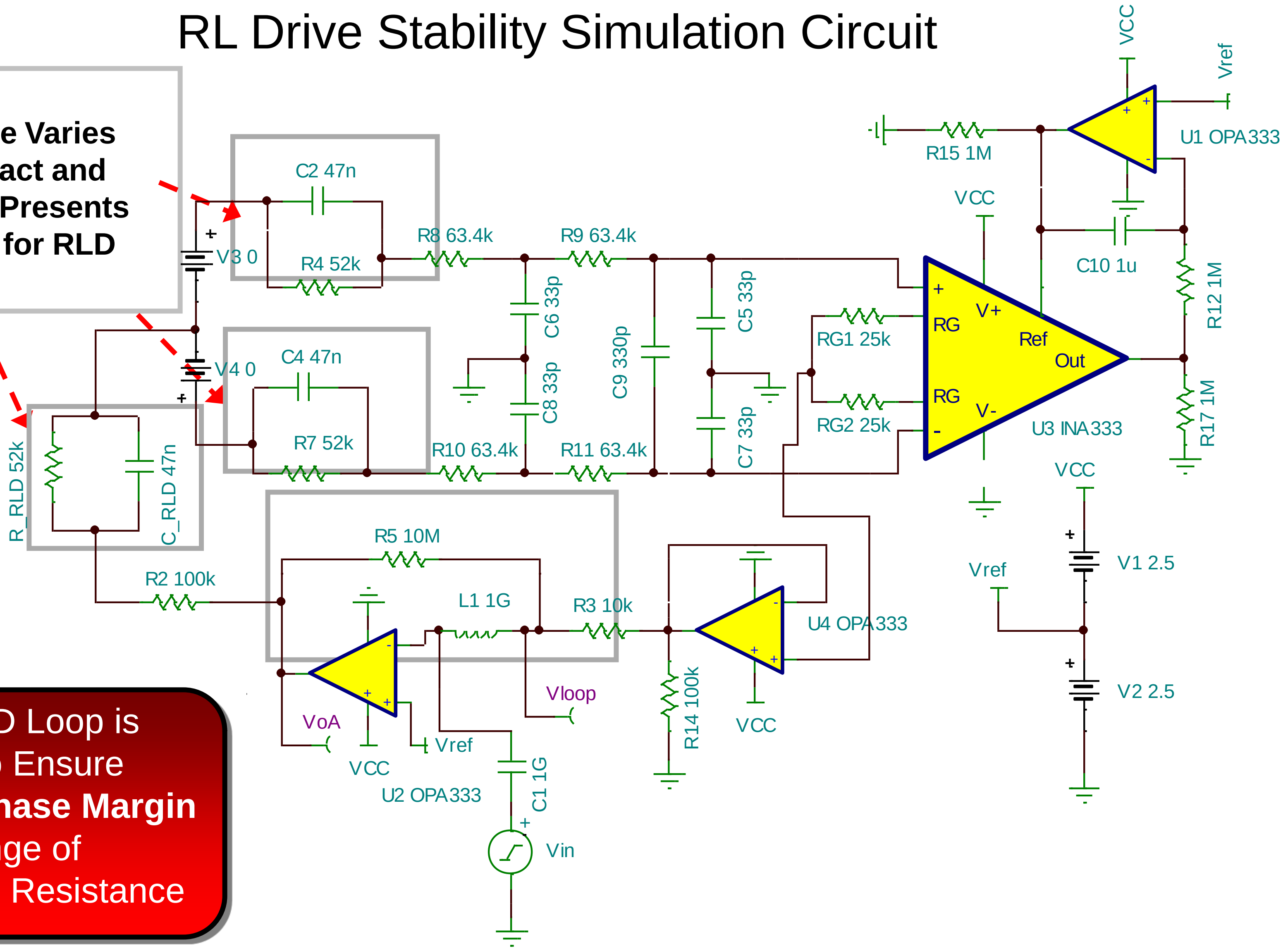
CMRR Plots vs. Gain in RLD Loop



The RL Drive Amplifier

RL Drive Stability Simulation Circuit

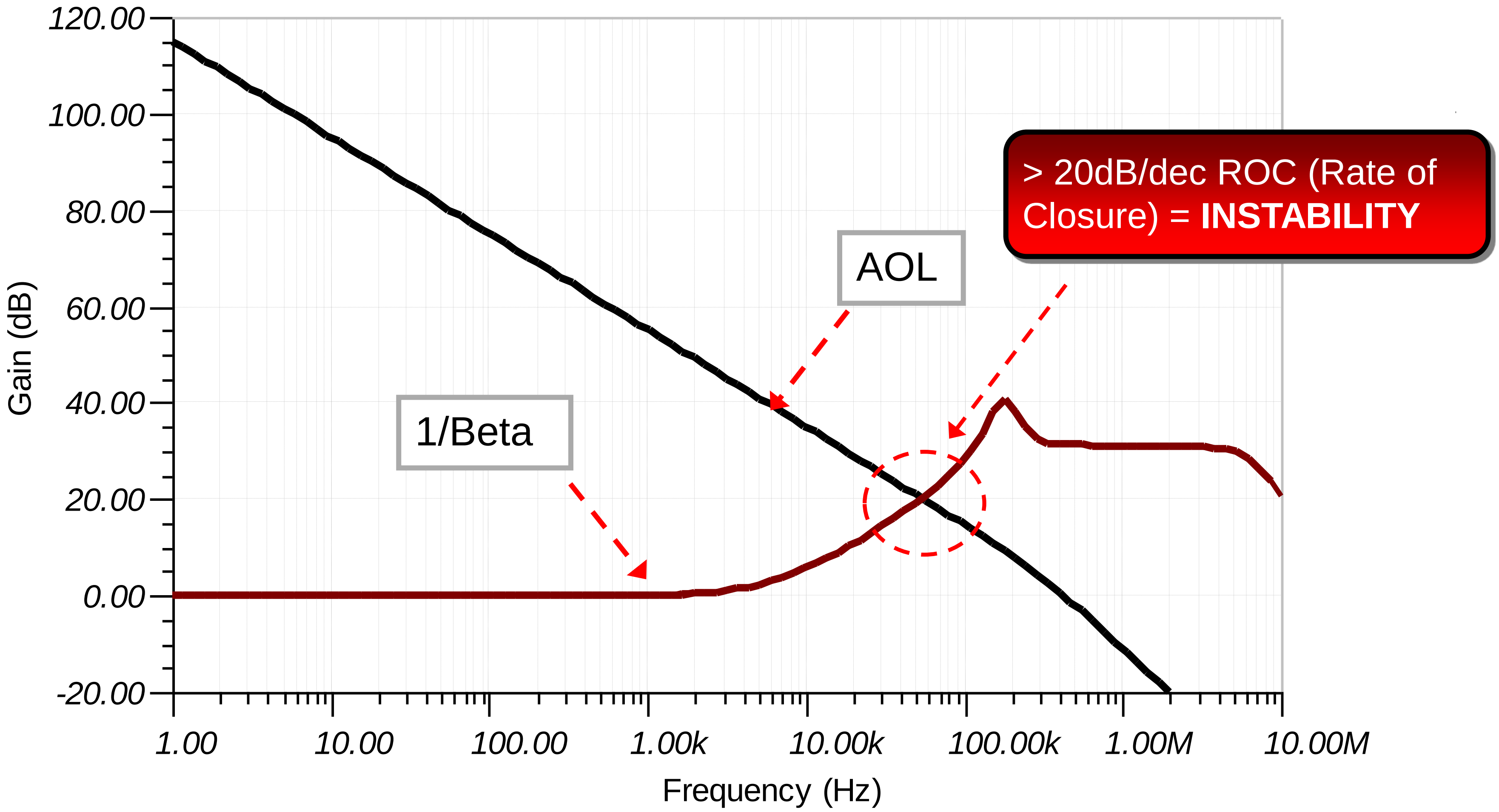
Electrode Resistance Varies With Contact and Moisture, Presents Problems for RLD Stability



Local RLD Loop is Broken to Ensure Proper **Phase Margin** Over Range of Electrode Resistance

The RL Drive Amplifier

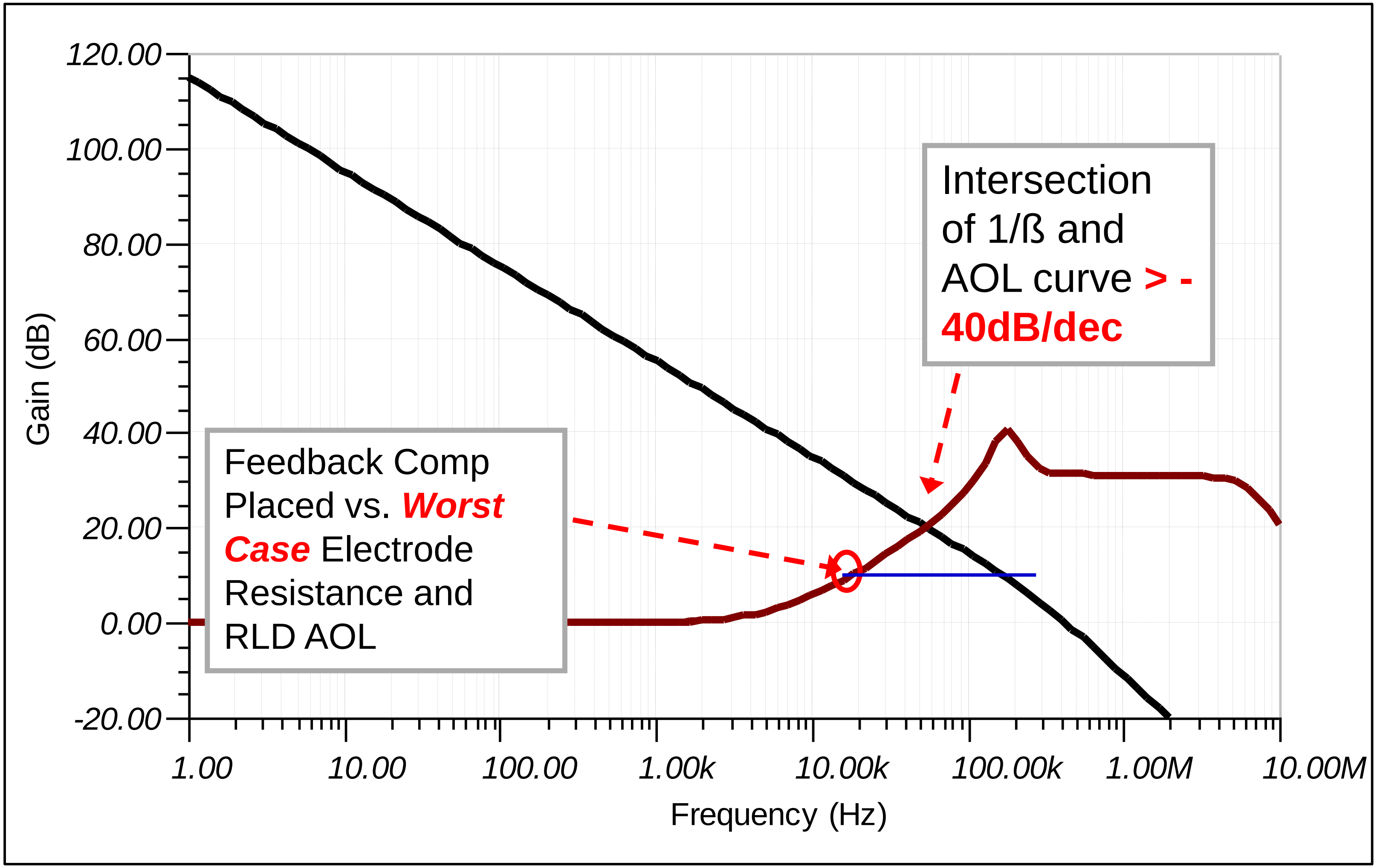
RL Drive Simulation Showing Instability in the RLD Feedback Loop



> 20dB/dec ROC (Rate of Closure) = **INSTABILITY**

The RL Drive Amplifier

Using RLD Simulation to Compensate for 1/Beta Variation With Electrode Resistance

