A Comprehensive Model for Power Line Interference in Biopotential Measurements

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Abstract—**Power line interference is a major problem in high-resolution biopotential measurements. Because interference coupling is mostly capacitive, shielding electrode leads and a high common-mode rejection ratio (CMRR) are quite effective in reducing power-line interference but do not completely eliminate it. We propose a model that includes both interference external to the measuring system and interference coming from its internal power supply. Moreover, the model considers interference directly coupled to the measuring electrodes, because, as opposed to connecting leads, electrodes are not usually shielded. Experimental results confirm that reducing interference coupled through electrodes yields a negligible interference. The proposed model can be applied to other differential measurement systems, particulary those involving electrodes or sensors placed far apart.**

Index Terms—**Amplifier, biomedical measurements, electrodes, interference, shielding.**

I. INTRODUCTION

POWER line interference is often a nuisance in biopotential measurements, mostly because of the long wires between the subject and the amplifier, the separation between the measurement points (electrodes), capacitive coupling between the subject (a volume conductor) and power lines, and the low amplitude of the desired signals. High-resolution measurements searching for potentials as small as 1 μV further exacerbate the problem.

Huhta and Webster [1] analyzed power line interference in three-electrode ECG recordings with grounded amplifiers. They identified four different contributions

1) magnetic induction in input leads;
2) displacement currents in those leads;
3) displacement currents in the body;
4) common-mode voltage contribution because of the amplifier's limited common-mode rejection ratio (CMRR) compounded by electrode and common-mode input impedance imbalance. They recommended twisting input leads together and using a right-leg drive circuit to reduce the common-mode voltage.

Thakor and Webster [2] analyzed power line interference in two-electrode ECG recordings. Ground-free amplifiers are safer than grounded amplifiers, and two-electrode amplifiers are common in biotelemetry and ambulatory monitoring. They realized that in two-electrode amplifiers the interference is larger in grounded amplifiers than in ground-free amplifiers.

Hence, two-electrode grounded amplifiers need high CMRR and common-mode input impedance.

Winter and Webster [3] considered interference reduction in both isolated and nonisolated amplifiers. They proposed to reduce interference by increasing the amplifier's effective CMRR or by reducing the common-mode voltage, v_{cm}, for example by increasing the isolation impedance.

Pallás-Areny [4] compared the interference-rejection characteristics of two- and three-electrode amplifiers, both isolated and nonisolated. He proposed the effective coupling impedance concept (Z_{CE}) to describe the external interference coupled into the patient-amplifier system and to compare different amplifiers. Z_{CE} includes the limited isolated-mode rejection ratio (IMRR), needed to account for the relatively large interference observed in amplifiers with high CMRR in the presence of a small v_{cm}. He concluded that isolated amplifiers, needed to ensure patient safety, only help in interference reduction if their IMRR is high enough.

Meeting van Rijn *et al.* [5] corroborated the importance of capacitive coupling from power lines to electrode wires and considered an additional interference in isolation amplifiers: displacement currents coupled to the (isolated) amplifier common. The same authors further demonstrated the need of a very high IMRR [6].

Wood *et al.* [7] analyzed power line interference in two- and three-electrode biopotential amplifiers. They pointed to the limited effect of shielded cables because shields did not extend to the electrodes. Also, their simulations showed that the interference from the potential across the isolation barrier was negligible for $\mathrm{IMRR} = 130$ dB, which is understandable.

Weicher [8] analyzed electrical interference in instruments. He mentioned the need to consider radiated interference coupled via power-supply wires, but otherwise did not provide a model for the effect of that interference on the measurement. The effect of currents injected from the instrument to the measurement circuit in instruments with guard terminal is well known [9]. Nevertheless, in measurement systems using electrodes it is not possible to have a low-impedance guard connection because of the high electrode impedance involved.