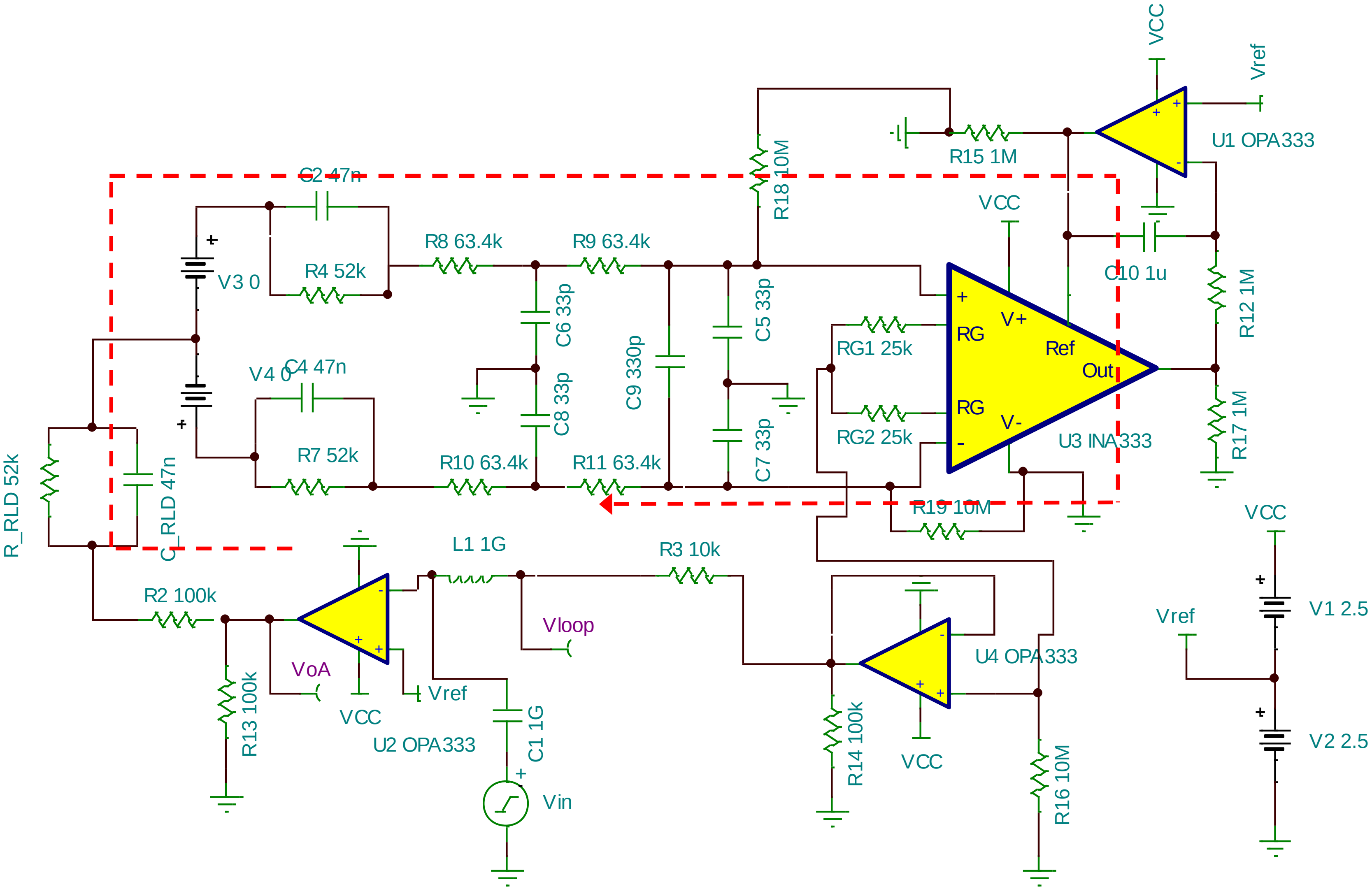


Intersection of 1/ß and AOL curve **> -40dB/dec**

Feedback Comp Placed vs. ***Worst Case*** Electrode Resistance and RLD AOL

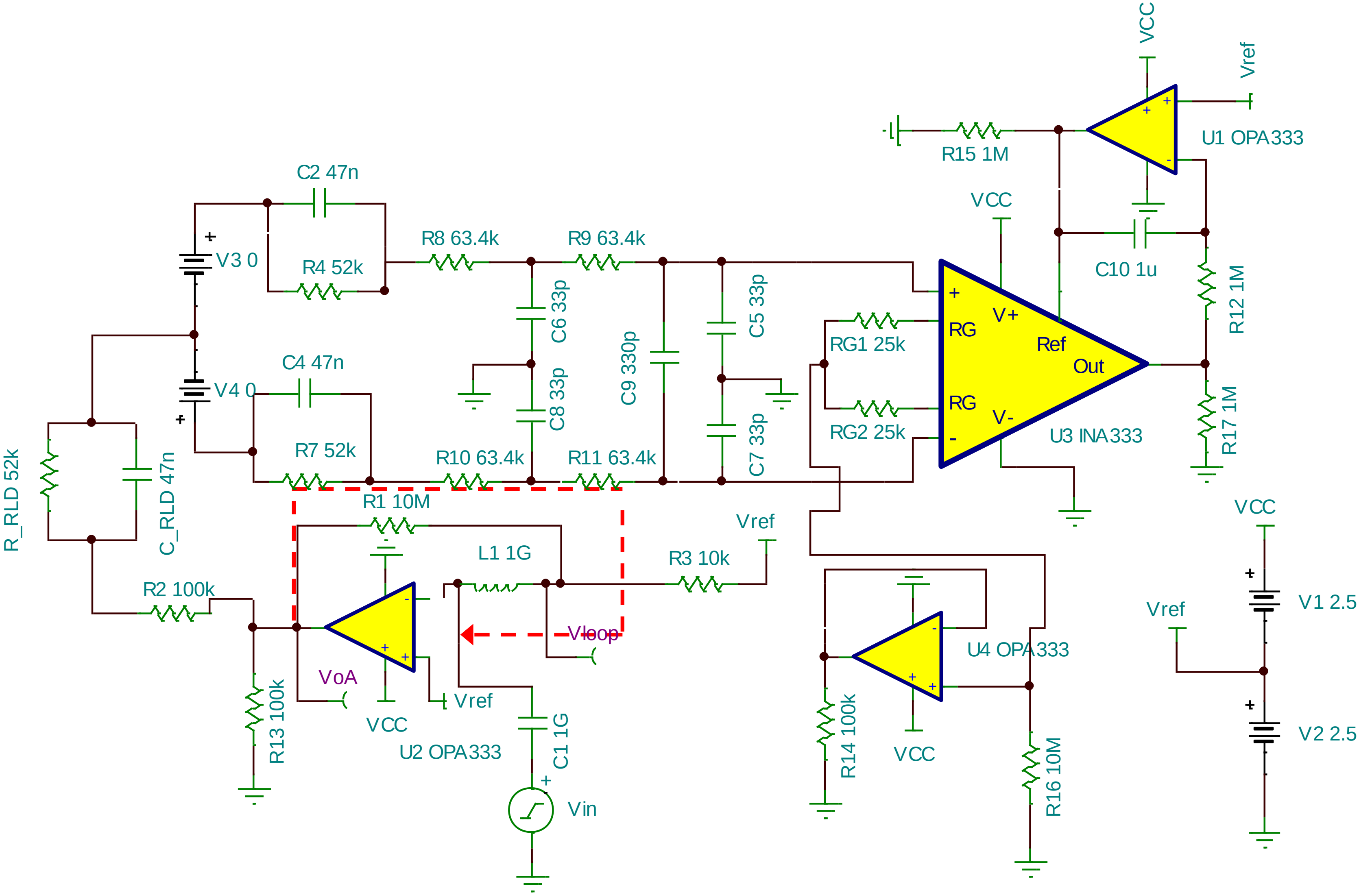
The RL Drive Amplifier

RL Drive Stability Simulation Circuit of Feedback #1



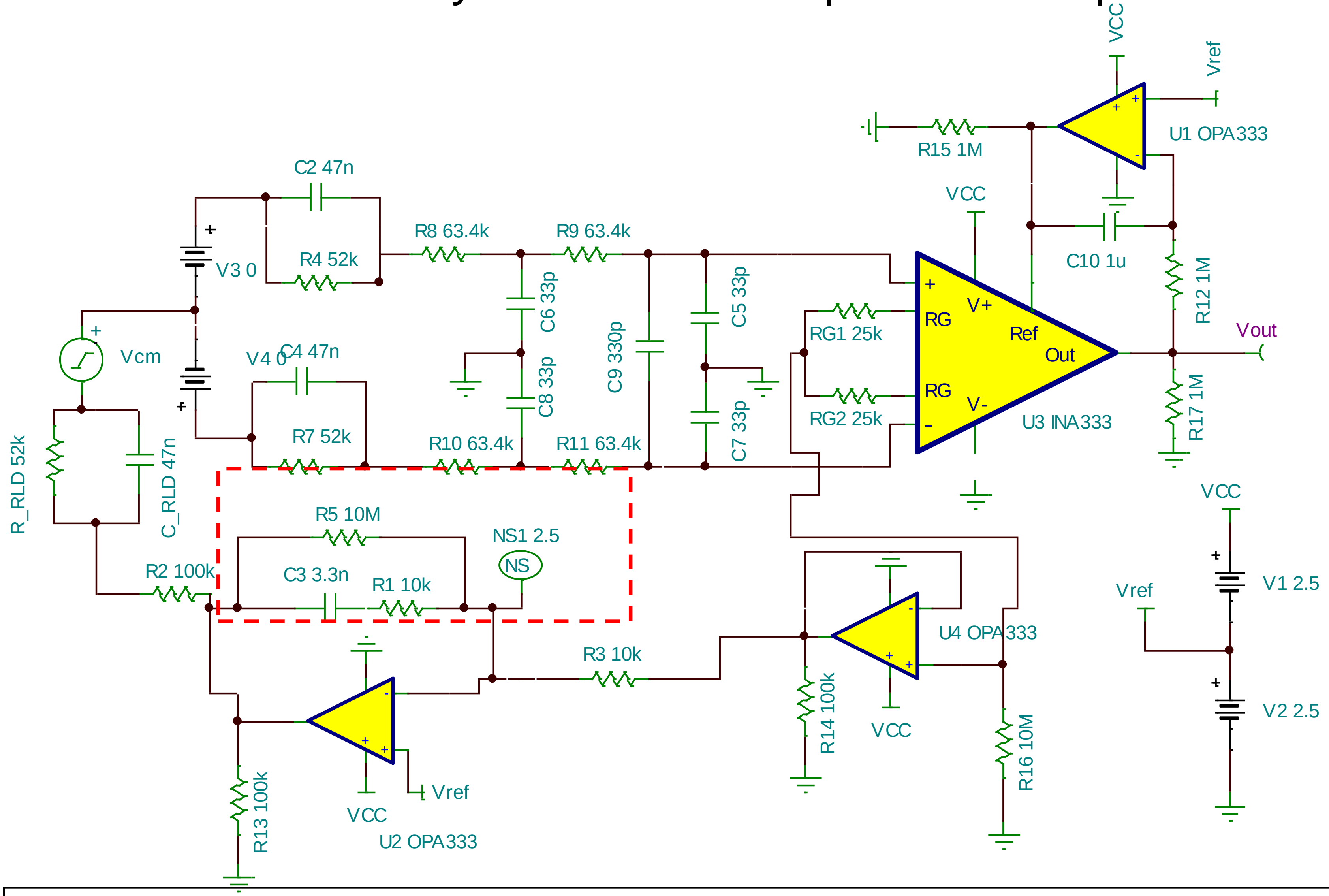
The RL Drive Amplifier

RL Drive Stability Simulation Circuit of Feedback #2



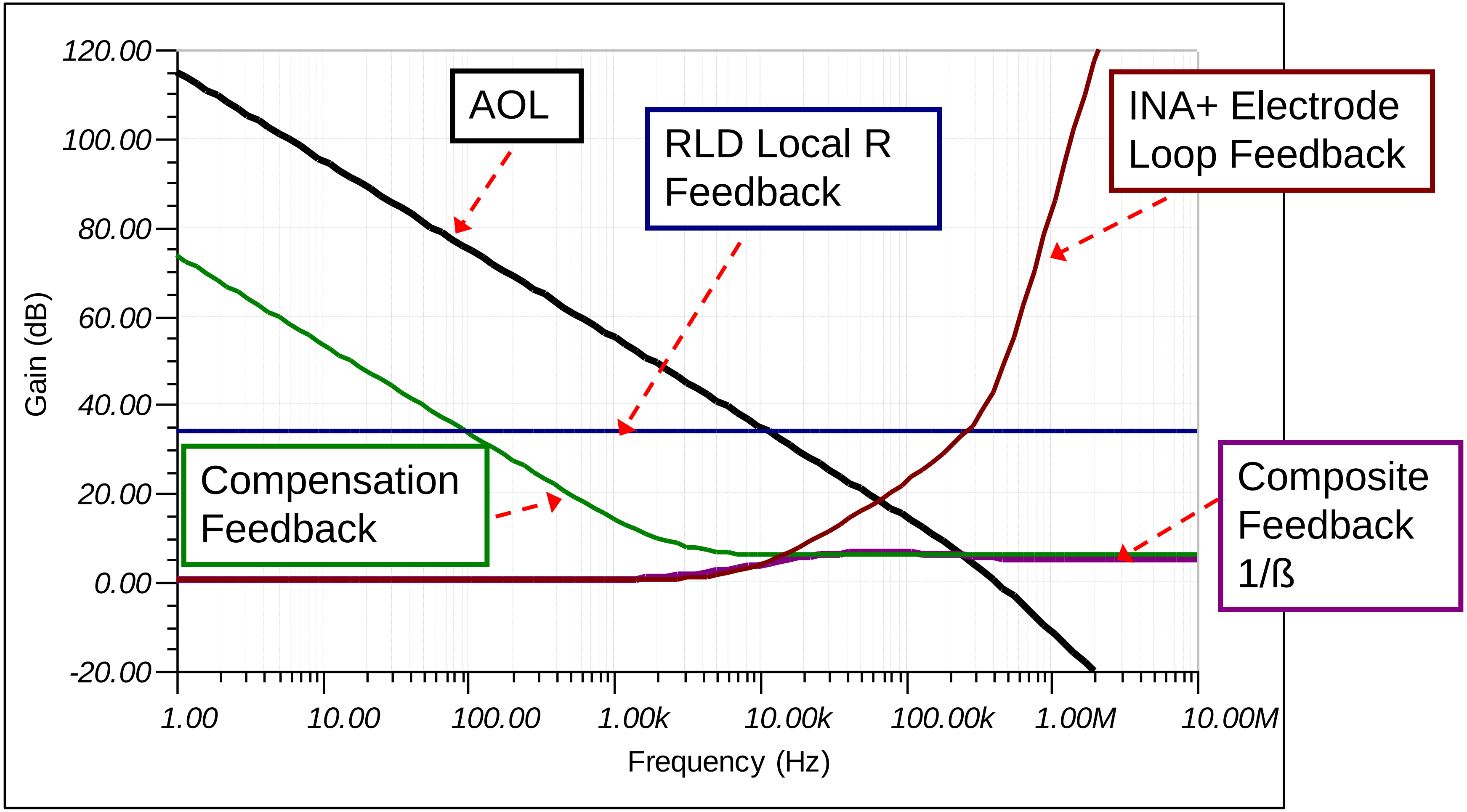
The RL Drive Amplifier

RLD Stability Circuit with Compensated Amplifier



The RL Drive Amplifier

RL Drive Stability Simulation of Separate Feedback Paths



AOL

RLD Local R Feedback

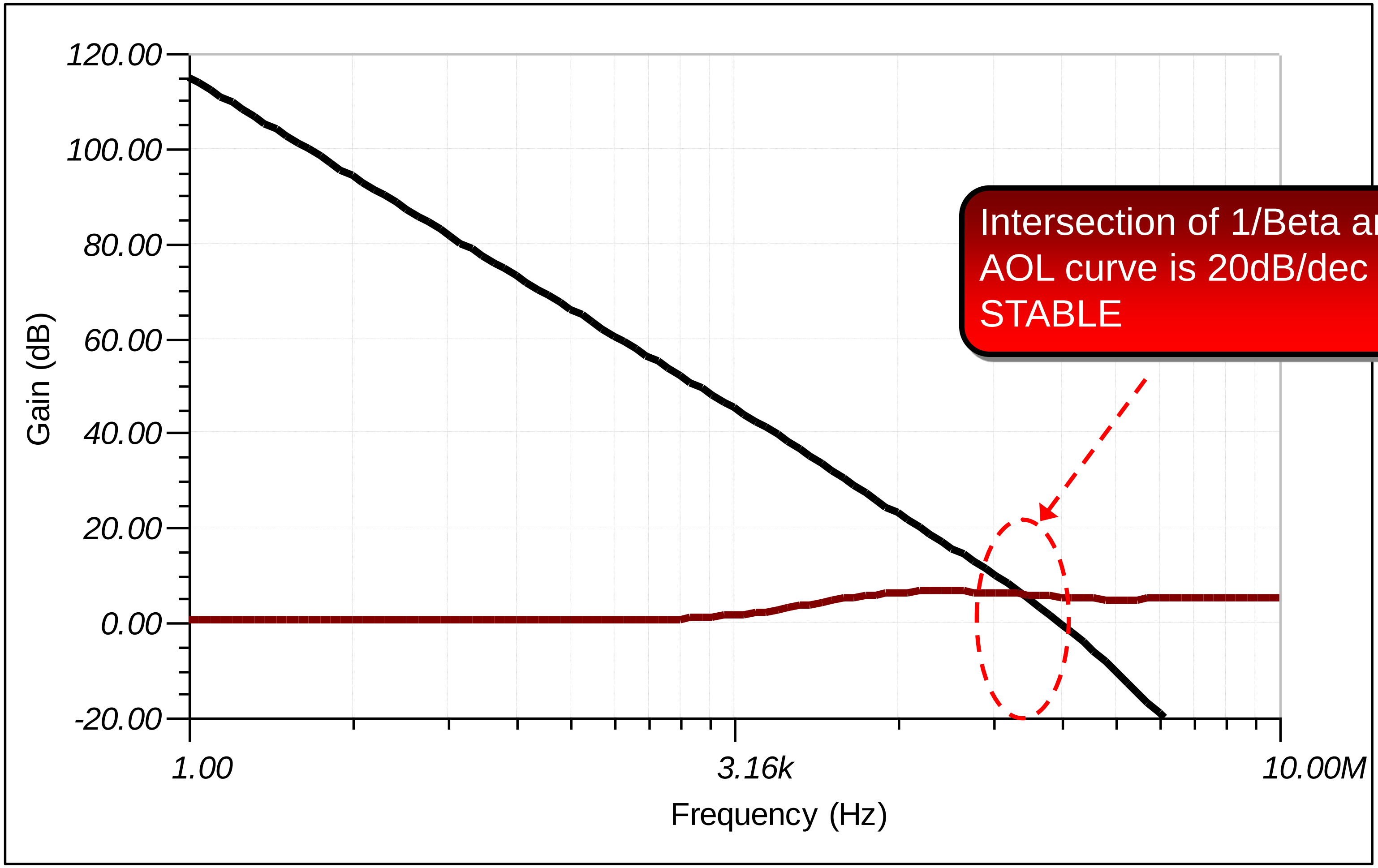
INA+ Electrode Loop Feedback

Compensation Feedback

Composite Feedback 1/ß

The RL Drive Amplifier

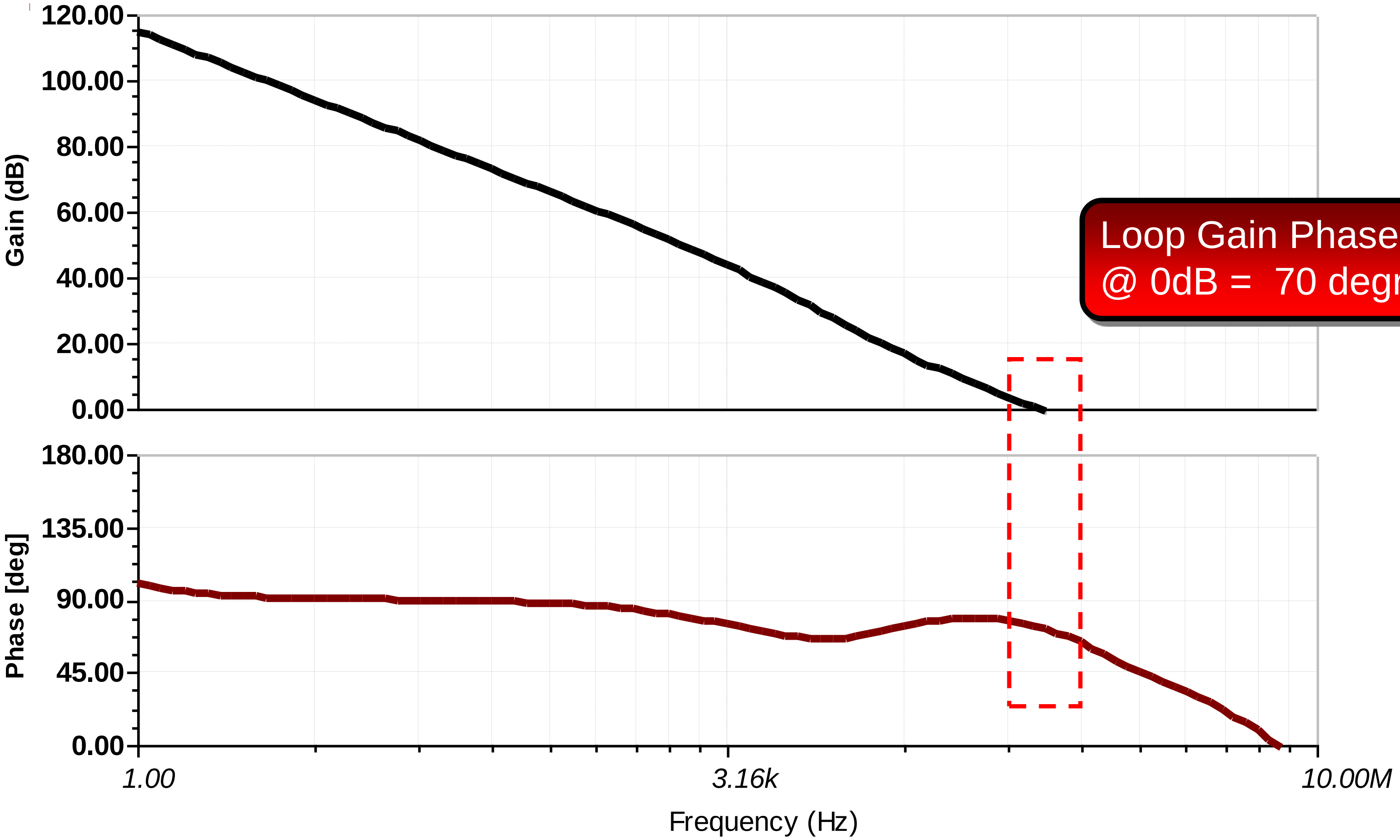
Compensated RLD Circuit Simulation of 1/Beta and AOL Intersection



Intersection of 1/Beta and AOL curve is 20dB/dec = STABLE

The RL Drive Amplifier

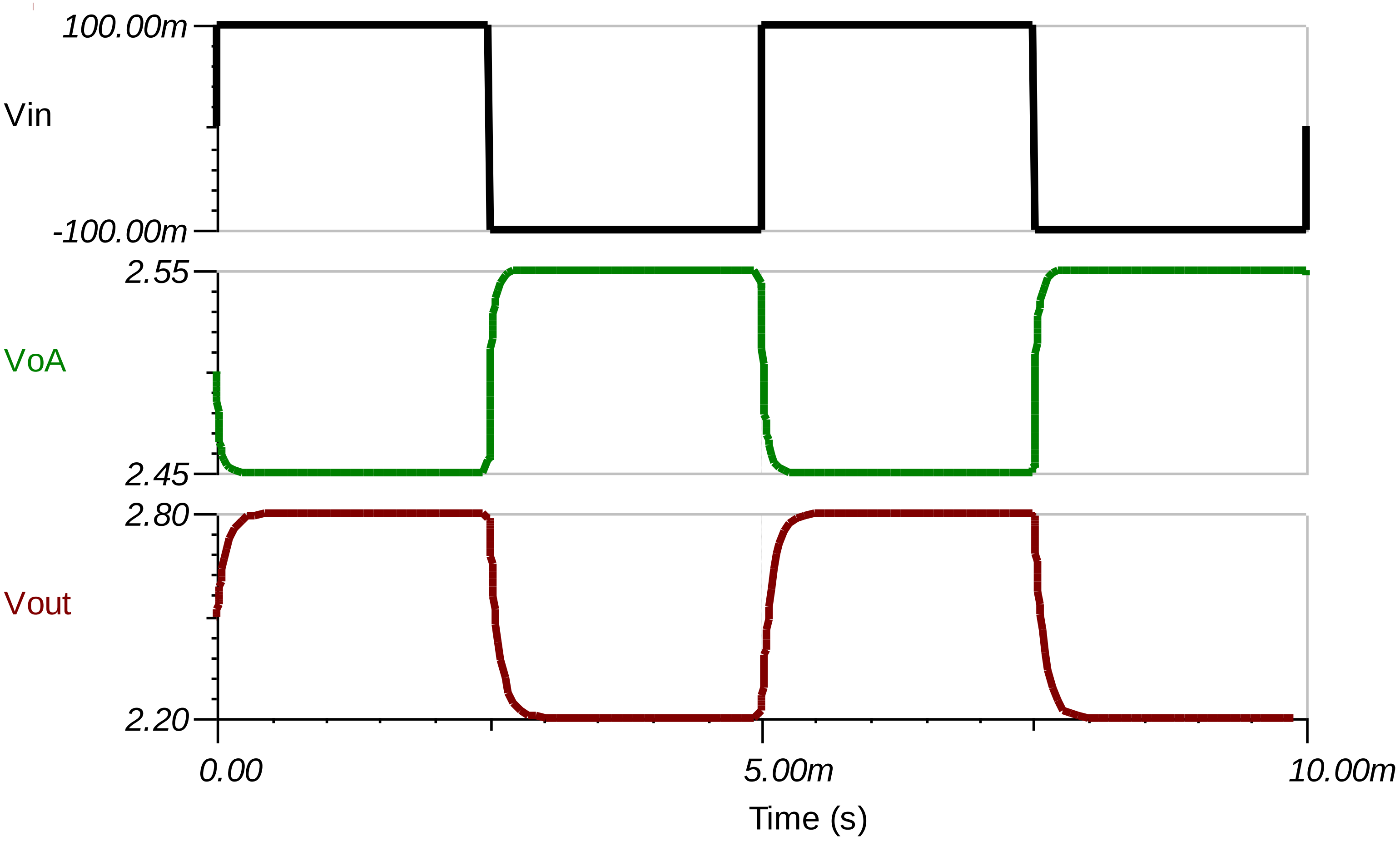
Gain and Phase Margin Plots of Compensated RLD Amplifier