We propose an interference model that includes the main coupling mechanisms described in the bibliography and adds an internal interference arising from the amplifier's power supply. The model separates coupling to the electrodes from coupling to their wires because shielded cables do not reduce displacement currents coupled to the electrodes. Our model applies to any measurement system with differential inputs connected to a volume conductor such as in soil or large-sample impedance measurements, or to distant sensors.
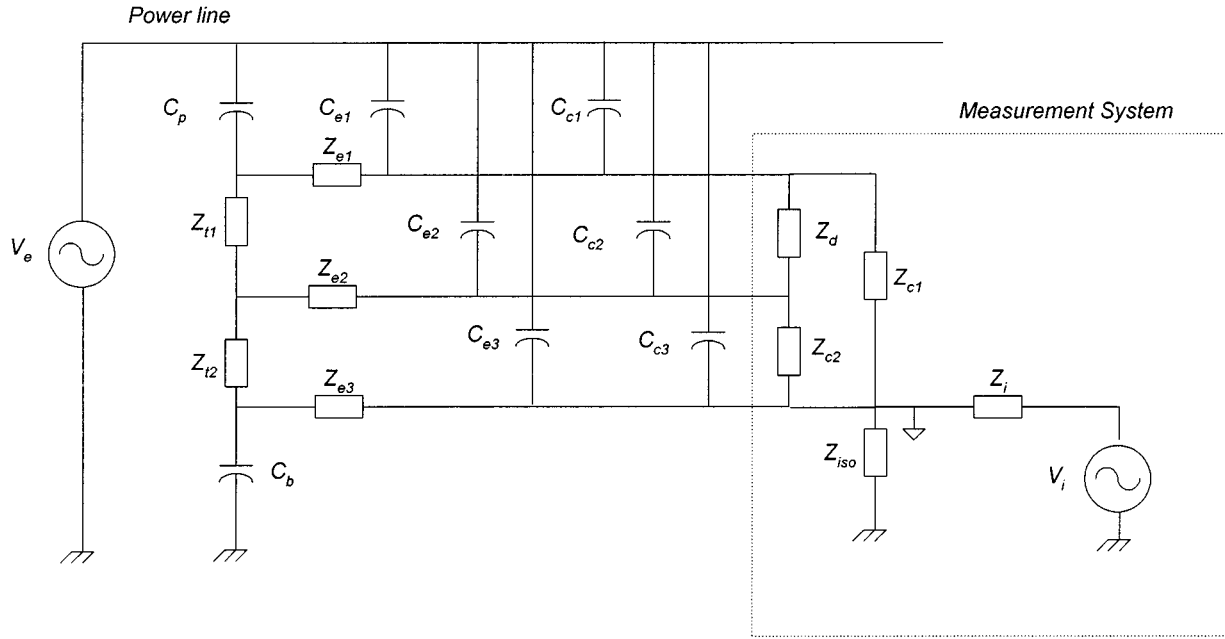
Fig. 1. Capacitive coupling to patient, electrodes, and electrode leads, and internal interference in a three-electrode isolated biopotential measurement system. Shielded electrode leads reduce capacitive coupling to them, but not capacitive coupling to electrodes. The same model applies to measurements involving electrodes in other volume conductors.

II. PROPOSED MODEL

Fig. 1 shows the proposed model for power line interference in a three-electrode, isolated biopotential measurement system. V_e is the power line voltage; Z_{t1} and Z_{t2} are the thorax impedances; Z_{e1}, Z_{e2}, and Z_{e3} are electrode impedances; the power line to patient and the patient to ground coupling capacitances are, respectively, C_p and C_b; the power line to electrode coupling capacitances are C_{e1}, C_{e2}, and C_{e3}; the power line to electrode lead coupling capacitances are C_{c1}, C_{c2}, and C_{c3}; Z_d, Z_{c1}, and Z_{c2} are the differential and common-mode input impedances for the measurement system; Z_{iso} is the isolation impedance; V_i and Z_i are the equivalent amplitude and impedance for interference coming from the internal power supply because of the imbalanced secondary (Fig. 2). Other displacement currents coupled to the isolated common, such as those described in [5], add to currents attributable to V_i. The power supply transformer is assumed to be shielded. Inductive interference is negligible because of the low magnetic fields arising from common power lines in buildings.