Loop Gain Phase Margin @ 0dB = 70 degrees

The RL Drive Amplifier

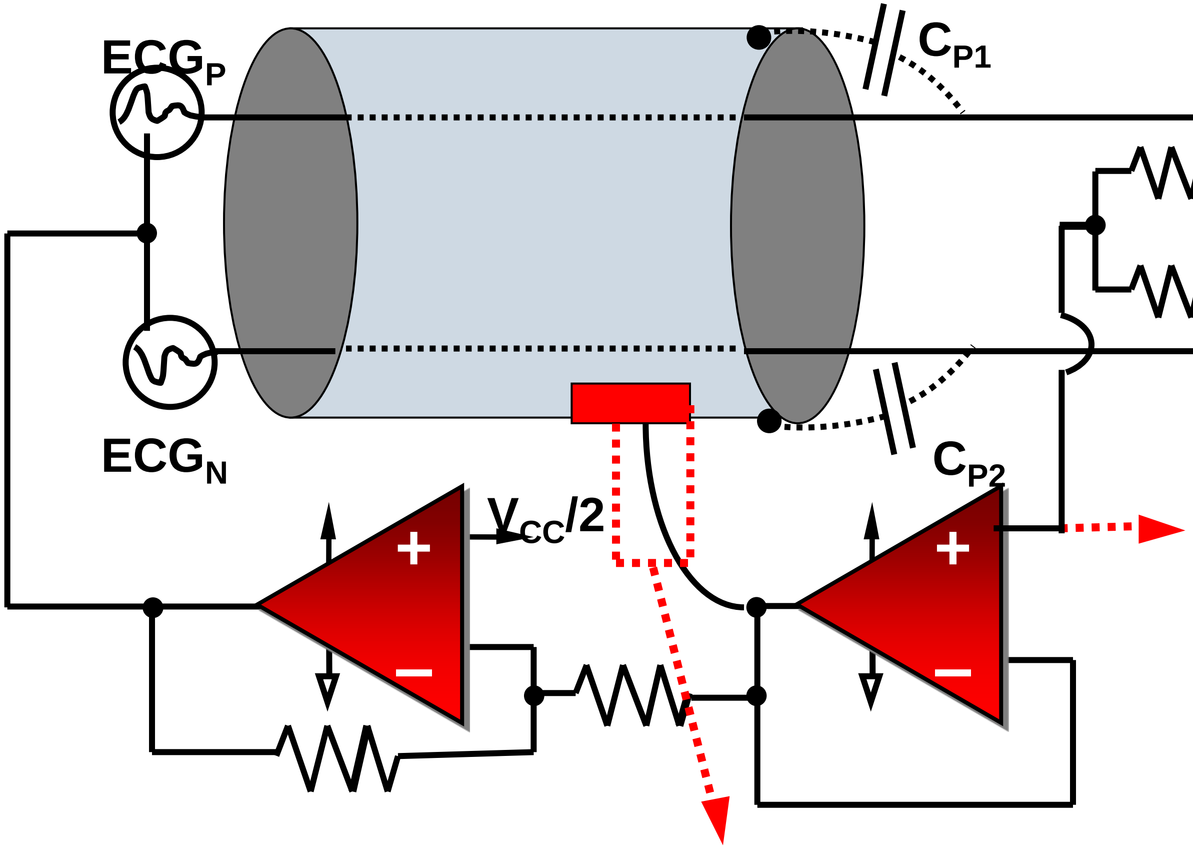
Step Response of RLD Amplifier and ECG Output



The ECG Shield Drive

The ECG Shield Drive

Shield drive eliminates leakage to ECG Inputs

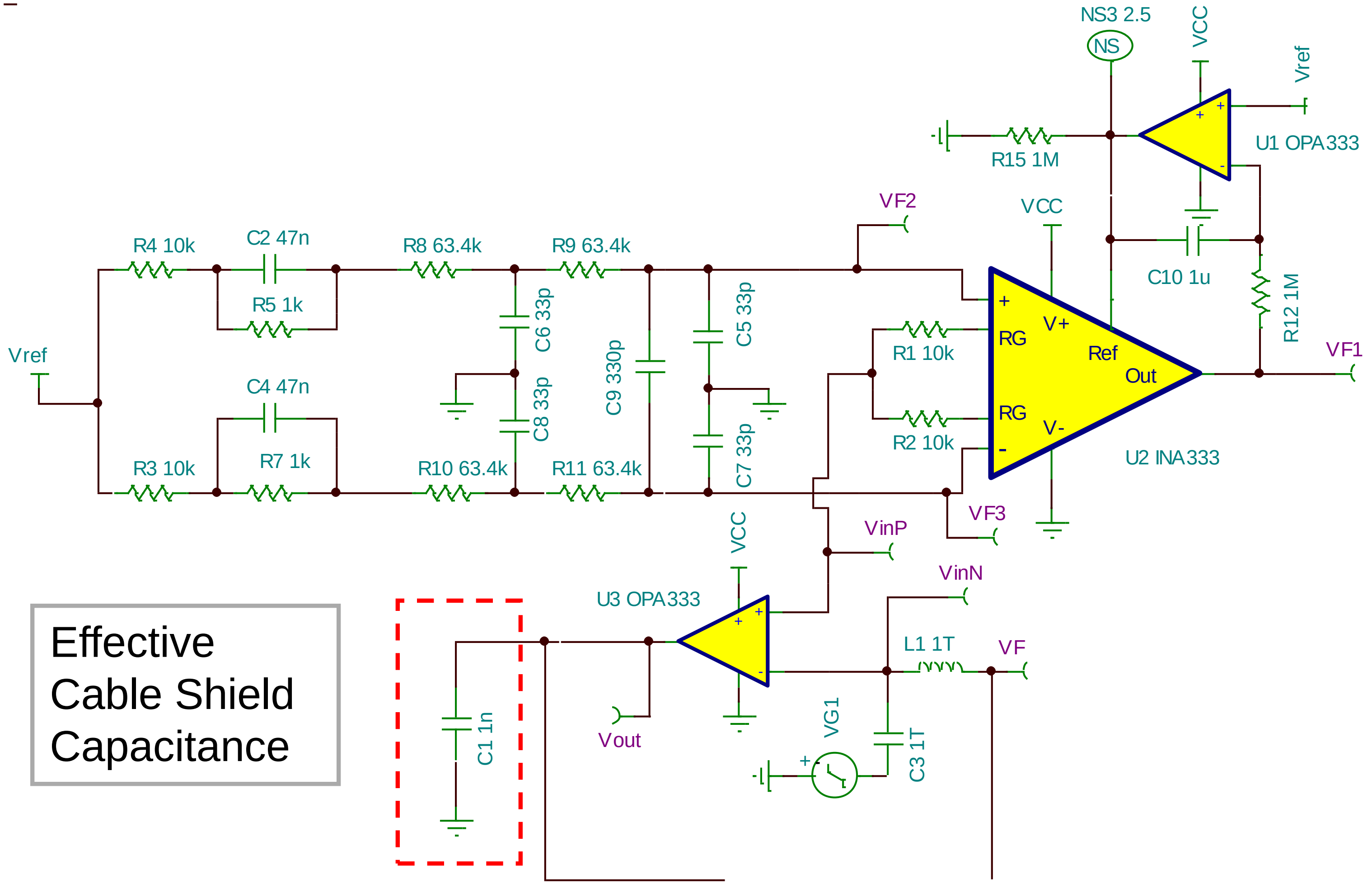


- **Shield is driven to ($V_{IN(+)}$ — $V_{IN(–)}$) /2**
- **Eliminates Leakage from C_{P1} and C_{P2}**

- **Capacitance of cable can be 500 pF to 1.5 nF**
- **Isolation resistor Necessary for improved EMI/RFI filtering**

The ECG Shield Drive

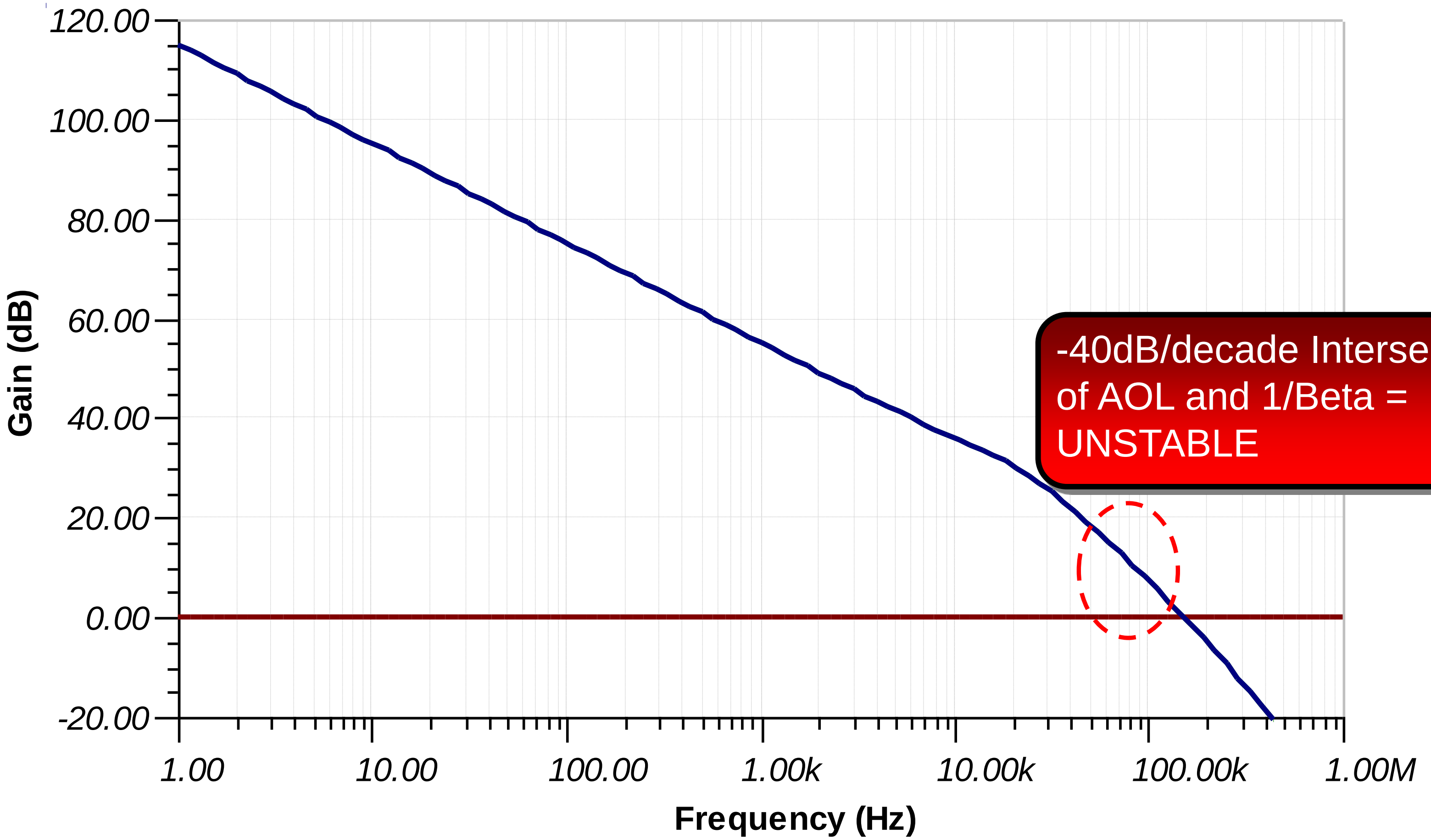
AC Stability Simulation Circuit for OPA333 as Shield Driver



Effective Cable Shield Capacitance

The ECG Shield Drive

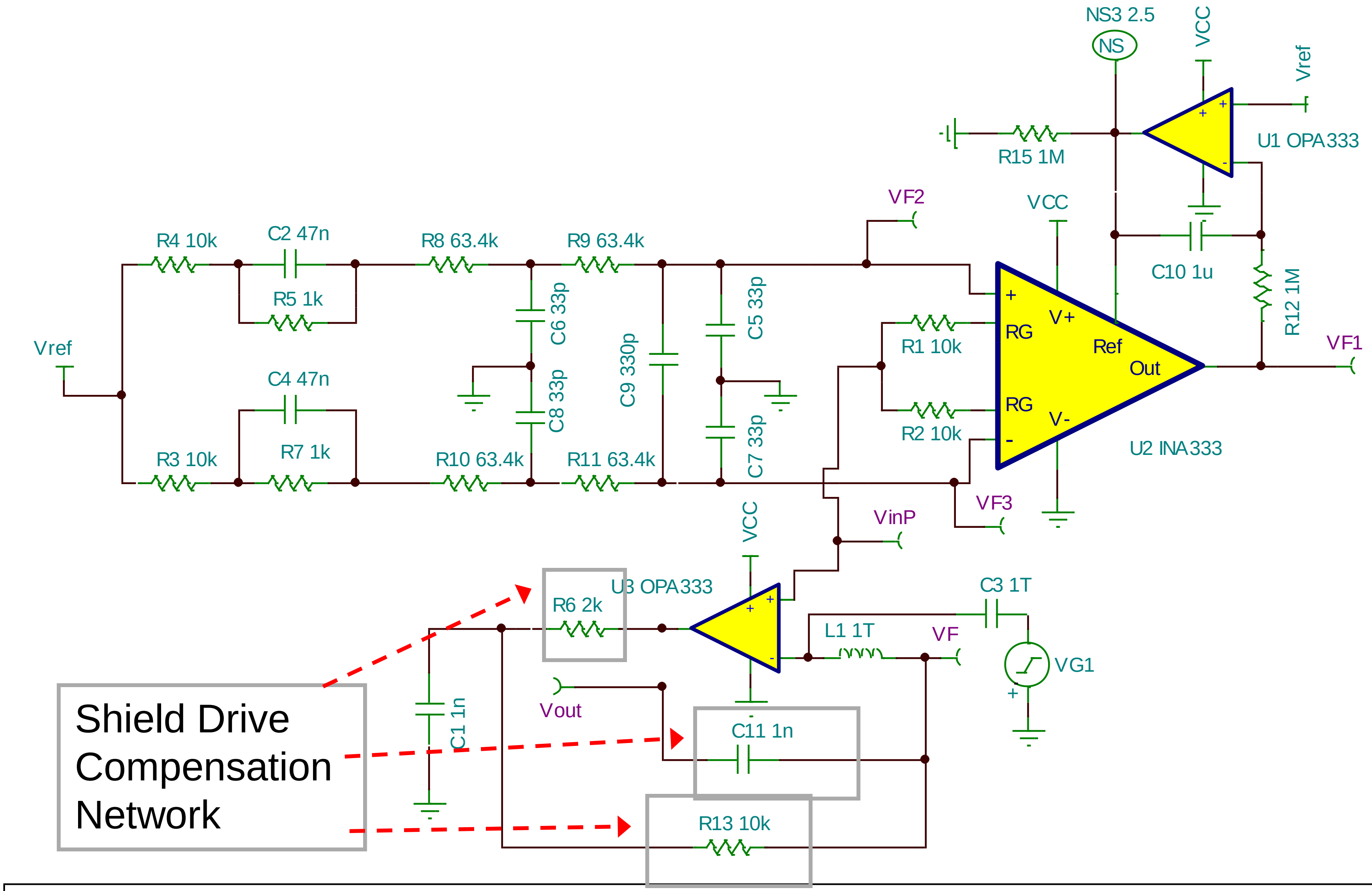
AOL + 1/Beta Response of OPA333 Shield Drive and 1nF Cable Capacitance



-40dB/decade Intersection of AOL and 1/Beta = UNSTABLE

The ECG Shield Drive

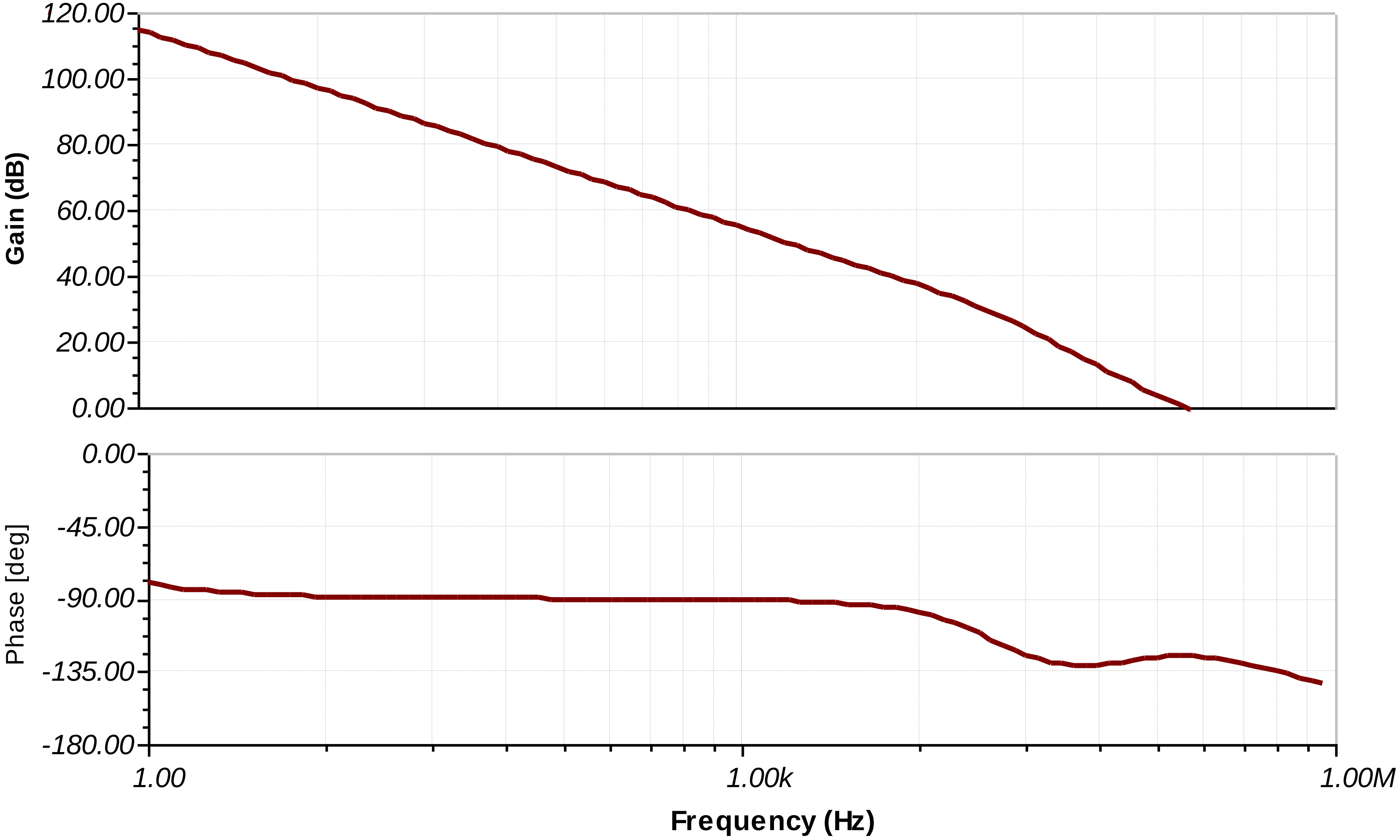
TINA Simulation Circuit for Stabilized OPA333 Shield Driver



Shield Drive
Compensation
Network

The ECG Shield Drive

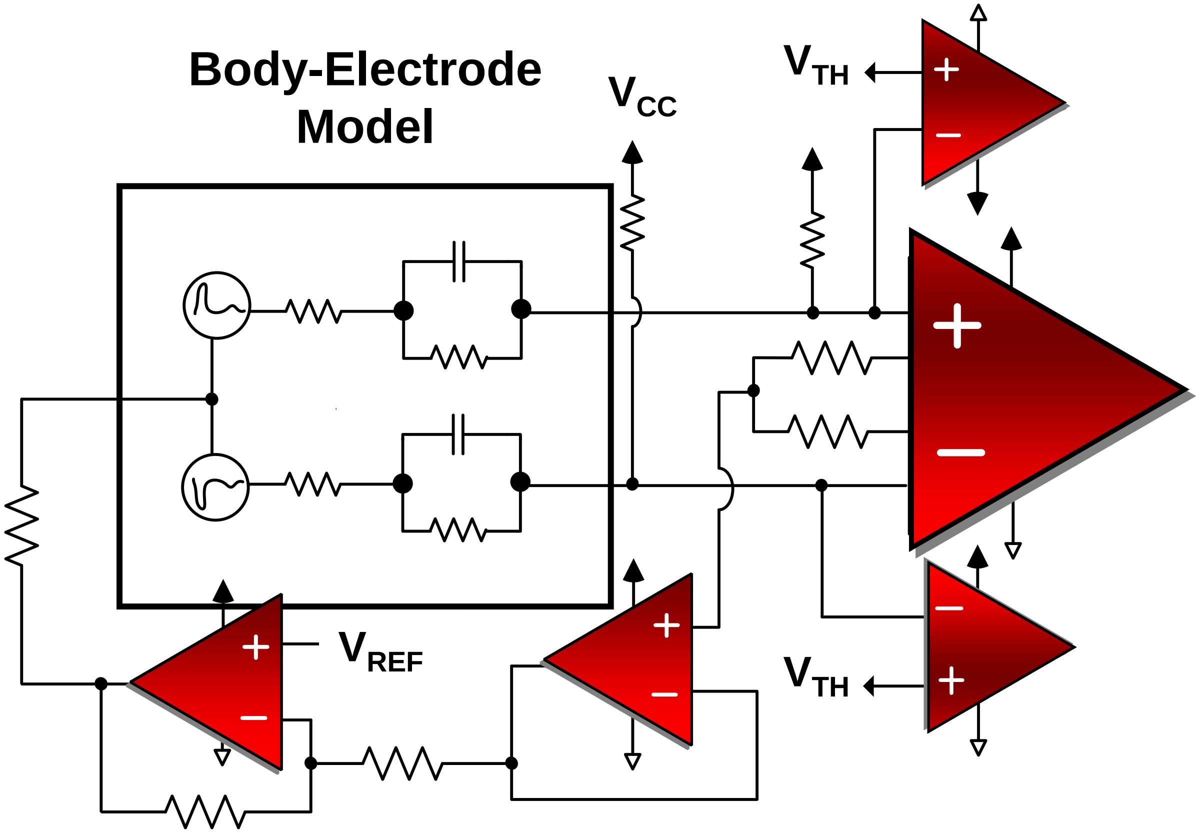
TINA Simulation Shows > 45 Degrees Phase Margin for OPA333 Shield Driver