The main external contributions to interference come from displacement currents coupled into the patient body and currents coupled to the electrodes. Currents coupled to shielded wires are negligible. The resulting voltages appear in differential, common, or isolated mode. Hence, the equivalent input interfering voltage is

$$V_n = V_d + \frac{V_{\text{cm}}}{\text{CMRR}} + \frac{V_{\text{iso}}}{\text{IMRR}} \tag{1}$$

where V_d is the differential-mode interfering voltage, V_{cm} is the common-mode interfering voltage, and V_{iso} is the isolated-mode interfering voltage. Equation (1) shows that differential-mode interference directly adds to the signal of interest.
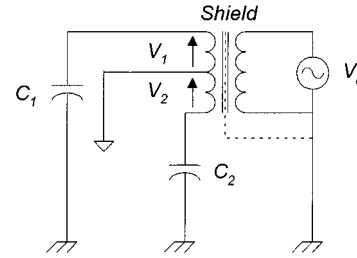

Fig. 2. Winding imbalance and stray capacitance to ground imbalance in the secondary yield internal interference in power supply transformers.

From Fig. 3(a), the respective voltages because of currents directly coupled to the patient are

$$V_{pd} \cong V_e \left[\frac{Z_{t1}}{Z_p} + \frac{Z_e}{Z_p} \frac{Z_b}{Z_{\text{iso}} + Z_b} \frac{Z_e}{Z_c} \left(\frac{\Delta Z_e}{Z_e} - \frac{\Delta Z_c}{Z_c} \right) \right] \tag{2}$$

$$V_{p\,\text{cm}} \cong V_e \frac{Z_e}{Z_p} \frac{Z_b}{Z_{\text{iso}} + Z_b} \tag{3}$$

$$V_{p\,\text{iso}} \cong V_e \frac{Z_{\text{iso}}}{Z_p} \frac{Z_b}{Z_{\text{iso}} + Z_b} \tag{4}$$

where V_{pd}, $V_{p\,\text{cm}}$, and $V_{p\,\text{iso}}$ are, respectively, the differential-mode, common-mode, and isolated-mode interfering voltages due to power line to patient capacitive coupling; Z_e is the average electrode impedance; $\Delta Z_e = Z_{e1} - Z_{e2}$; Z_c is the mean value for common-mode input impedances; $\Delta Z_c = Z_{c1} - Z_{c2}$; and Z_p and Z_b are the patient-power line and patient-ground impedances.

Interference decreases for a high patient-powerline impedance (Z_p), which depends on the closeness to power conductors. Interference decreases for a low patient-ground impedance (Z_b). Z_b decreases when the patient is close to grounded objects (e.g., a metal frame bed), but otherwise it should be high enough to ensure patient safety.
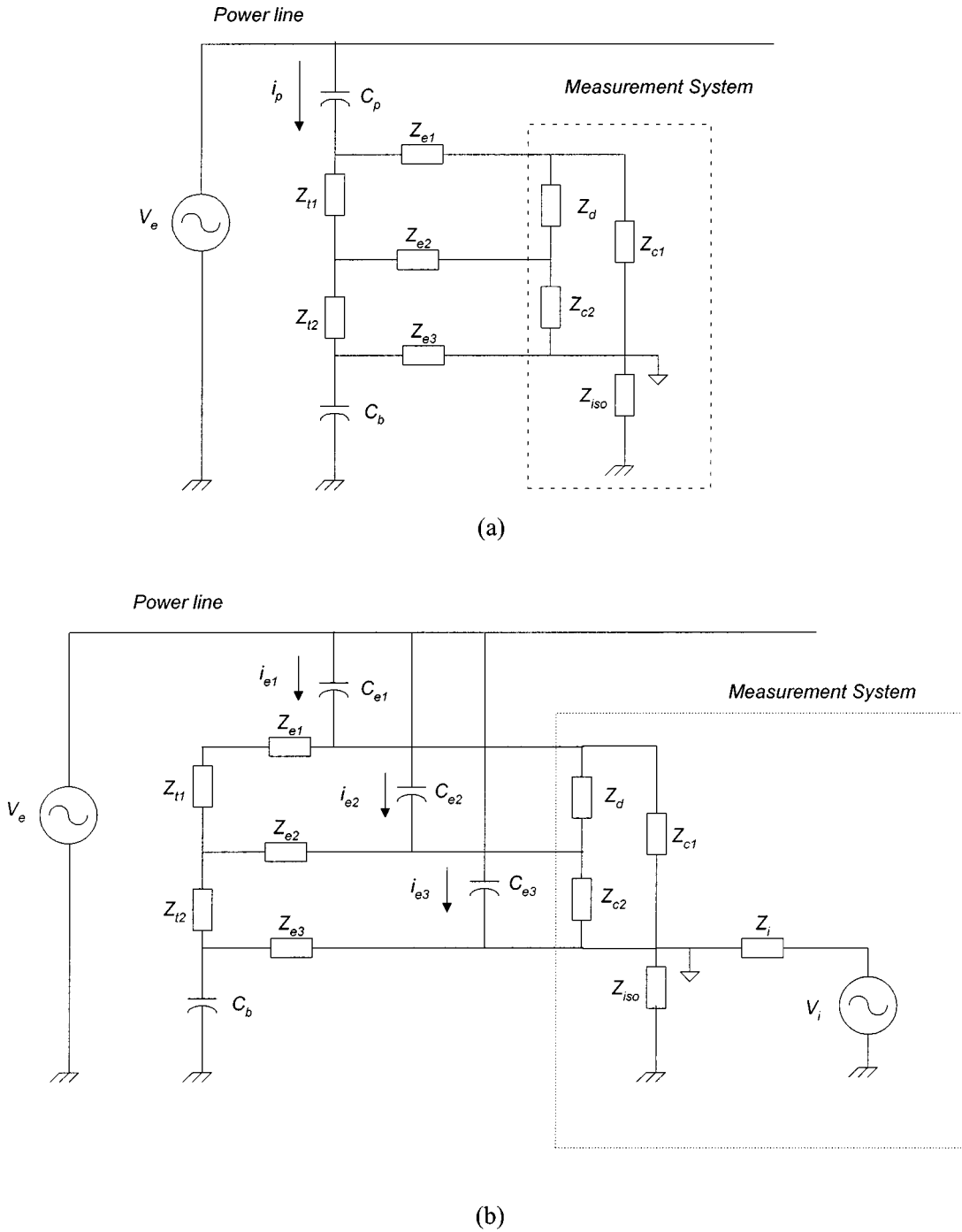
(a)

(b)

Fig. 3.   Equivalent circuit to analyze (a) capacitive coupling to the patient and (b) to the electrodes.

From Fig. 3(b), voltages from the displacement currents coupled to electrodes yield

$$V_{ed} \cong V_e \left\{ \frac{Z_{t1}}{Z_{ce}} + \frac{Z_e^2 Z_{iso}}{Z_c Z_{ce}(Z_{iso} + Z_b)} \right.$$
$$\times \left[\frac{Z_c(Z_{iso} + Z_b)}{Z_e Z_{iso}} \left(\frac{\Delta Z_e}{Z_e} - \frac{\Delta Z_{ce}}{Z_{ce}} \right) \right.$$
$$\left. \left. + \left(\frac{\Delta Z_{ce}}{Z_{ce}} - \frac{\Delta Z_c}{Z_c} \right) + \left(\frac{\Delta Z_e}{Z_e} - \frac{\Delta Z_c}{Z_c} \right) \right] \right\}$$

(5)

$$V_{e\,\mathrm{cm}} \cong V_e \frac{Z_e}{Z_{ce}} \frac{Z_{iso} + 2Z_b}{Z_{ce}(Z_{iso} + Z_b)} \tag{6}$$

$$V_{e\,\mathrm{iso}} \cong V_e \frac{3Z_{iso}Z_b}{Z_{ce}(Z_{iso} + Z_b)} \tag{7}$$