Lead Off Detection

Lead Off Detection

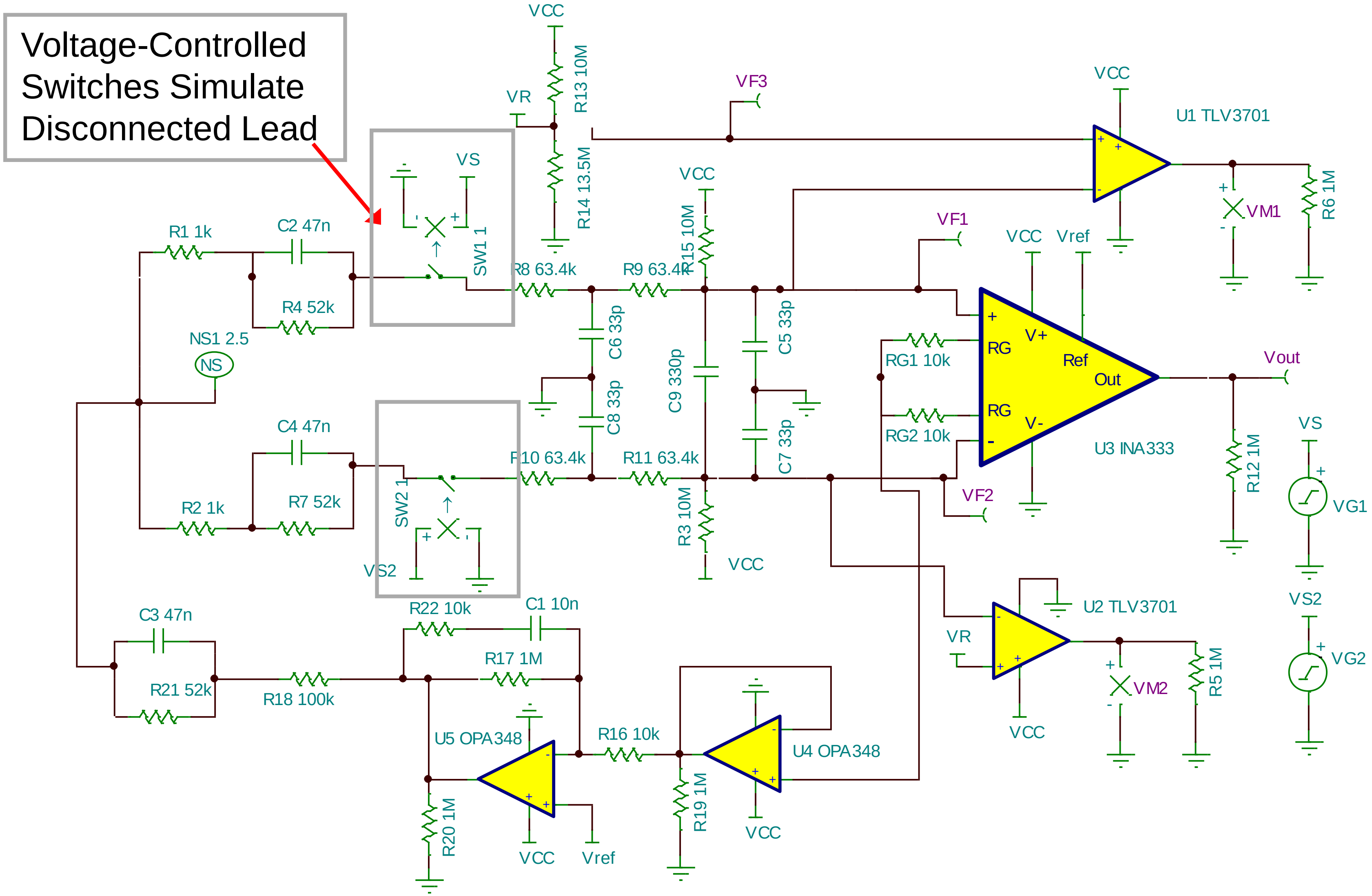
Lead Off Differentiates a Bad Lead from an Arrythmia



- Pull up Resistors Force +IN to Comparator High When Lead is Removed
- Comparator Voltage triggers ALERT
- Lead Off Indicative of “Weak Lead”

Lead Off Detection

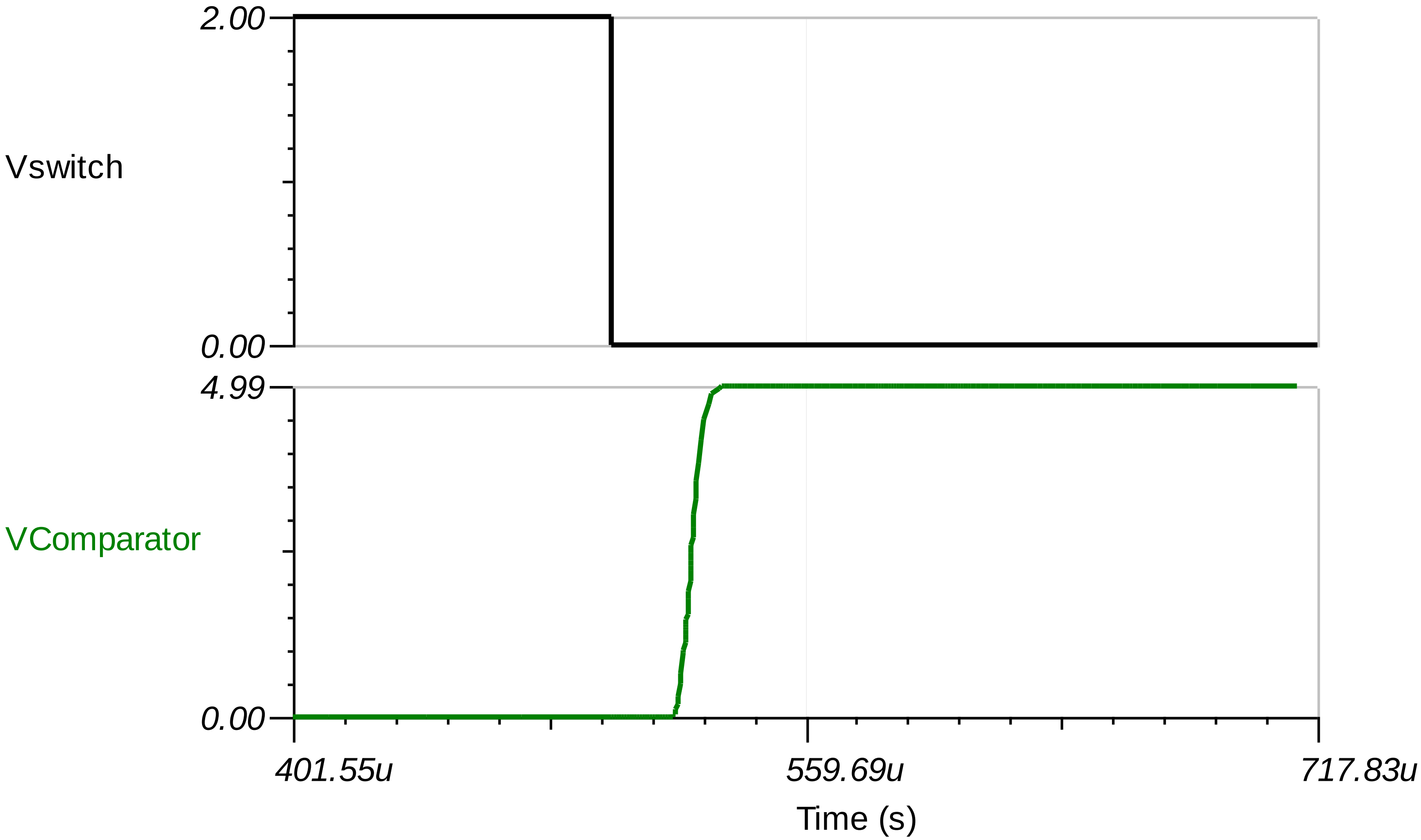
TINA Simulation Circuit for Lead Off Detect



Voltage-Controlled Switches Simulate Disconnected Lead

Lead Off Detection

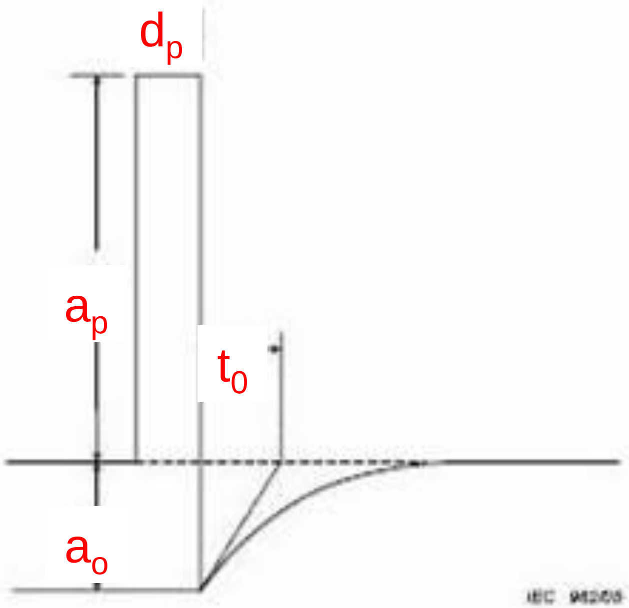
TINA Simulation Results for Lead Off Detect



Pace Detection

Pace Detect

Pace Maker Pulse Specifications



a_p = Amplitude (2-700mV)

a_o = Overshoot

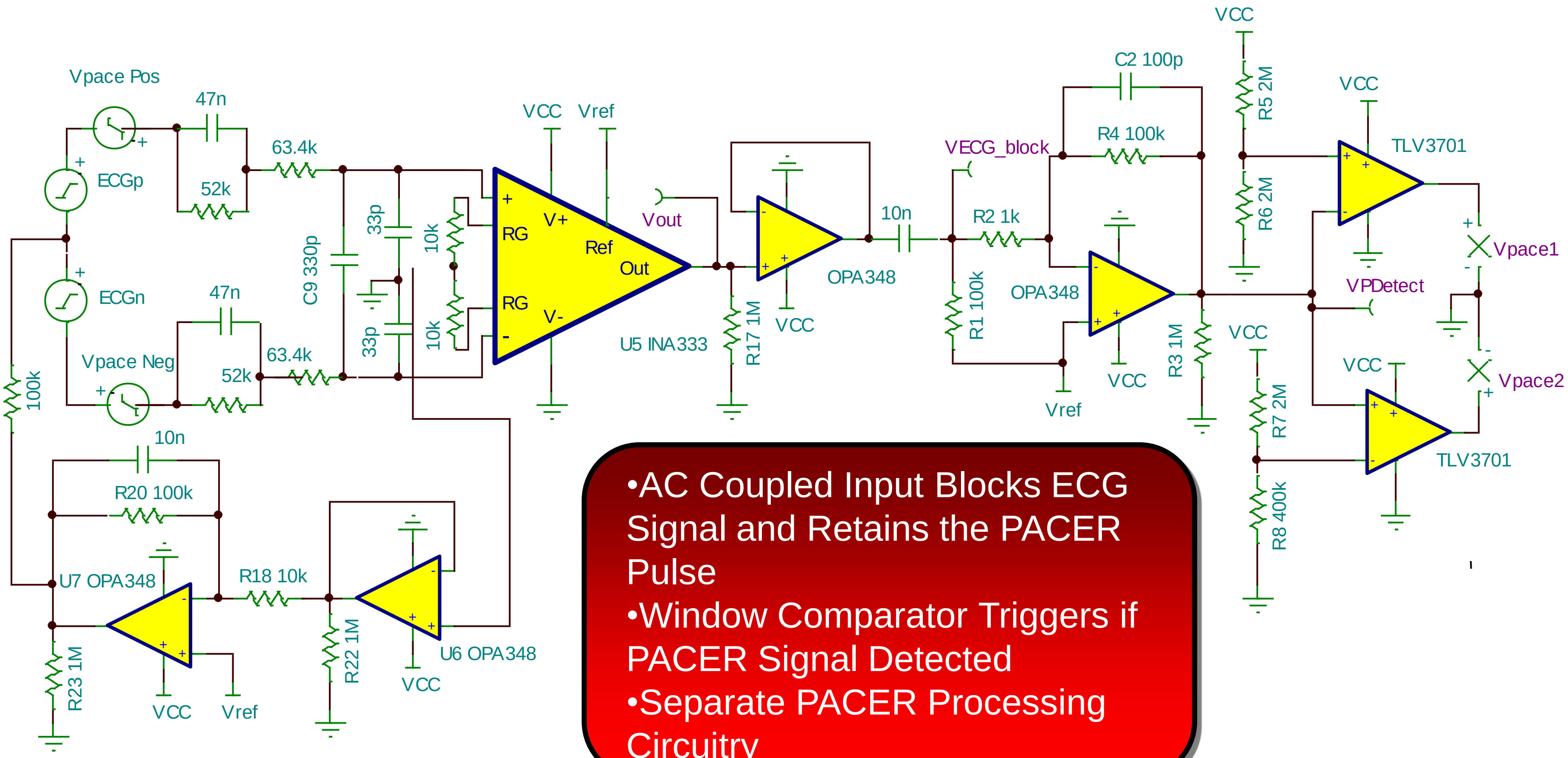
d_p = Pulse Width (.1-100us)

t_0 = Overshoot Time Constant (4-100ms)

Rise Time = 100us

Pace Detect

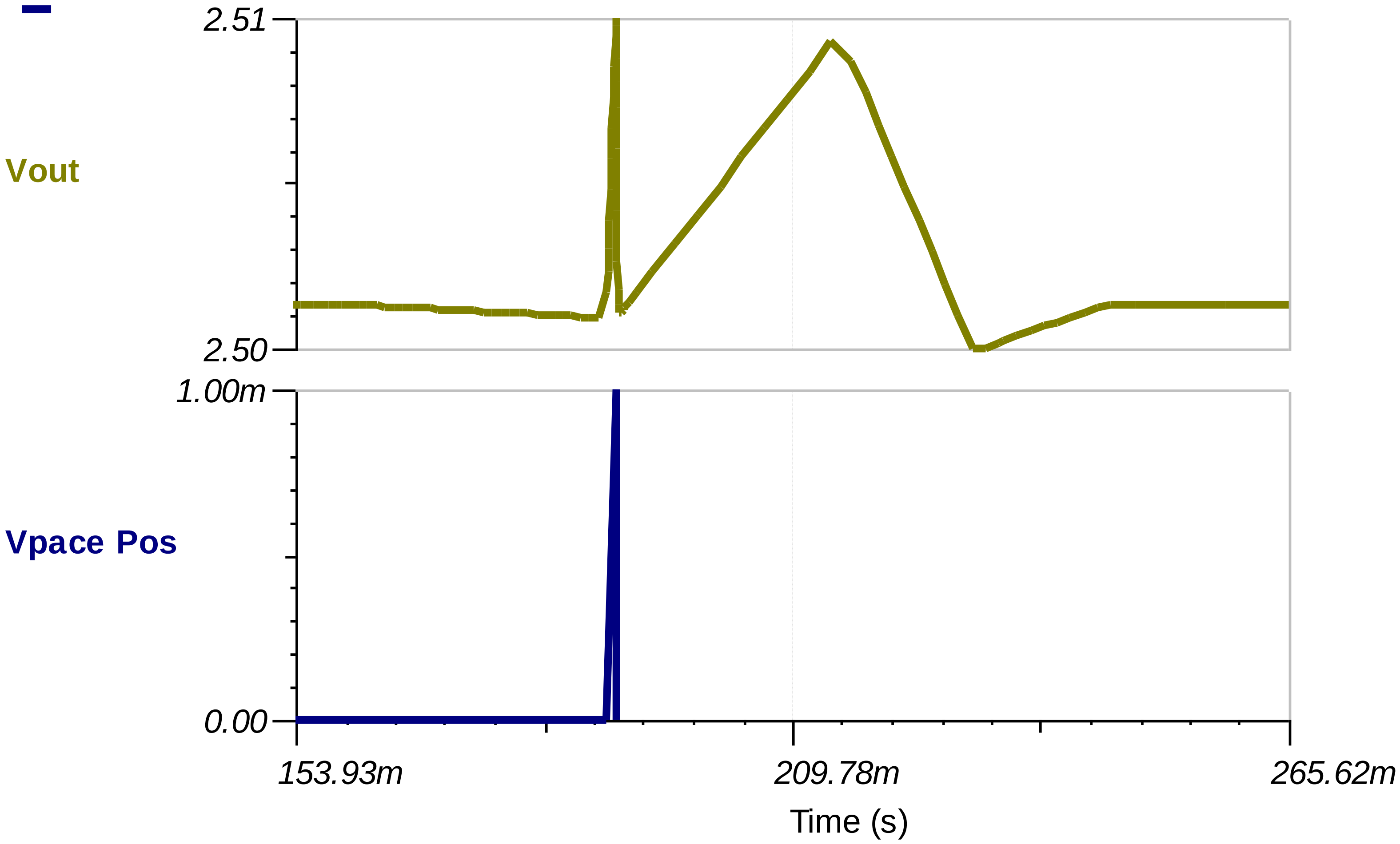
Pace Detect Circuitry in Parallel with ECG Signal Path



- AC Coupled Input Blocks ECG Signal and Retains the PACER Pulse
- Window Comparator Triggers if PACER Signal Detected
- Separate PACER Processing Circuitry

Pace Detect

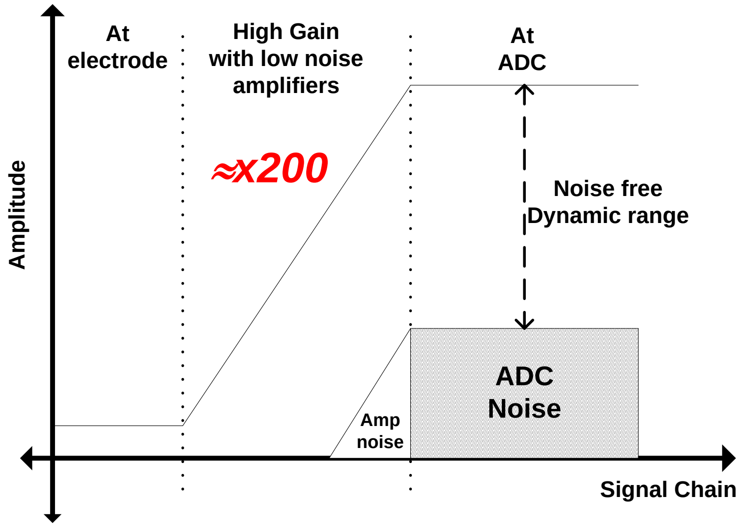
PACE Signal Extracted From PACE + ECG Waveform