where $V_{ed}, V_{e\,\mathrm{cm}}$, and $V_{e\,\mathrm{iso}}$ are, respectively, the differential-mode, common-mode, and isolated-mode interfering voltages due to capacitive coupling to electrodes; Z_{ce} is the average power line-electrode coupling impedance, and $\Delta Z_{ce} = Z_{ce1} - Z_{ce2}$.
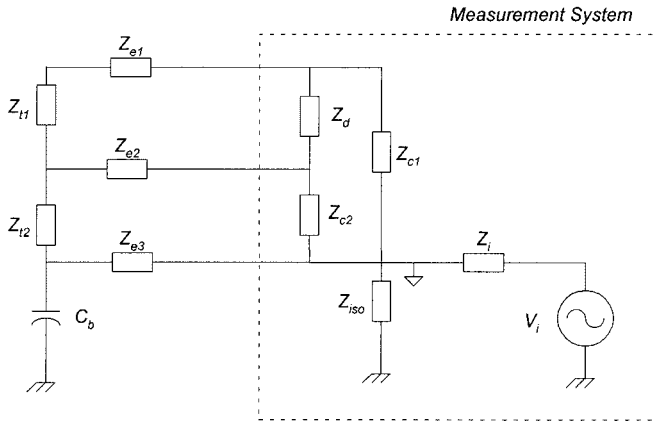
Fig. 4.   Equivalent circuit to analyze internal interference coupling.

The differential-mode component in (2) and (5) depends on electrode and common-mode input impedance balance. The common-mode component in (3) and (6) decreases for small electrode impedance. The isolated-mode component decreases for reduced isolation impedance. Nevertheless, safety regulations impose a minimal 22 MΩ isolation impedance. A large Z_{ce} (small coupling capacitance from power lines to electrodes) reduces interference.

The equivalent voltage source resulting from winding and coupling imbalance in the secondary of the power supply transformer (Fig. 2) has amplitude

$$V_i = \frac{\Delta V_s}{2} + \frac{V_s \Delta C_s}{2 C_s} = \frac{V_s}{2} \left(\frac{\Delta V_s}{V_s} + \frac{\Delta C_s}{C_s} \right) \qquad (8)$$

and impedance

$$Z_i = \frac{1}{j 2 \pi f 2 C_s} \qquad (9)$$

where V_s is the average secondary winding voltage, $\Delta V_s = V_1 - V_2$, C_s is the average stray capacitance to ground, and $\Delta C_s = C_1 - C_2$.

Therefore, internal interference is negligible only for power supply transformers with balanced secondary windings. From Fig. 4, the resulting interference voltages at the amplifier input because of V_i are

$$V_{id} = V_i \frac{Z_e}{Z_c Z_i} \frac{Z_{iso} \| Z_i}{Z_b + Z_{iso} \| Z_i}$$
$$\times \left[Z_{t1} + Z_e \left(\frac{\Delta Z_e}{Z_e} - \frac{\Delta Z_c}{Z_c} \right) \right] \qquad (10)$$

$$V_{i\,cm} = V_i \frac{Z_e}{Z_i} \frac{Z_{iso} \| Z_i}{Z_b + Z_{iso} \| Z_i} \qquad (11)$$

$$V_{i\,iso} = V_i \frac{Z_b}{Z_i} \frac{Z_{iso} \| Z_i}{Z_b + Z_{iso} \| Z_i} \qquad (12)$$

where V_{id}, $V_{i\,cm}$, and $V_{i\,iso}$ are, respectively, the differential-mode, common-mode, and isolated-mode interfering voltages due to internal interference.

Depending on the ratios between the isolation impedance Z_{iso}, the patient-to-ground impedance Z_b, and the amplifier common-mode impedance Z_c, a fraction of the ground-seeking current from the power supply (V_i/Z_i) flows to ground through

TABLE I
PARAMETER VALUES FOR CAPACITIVE
COUPLING AND INTERNAL INTERFERENCE IN FIG. 1

C_p (pF)	C_b (pF)	C_e (pF)	Z_t (Ω)	Z_e (kΩ)	ΔZ_e (kΩ)
1	200	0.04	20	10	2

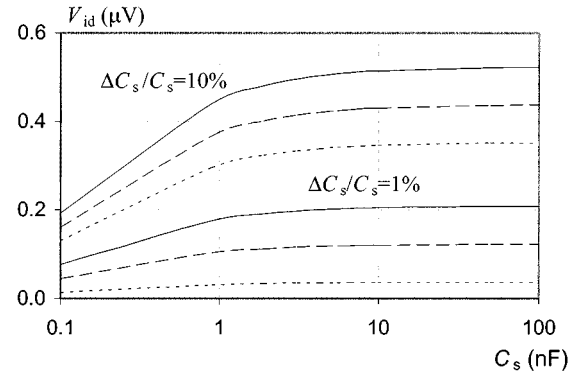|	Z_{iso}(MΩ)		Z_d (MΩ)	Z_c (MΩ)	ΔZ_c (MΩ)
Measurement System	22		10	10	1
	C_s (pF)		ΔC_s (pF)	V_s (V)	ΔV_s (V)
	500		100	21	1

Fig. 5.   Differential mode voltage (V_{id}) due to internal interference for different values of imbalance of stray capacitance to ground and different values of secondary voltage imbalance when $V_s = 10$ V ($\Delta V_s/V_s = 100\%$ solid line, $\Delta V_s/V_s = 50\%$ dashed line and $\Delta V_s/V_s = 5\%$ dotted line).

the patient and yields a differential and a common-mode voltage.

The overall interference because of the three factors considered (direct coupling to the patient, coupling to electrodes and internal interference) can be estimated by

$$V_n^2 = V_{np}^2 + V_{ne}^2 + V_{ni}^2 \qquad (13)$$

where subscripts p, e, and i stand, respectively, for patient, electrode, and internal.

III. RESULTS AND DISCUSSION

Parameter values in Fig. 1 heavily depend on the measurement setup. Table I lists some typical values reported by different authors. Capacitance coupling to the electrodes was estimated from measurements involving shielded cables. We have also assumed a maximal 5% imbalance in the secondary voltages of the power supply transformer and a 10% imbalance in their stray capacitance to ground. The assumed voltage imbalance agrees with our measurements in several medical-grade commercial linear power supplies. Some dc/dc converters have a similar imbalance. However, their specifications do not include that parameter. Furthermore, from (8) the internal interference voltage is directly proportional to the secondary voltage (V_s). Hence, medium- and high-power supply voltages increase interference. Fig. 5 shows the differential-mode voltage due to internal interference (V_{id}) versus average stray capacitance to ground (C_s) for different imbalance in secondary voltage (ΔV_s) and stray capacitance to ground (ΔC_s) when $V_s = 10$ V.
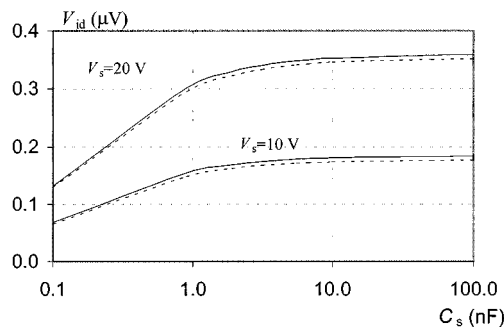
Fig. 6. Differential mode voltage (V_{id}) due to internal interference for different values of secondary voltage (V_s) and different values of secondary voltage imbalance $(\Delta V_s = 10\%$ solid line and $\Delta V_s = 5\%$ dotted line).

TABLE II
DIFFERENTIAL MODE (V_d), COMMON MODE (V_{cm}), AND ISOLATION MODE (V_{iso}) VOLTAGES FOR CAPACITIVE COUPLING AND INTERNAL INTERFERENCE, CALCULATED FROM VALUES IN TABLE I. FINAL INTERFERENCE (V_n), CALCULATED FROM (1) USING CMRR $= 10^4$ AND IMRR $= 10^6$

	V_d(average)	V_{cm}	V_{iso}	V_n(average)
Patient	2.1 µV	426 µV	900 mV	3 µV
Electrodes	8 µV	45 µV	100 mV	9 µV
Internal	2.4 µV	2.5 mV	3.6 V	6.2 µV

Fig. 6 shows the differential mode voltage due to internal interference (V_{id}) versus the average stray capacitance to ground (C_s) for different mean (V_s) and incremental values (ΔV_s) secondary voltages when $\Delta C_s/C_s = 20\%$.

The input differential voltage due to internal interference increases when the equivalent impedance Z_i decreases (i.e., C_s increases) until this impedance equals the isolation impedance. Afterwards the differential voltage is approximately constant. A grounded shield between the primary and the secondary windings does not solve the problem. That shield is only effective in reducing interference coupling between the primary and the secondary. Connecting capacitors from the secondary to ground would increase the interference.