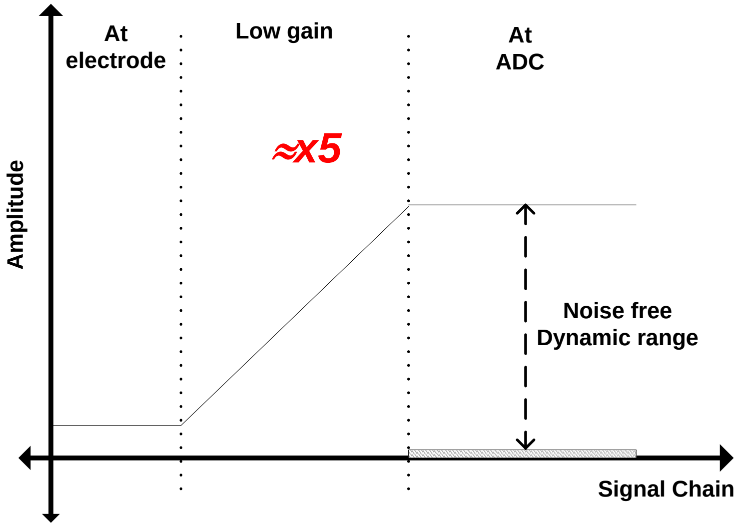
INA Post Gain and Filtering

INA Post Gain and Filtering

Choice of High Gain + SAR ADC OR Low Gain + 24 bit Delta Sigma ADC



a) Using a low resolution ADC

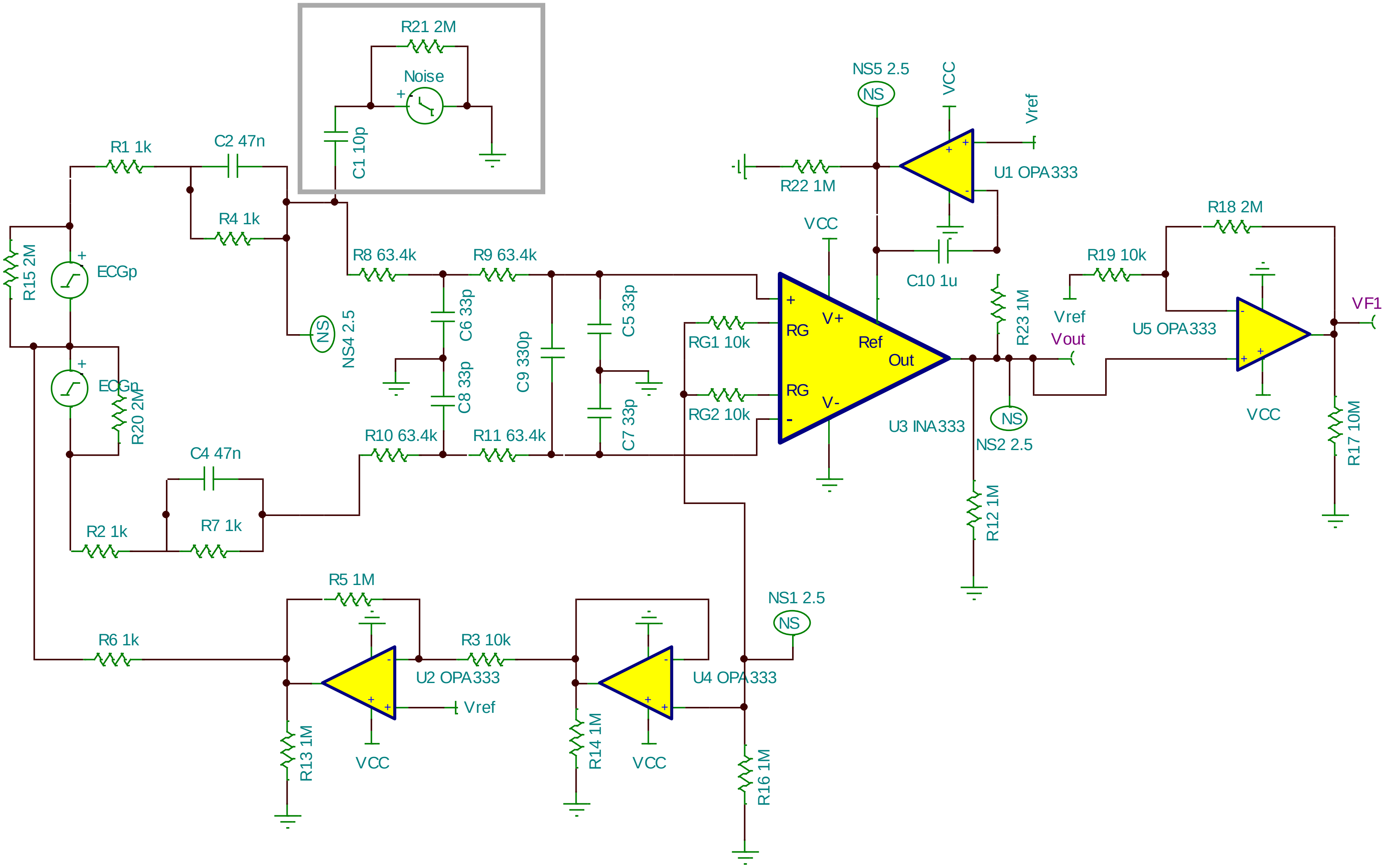


b) Using a high resolution ADC

SAR + filter Option Results in Same Input-Referred Noise as the DC Coupled Delta-Sigma, but at what COST?

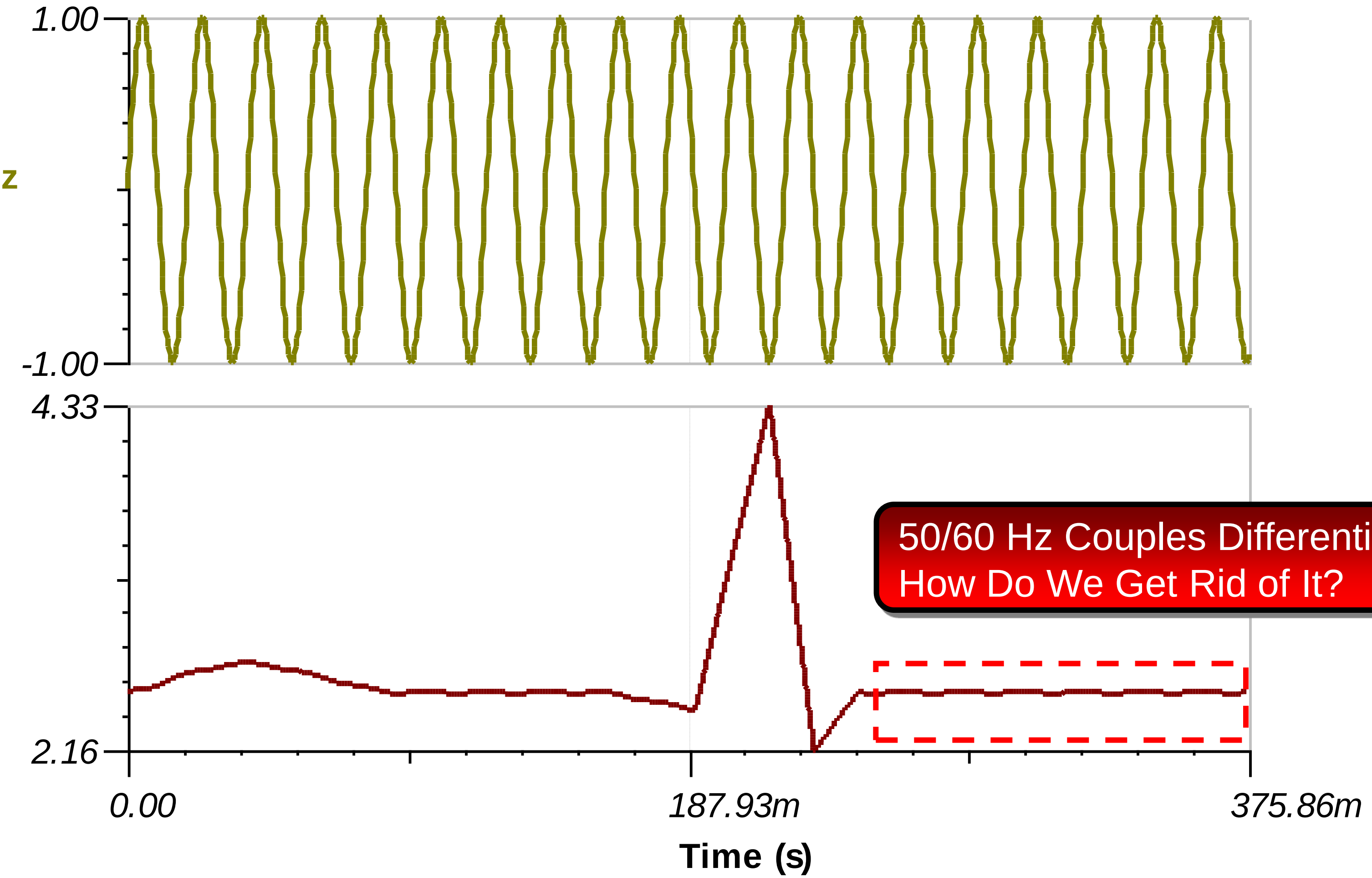
INA Post Gain and Filtering

INA + Post Gain Amp With Differential Noise Source



INA Post Gain and Filtering

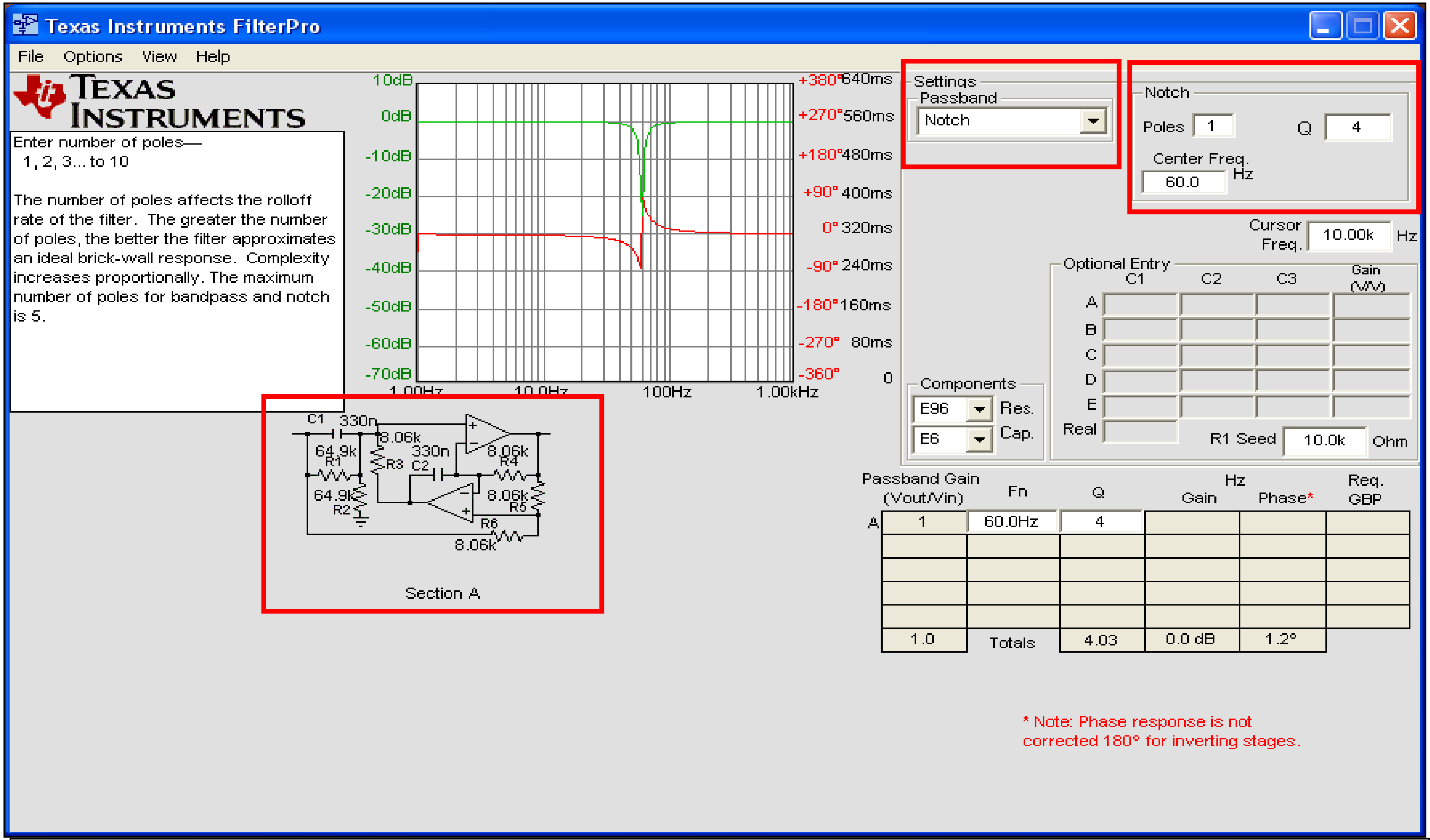
Noise Coupled Differentially Translates to Output



50/60 Hz Couples Differentially; How Do We Get Rid of It?

INA Post Gain and Filtering

Use Filter Pro to Design a 50/60 Hz Notch



Enter number of poles—
1, 2, 3... to 10

The number of poles affects the rolloff rate of the filter. The greater the number of poles, the better the filter approximates an ideal brick-wall response. Complexity increases proportionally. The maximum number of poles for bandpass and notch is 5.

Settings

Passband: Notch

Notch

Poles: 1

Q: 4

Center Freq. Hz: 60.0

Cursor Freq.: 10.00k Hz

Optional Entry

	C1	C2	C3	Gain (V/V)
A				
B				
C				
D				
E				
Real				

R1 Seed: 10.0k Ohm

Components

Res.: E96

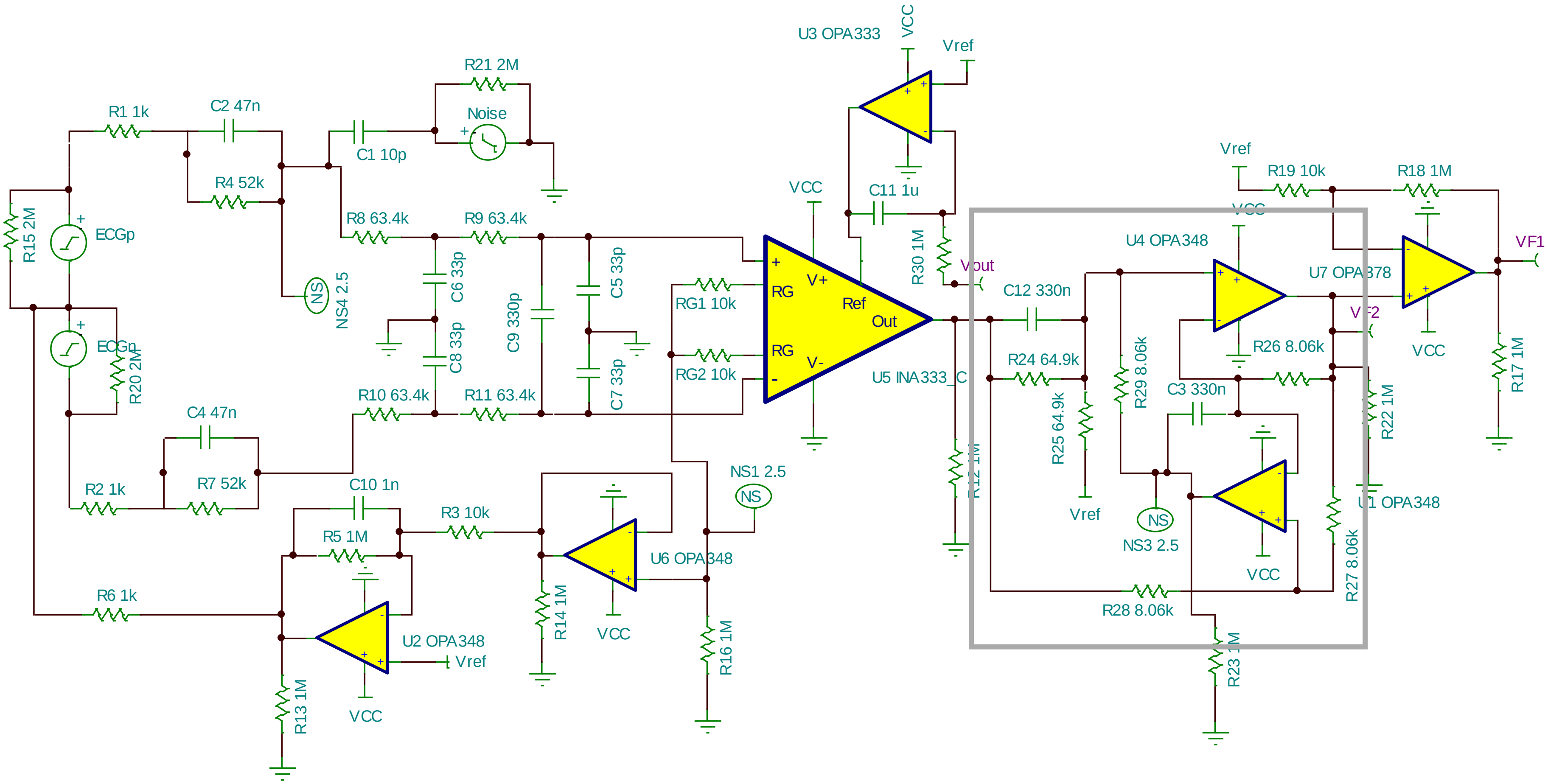
Cap.: E6

	Passband Gain (Vout/Vin)	Fn	Q	Hz Gain	Phase*	Req. GBP
A	1	60.0Hz	4			
	1.0	Totals	4.03	0.0 dB	1.2°	

* Note: Phase response is not corrected 180° for inverting stages.

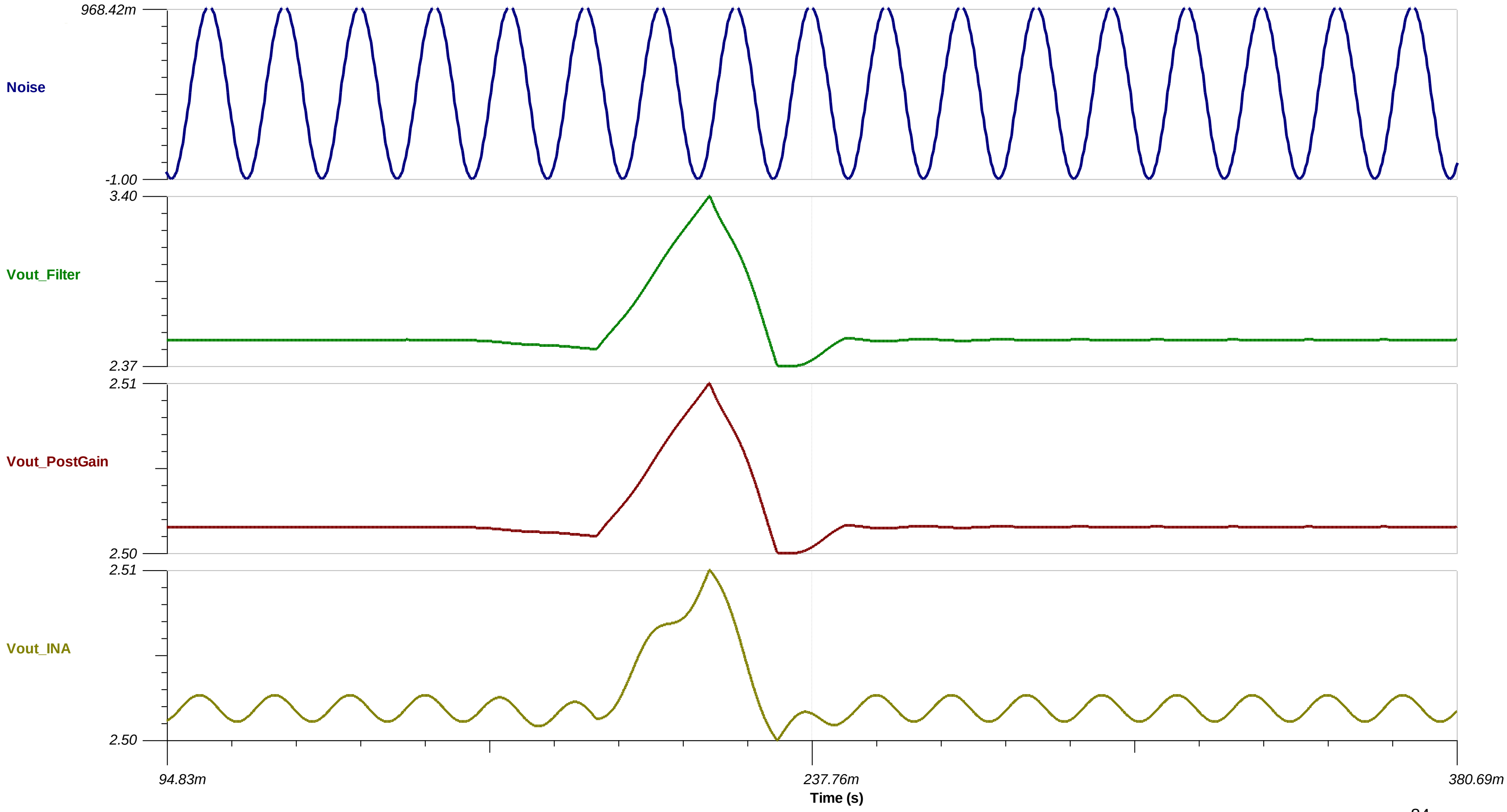
INA Post Gain and Filtering

ECG Circuit with Added 50/60Hz Notch + Post Gain



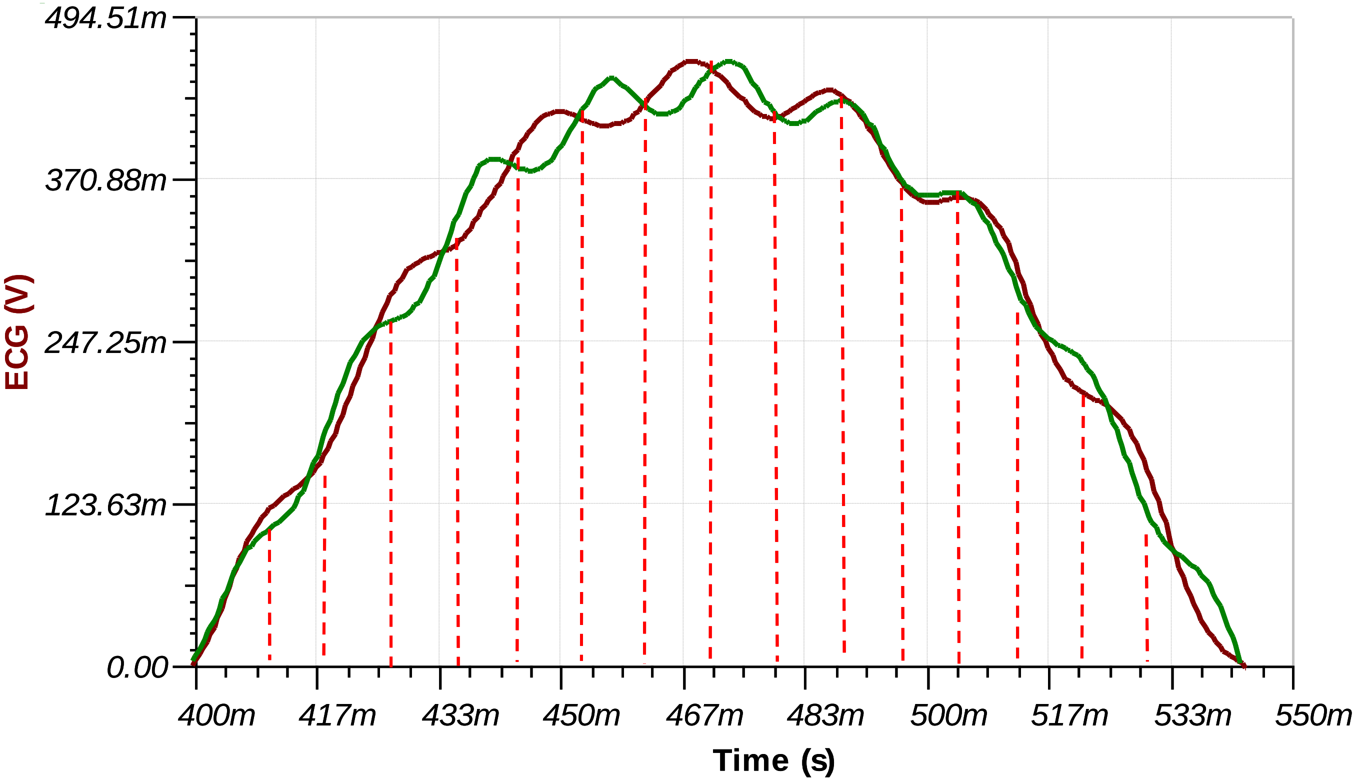
INA Post Gain and Filtering

Plots of ECG Output with Gain and 60Hz Notch



INA Post Gain and Filtering

Line Cycle Sampling with SAR converter on 'T' Wave at Common Frequency Multiples of 50/60Hz

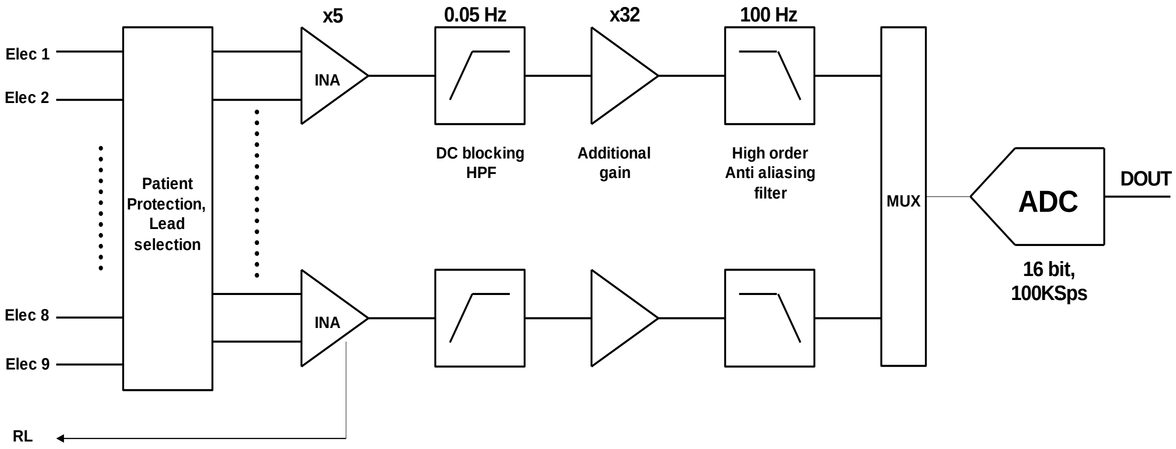


$$F_{sample}\ (\text{Hz}) = n^*(1/50\text{Hz} + 1/60\text{Hz})^{-1} = n^*(27.27)\ \text{Hz}$$

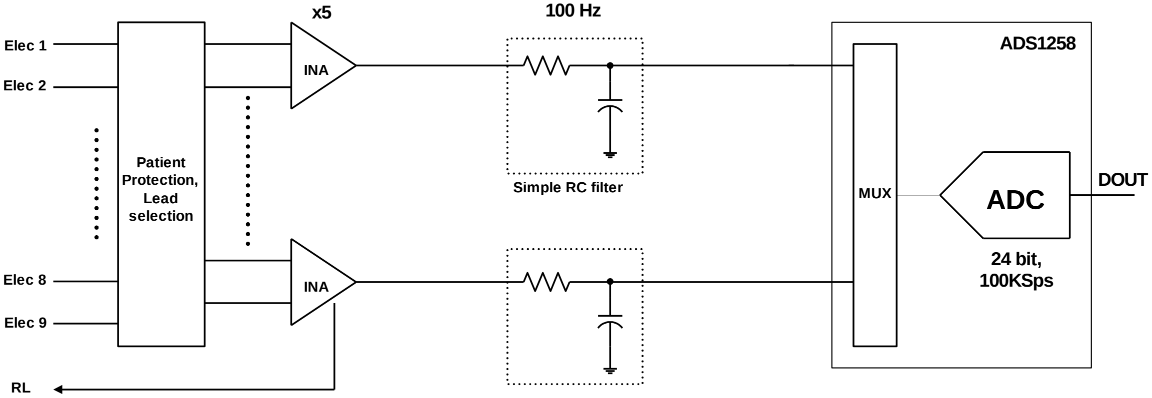
INA Post Gain and Filtering

Comparison of Delta Sigma ADC vs. Lower Resolution SAR ADC

Using a low resolution ADC00



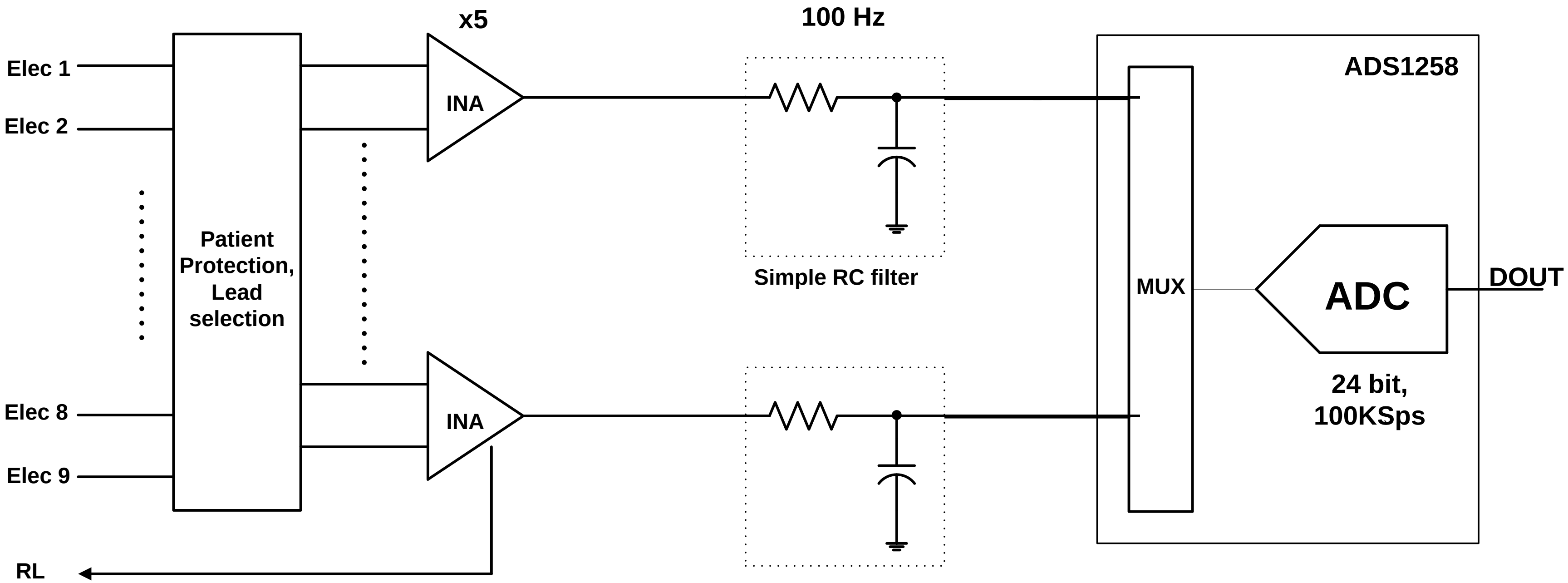
Using a high resolution ADC



- Reduced Hardware
- Filter Requirements Relaxed
- Lower Power
- Lower System Cost
- Electrode Offset Info Retained

INA Post Gain and Filtering

Block Diagram of INA Gain, Simple RC Filter, and ADS1258

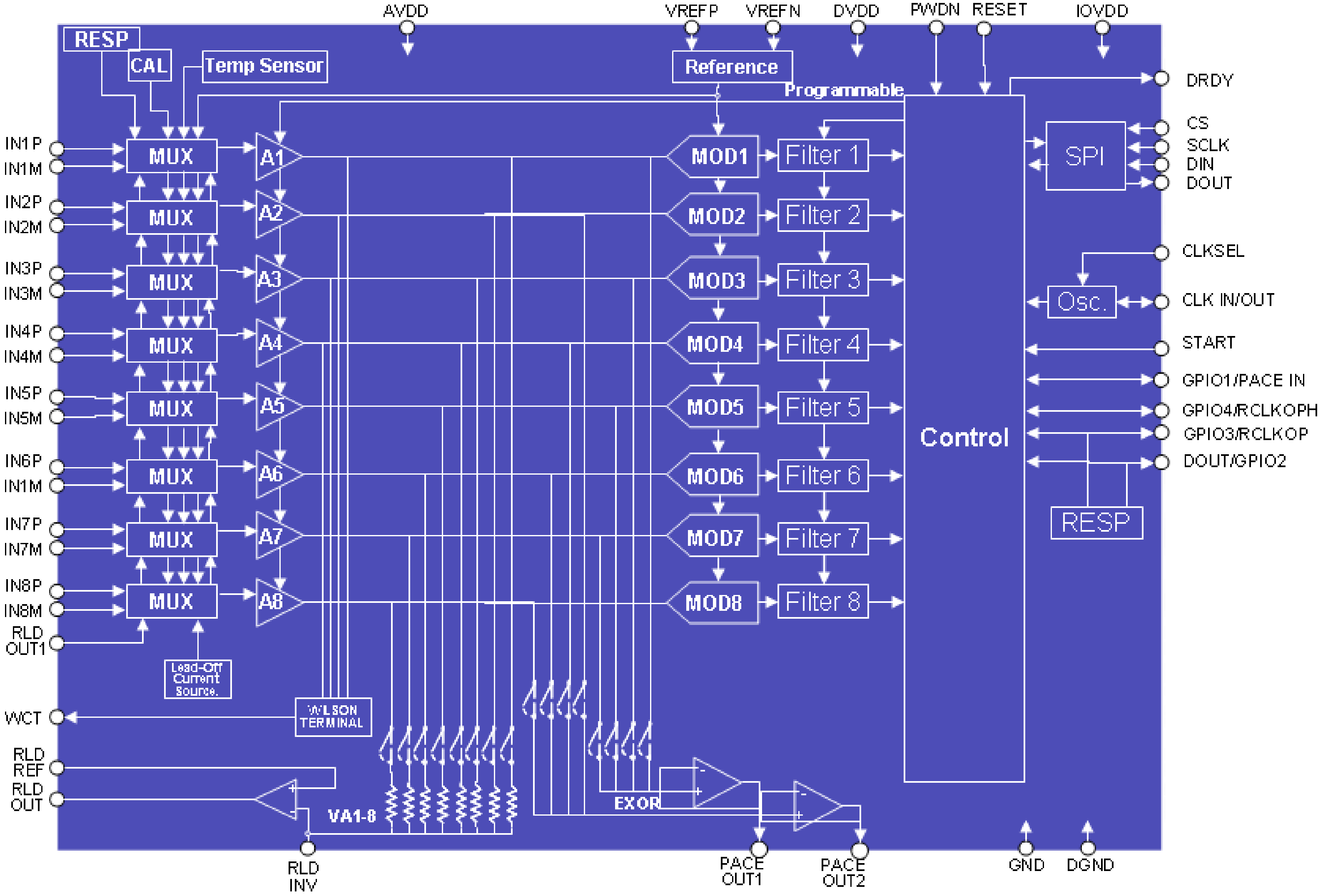


A single ADC in the MUX approach does not necessarily mean lower power due to the higher speed needed to perform MUX switching

ADS1298 Introduction

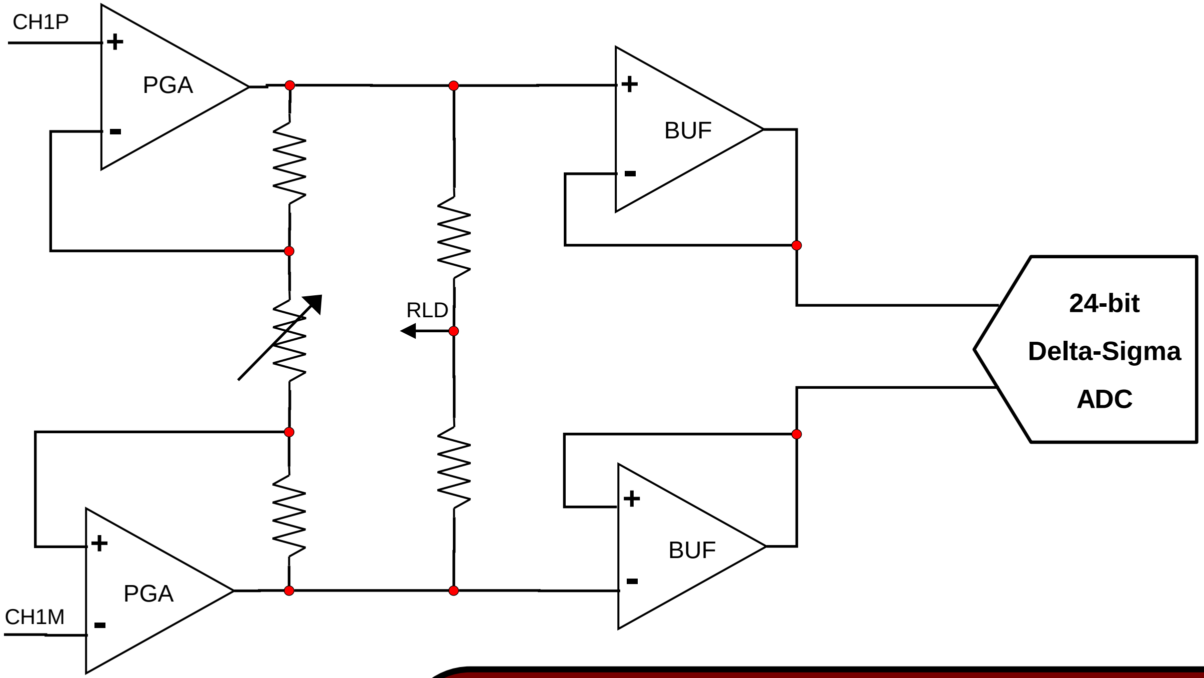
The ADS129x

The All-In-One ECG Chip



ADS129x

Input Amplifier Specifications for Single Channel AFE



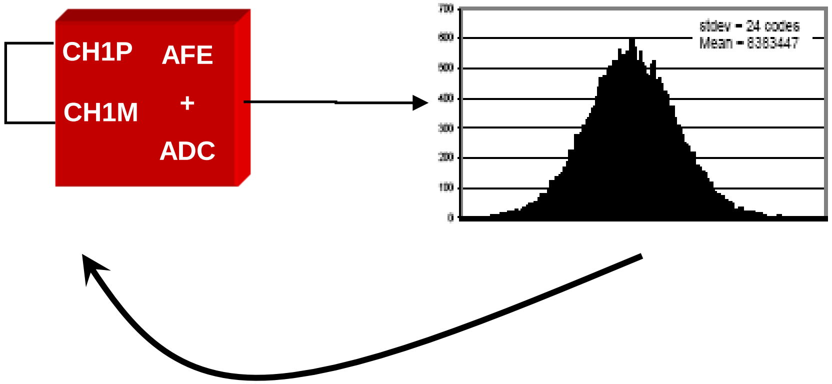
- ✓CMOS input PGA
- ✓High input impedance
- ✓Low input current noise
- ✓Rail to rail input

- •Low Input Voltage Noise
- •Current Noise = 0.1pA√Hz Over Bandwidth
- •No Output Phase Reversal
- •100dB AOL @ 50/60 Hz
- •90dB CMR @ 50/60 Hz
- •IB = 150pA MAX vs. Temperature
- •ZIN >100MΩ, CIN = 100pF max

ADS129x

Noise Model for ECG AFE

- Noise is optimized with amplifier gain=4
- The 4uV p-p includes the crest factor of 6.6 to convert rms to pk-pk
- Noise is referred to the input



ADS129x

Programmable Data Rates for Low Power and High Resolution Modes

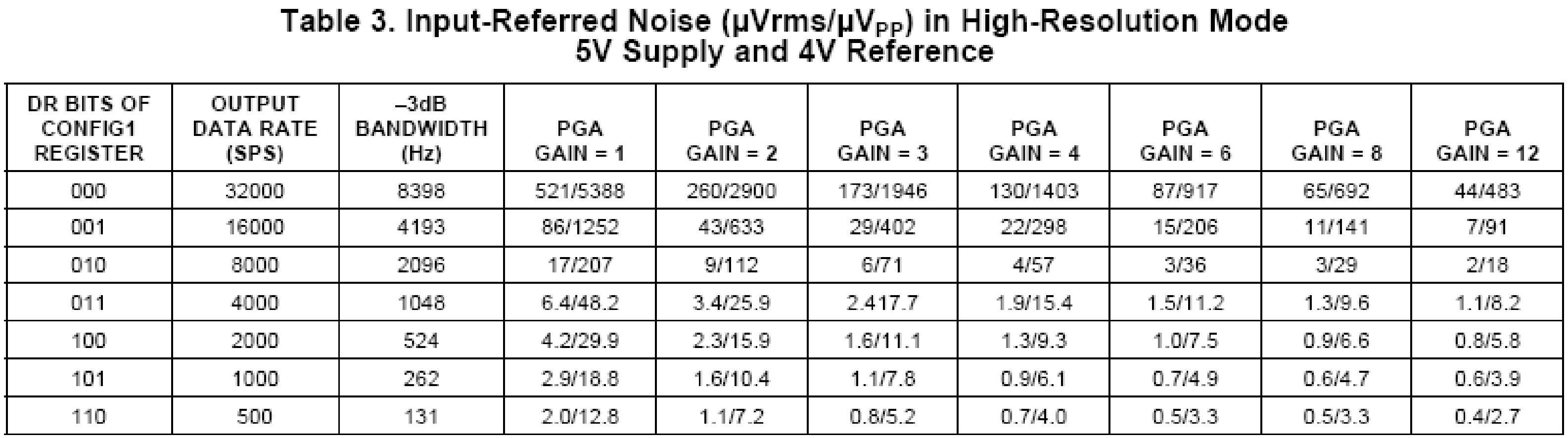


Table 3. Input-Referred Noise (μVrms/μV_{PP}) in High-Resolution Mode
5V Supply and 4V Reference

DR BITS OF CONFIG1 REGISTER	OUTPUT DATA RATE (SPS)	–3dB BANDWIDTH (Hz)	PGA GAIN = 1	PGA GAIN = 2	PGA GAIN = 3	PGA GAIN = 4	PGA GAIN = 6	PGA GAIN = 8	PGA GAIN = 12
000	32000	8398	521/5388	260/2900	173/1946	130/1403	87/917	65/692	44/483
001	16000	4193	86/1252	43/633	29/402	22/298	15/206	11/141	7/91
010	8000	2096	17/207	9/112	6/71	4/57	3/36	3/29	2/18
011	4000	1048	6.4/48.2	3.4/25.9	2.417.7	1.9/15.4	1.5/11.2	1.3/9.6	1.1/8.2
100	2000	524	4.2/29.9	2.3/15.9	1.6/11.1	1.3/9.3	1.0/7.5	0.9/6.6	0.8/5.8
101	1000	262	2.9/18.8	1.6/10.4	1.1/7.8	0.9/6.1	0.7/4.9	0.6/4.7	0.6/3.9
110	500	131	2.0/12.8	1.1/7.2	0.8/5.2	0.7/4.0	0.5/3.3	0.5/3.3	0.4/2.7

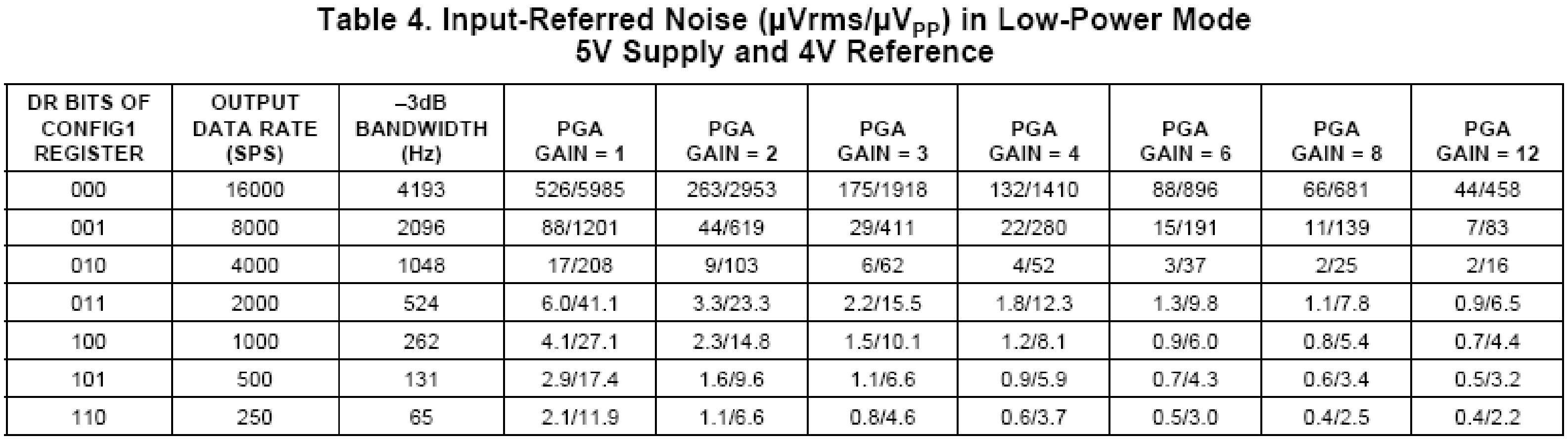
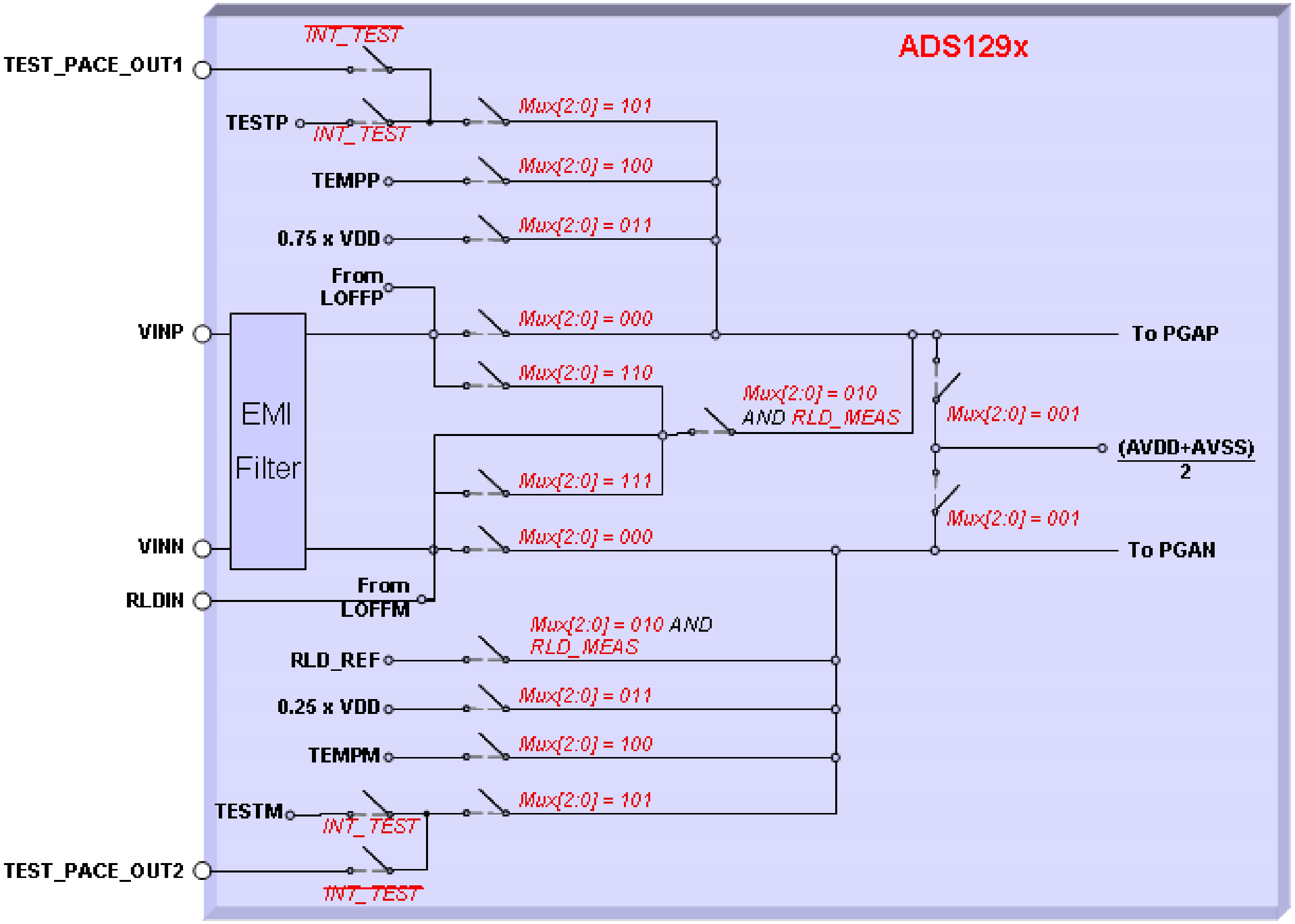


Table 4. Input-Referred Noise (μVrms/μV_{PP}) in Low-Power Mode
5V Supply and 4V Reference

DR BITS OF CONFIG1 REGISTER	OUTPUT DATA RATE (SPS)	–3dB BANDWIDTH (Hz)	PGA GAIN = 1	PGA GAIN = 2	PGA GAIN = 3	PGA GAIN = 4	PGA GAIN = 6	PGA GAIN = 8	PGA GAIN = 12
000	16000	4193	526/5985	263/2953	175/1918	132/1410	88/896	66/681	44/458
001	8000	2096	88/1201	44/619	29/411	22/280	15/191	11/139	7/83
010	4000	1048	17/208	9/103	6/62	4/52	3/37	2/25	2/16
011	2000	524	6.0/41.1	3.3/23.3	2.2/15.5	1.8/12.3	1.3/9.8	1.1/7.8	0.9/6.5
100	1000	262	4.1/27.1	2.3/14.8	1.5/10.1	1.2/8.1	0.9/6.0	0.8/5.4	0.7/4.4
101	500	131	2.9/17.4	1.6/9.6	1.1/6.6	0.9/5.9	0.7/4.3	0.6/3.4	0.5/3.2
110	250	65	2.1/11.9	1.1/6.6	0.8/4.6	0.6/3.7	0.5/3.0	0.4/2.5	0.4/2.2

ADS129x

MUX Selects Inputs to Front End PGA



- Normal Electrode
- Input Shorted
- RLD Input
- VDD
- TMP Sensor
- Input Test Signal

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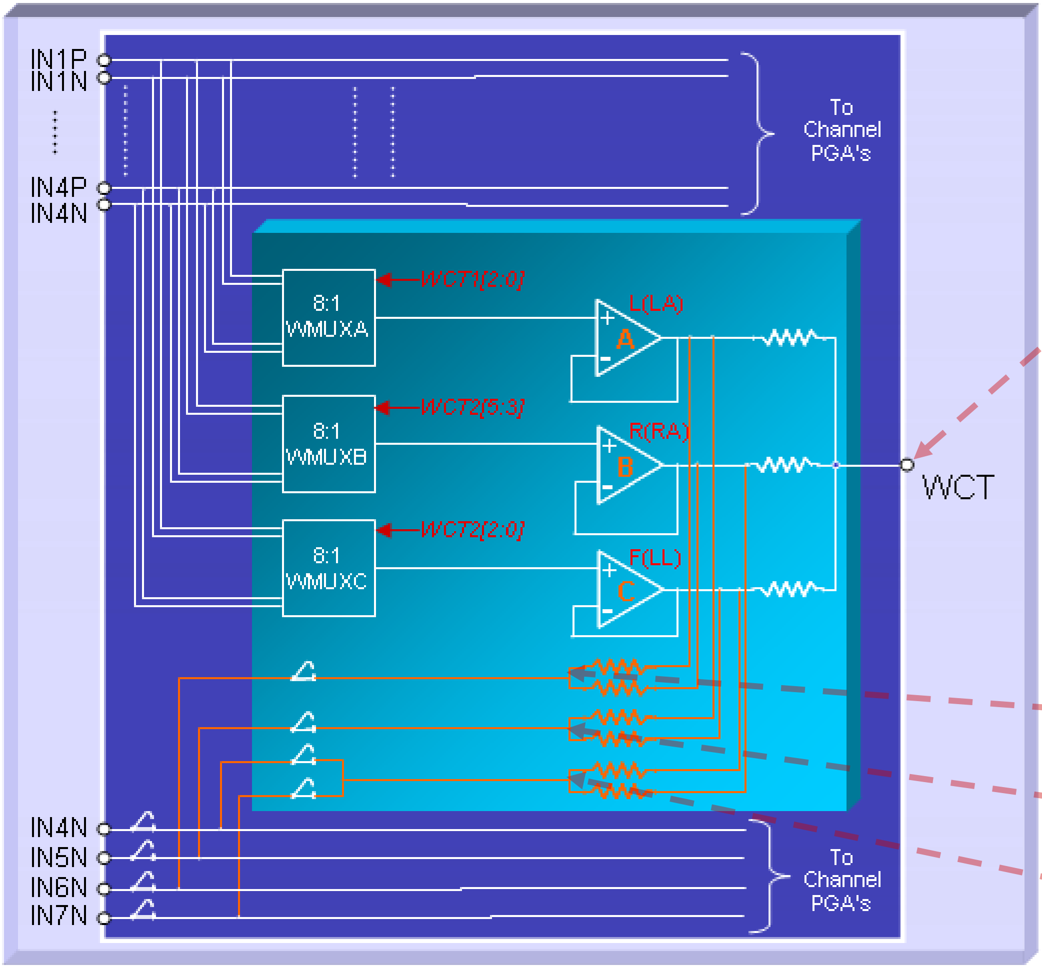
Wilson Central Terminal



Wilson Central Voltage

0.333(RA+LA+LL)

Augmented Lead Voltages

0.5(RA+LA)

0.5(LA+LL)

0.5(RA+LL)

*The Same Amplifiers Used to Derive the WCT Voltage Can be Switched to Obtain the Augmented Leads

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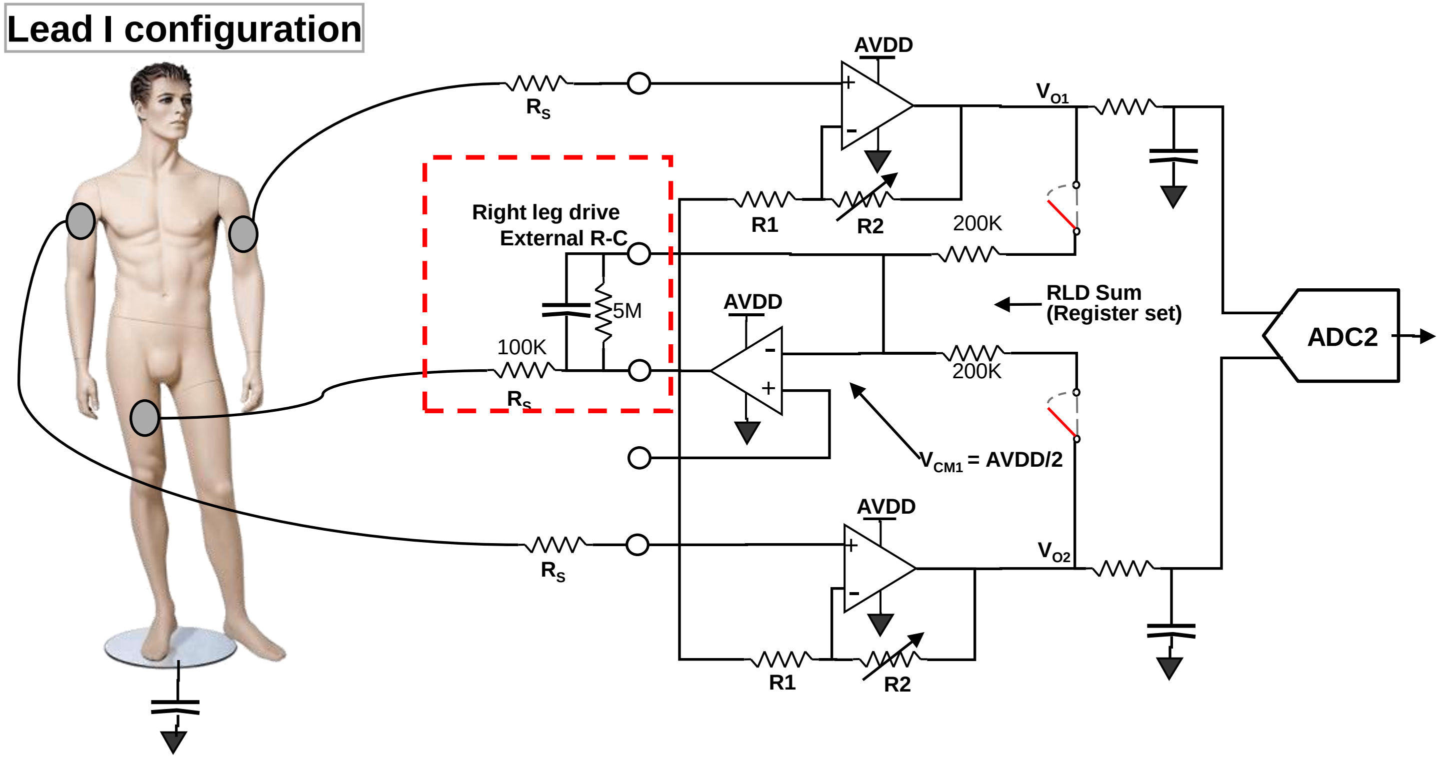
Input Amplifier and RLD Selection



*Compensation of RLD Amplifier is Based on the Gain Selected and the Number of Amplifiers in the Loop

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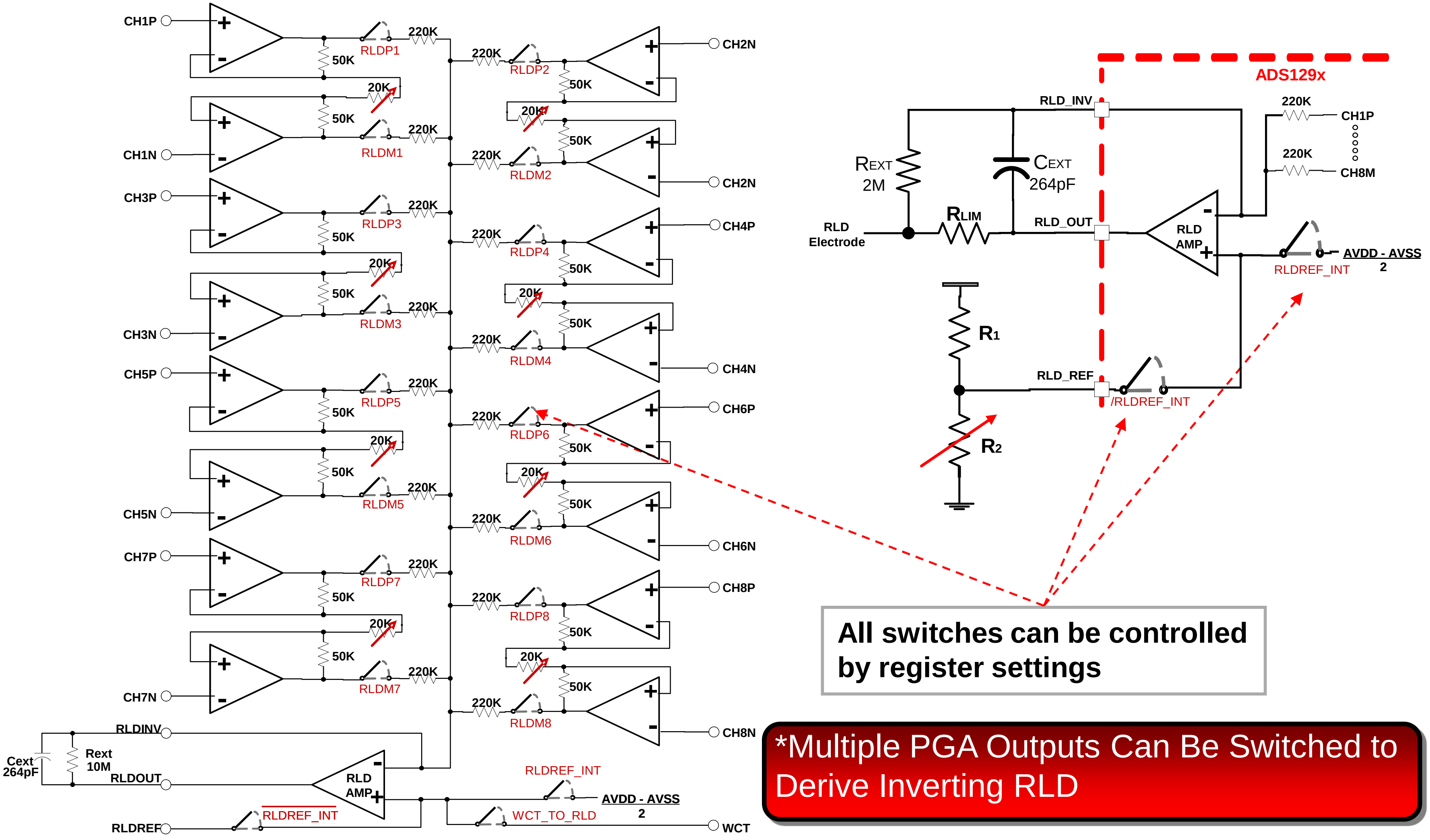
RLD Selection—8 Channel Case



All switches can be controlled by register settings

*Multiple PGA Outputs Can Be Switched to Derive Inverting RLD

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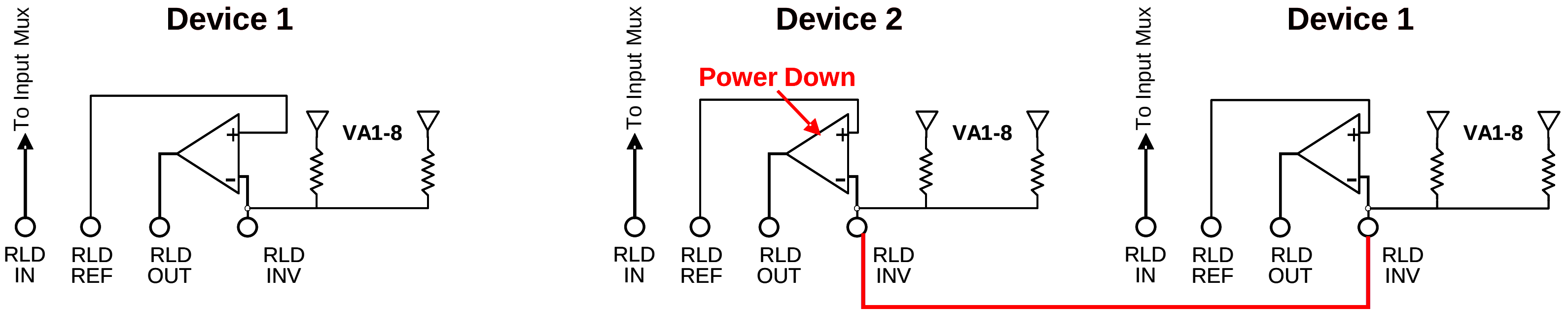
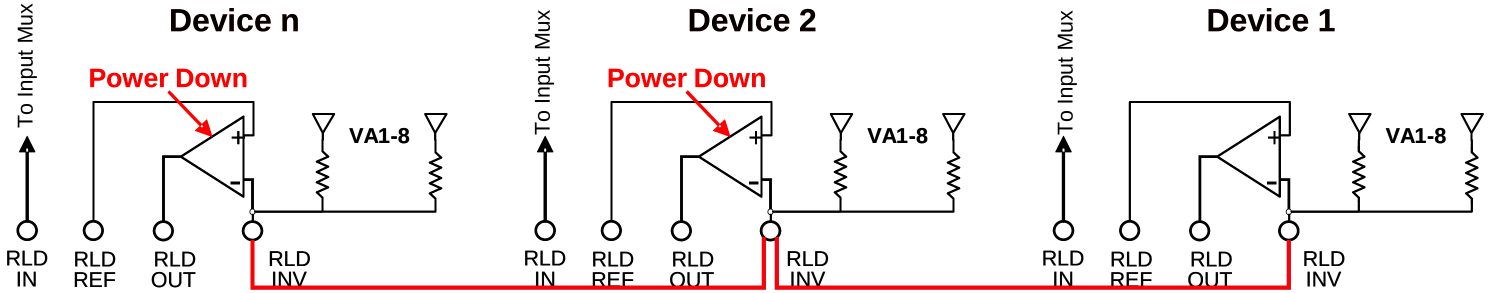
RLD with Multiple Devices





*With Multiple Devices the RLD Output Becomes the Amplified Difference Between RLD REF and the Summation of Multiple Lead Outputs

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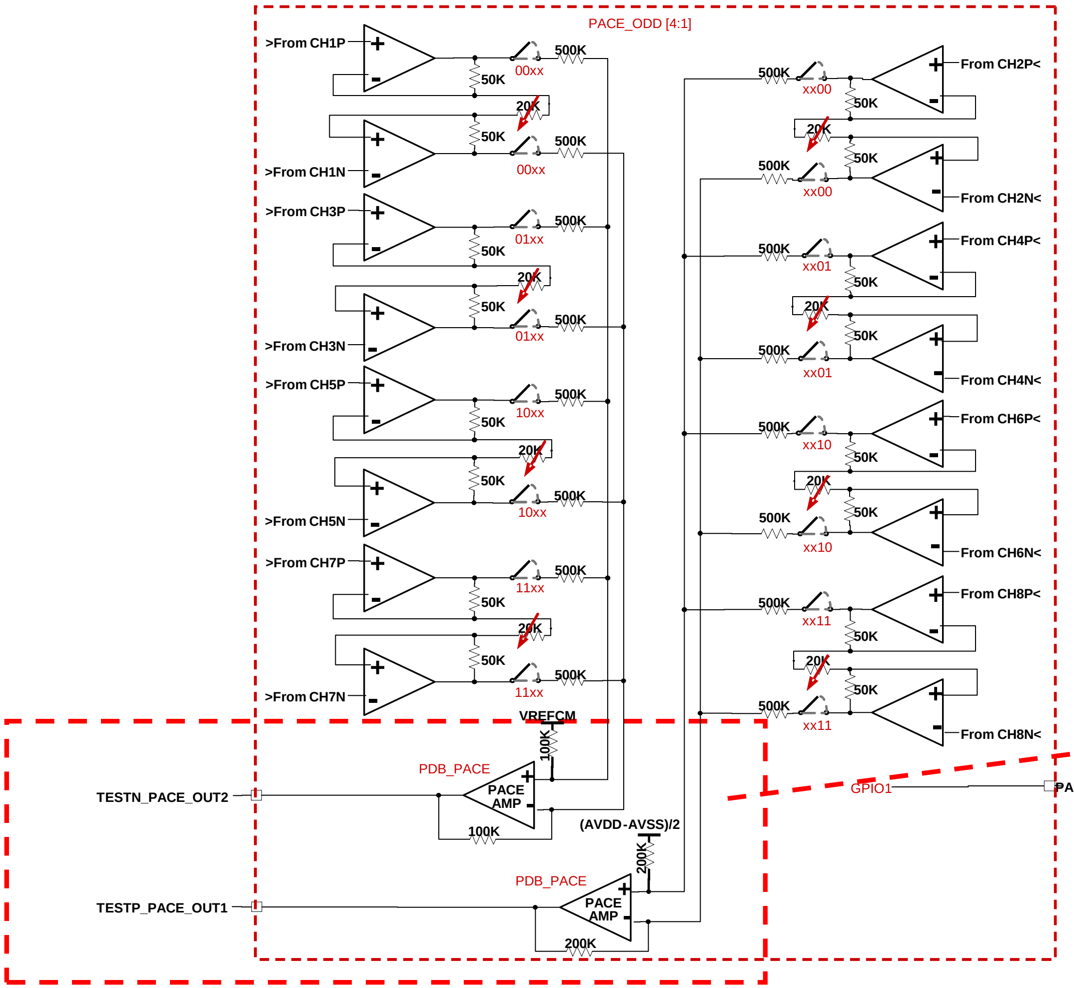
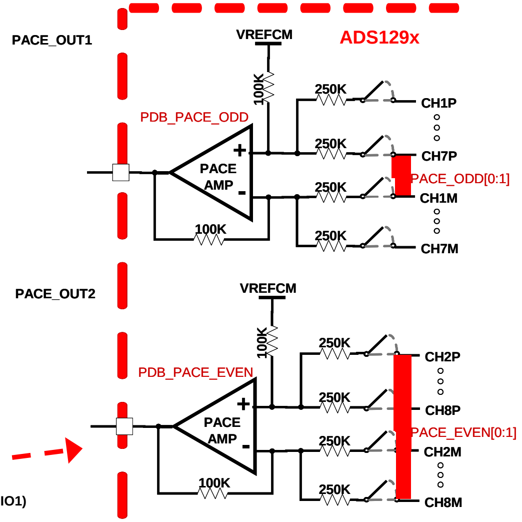
Pace Detect



*Separate Pace Amplifiers Allow External Processing of Pace Signal; All Channels do Not Have to Operate at Higher Data Rates



PACE ODD		PACE EVEN		CHANNEL SELECTED
PACEO1	PACEO0	PACEE1	PACEE0	
x	x	0	0	Channel #1
0	0	x	x	Channel #2
x	x	0	1	Channel #3
0	1	x	x	Channel #4
x	x	1	0	Channel #5
1	0	x	x	Channel #6
x	x	1	1	Channel #7
1	1	x	x	Channel #8

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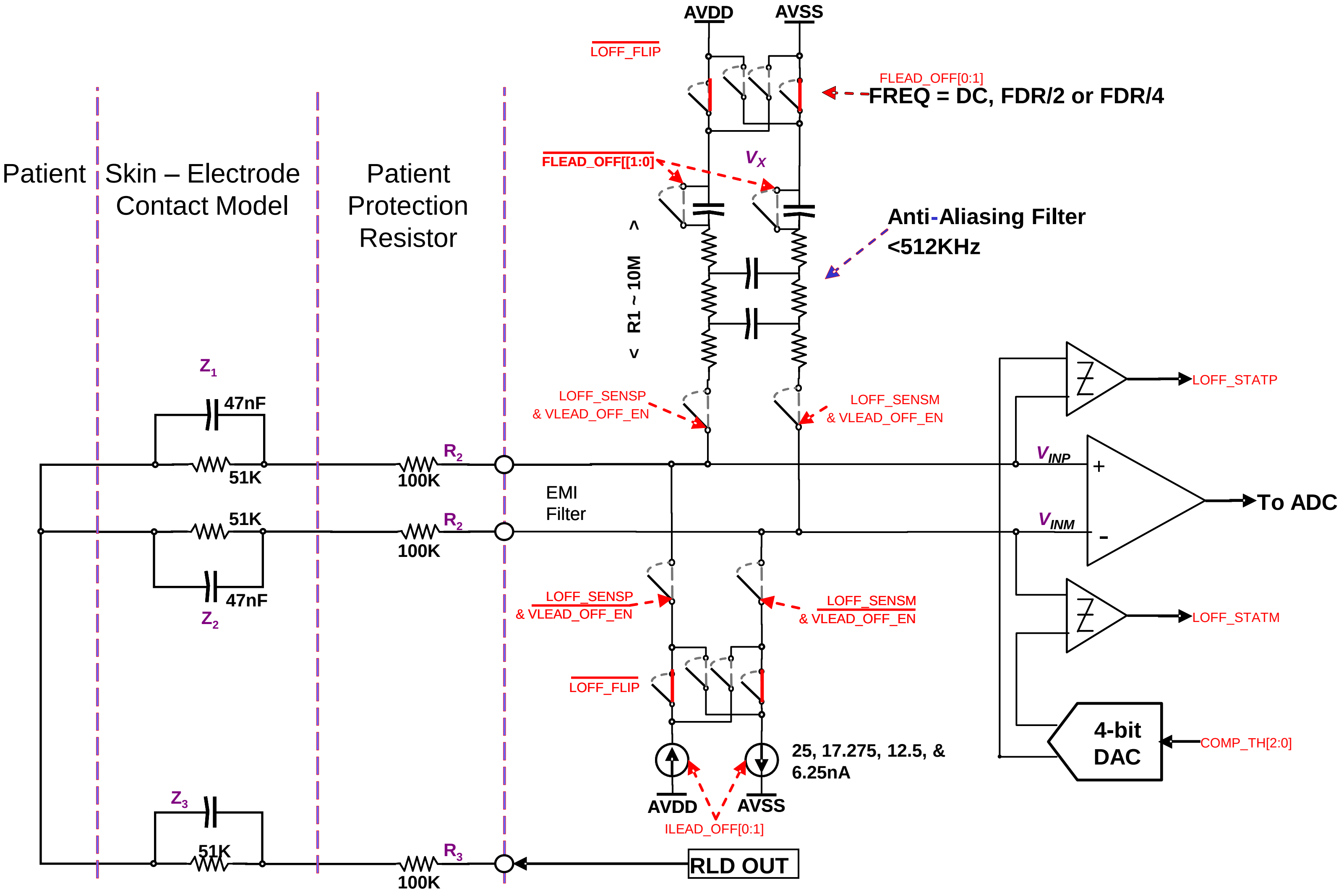
Lead Off Detection



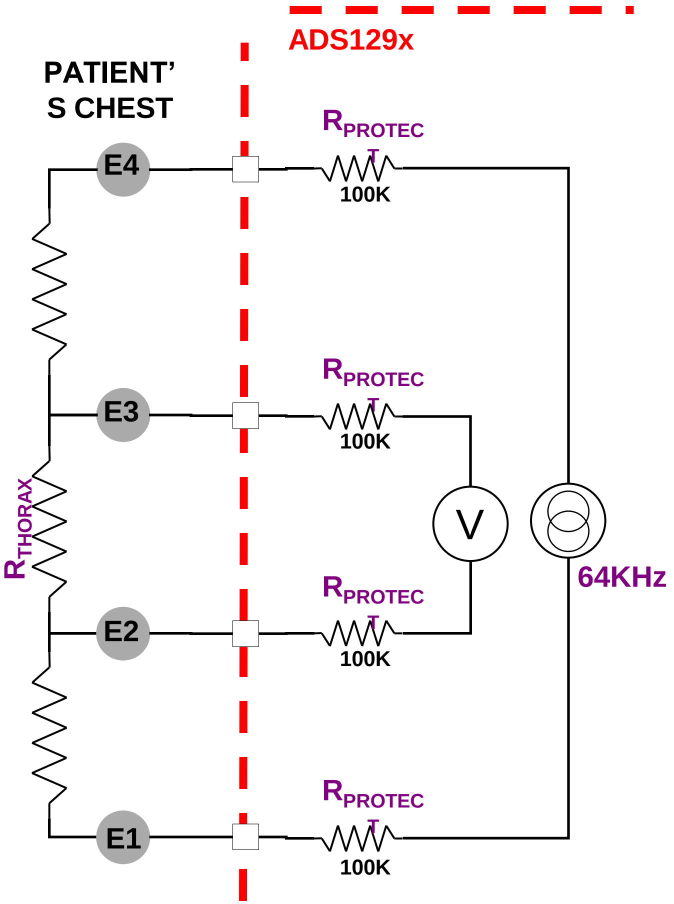
Patient

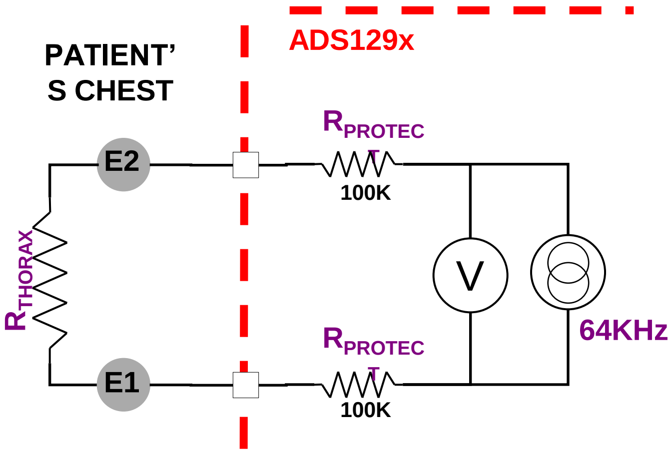
Skin – Electrode Contact Model

Patient Protection Resistor

ADS129x

Respiration Testing Measures the Change in Thoracic Impedance with Inhalation of O_2





*AC Current is injected into the Patient's Thorax and the Change in Voltage is Measured to Calculate Change in Impedance

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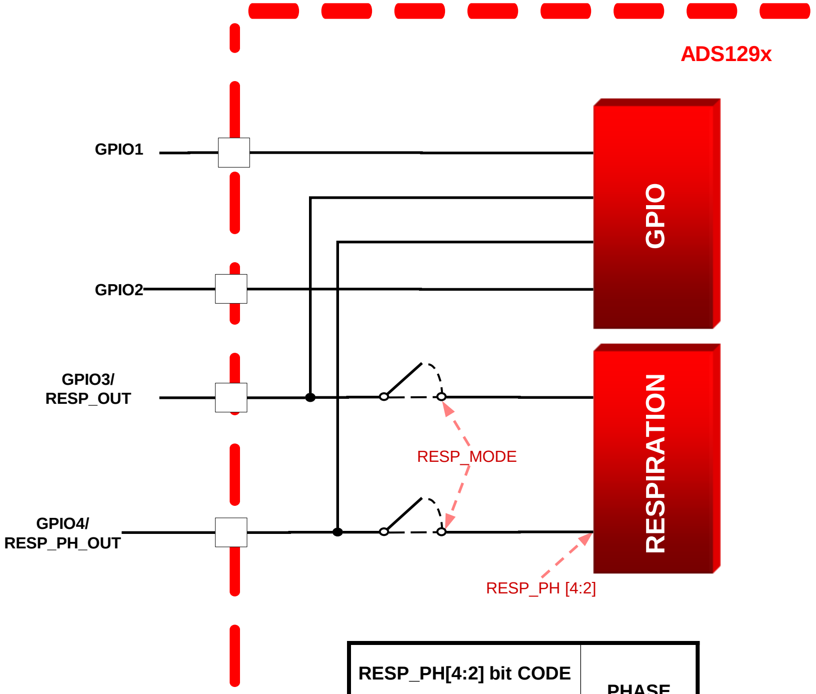
Respiration Functions



RESP_PH[4:2] bit CODE			PHASE SELECTED
RESP_PH2	RESP_PH1	RESP_PH0	
0	0	0	22.5 Deg
0	0	1	45 Deg
0	1	0	67.5 Deg
0	1	1	90 Deg
1	0	0	112.5 Deg
1	0	1	135 Deg
1	1	0	157.5 Deg
1	1	1	180 Deg

Changing Phase Allows Measurement/Compensation for Complex Impedance Phase Shifts Between Modulator and Demodulator

ADS129x

Internal Voltage Reference

Simplified ADS129x internal reference block diagram

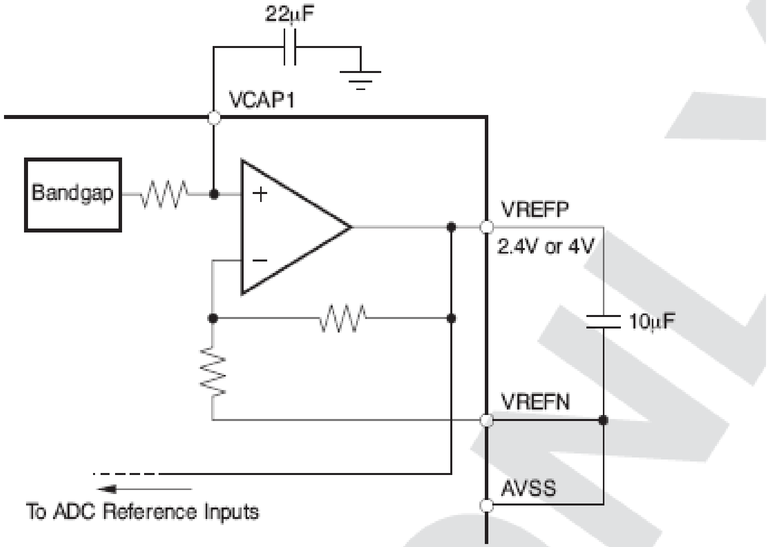


Figure 21. Internal Reference

The Internal Band Gap Accuracy = 1%

Internal REF can be Powered Down

VREFP Can Be Supplied Externally

Thank You

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Questions?

Acknowledgements

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