Table II shows interference contributed by different coupling channels. Differential-mode interference is the worst because, according to (1), it adds to the desired signal. The CMRR and IMRR, respectively, attenuate common-mode and isolated-mode interference. Differential mode interference depends on electrode and amplifier common-mode input impedance balance, but a balanced system still yields some differential-mode interference because of Z_{t1}. This impedance depends on the patient orientation with respect to power conductors, which is not usually controllable.

A relatively simple method to reduce patient interference is by covering his/her body with a metal foil connected to the amplifier common, e.g., by a foil blanket, which works as electric shield. Shielding electrodes increases Z_{ce}, hence reducing interference. However, the amplifier input impedance decreases because of the increased capacitance from electrodes to the amplifier common, connected to the electrodes' shields. Shielded active electrodes, which connect a battery-supplied amplifier directly to the electrode using a short wire, keep high-input impedance yet reducing displacement currents into electrodes. Fig. 7 shows two ECG recordings, the upper trace obtained
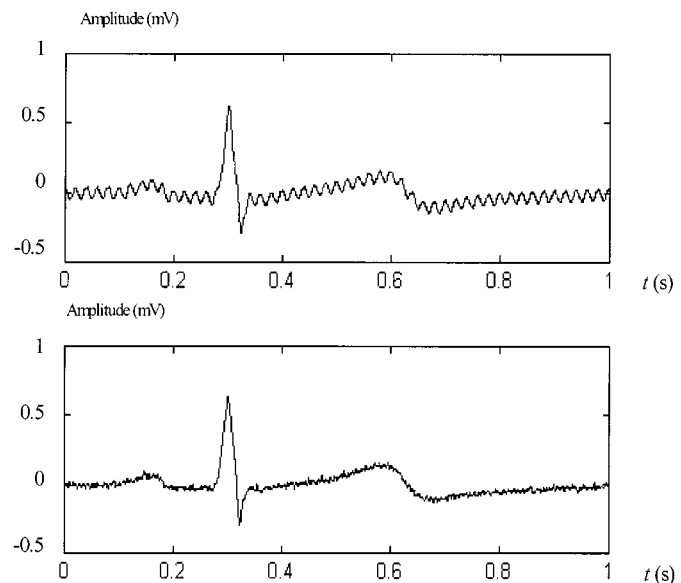

Fig. 7. Simultaneous ECG recordings, with standard electrodes (upper trace) and with active electrodes (lower trace).
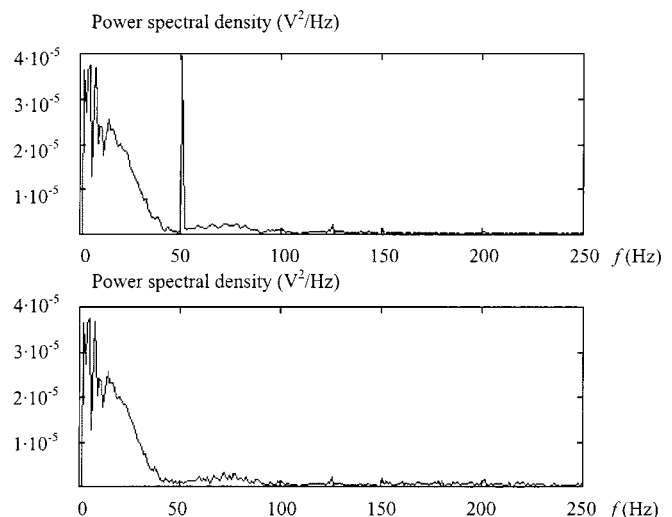

Fig. 8. Power spectral densities for the ECG signals in Fig. 7 show that active electrodes greatly reduce power line interference.

by standard electrodes and the lower trace obtained by active electrodes. Their respective spectra, Fig. 8, show that electrode shielding is quite effective in reducing power line interference (50 Hz). In this particular instance there was no need to shield the patient.

Internal interference highly depends on power supply cabling. Table II shows that the isolation-mode voltage because of internal interference can be high enough to contribute to the equivalent input interference v_n in spite of a high IMRR. This agrees with the common experience that battery-supplied amplifiers yield cleaner biopotential recordings than those from amplifiers supplied from the power line.

IV. CONCLUSION

Current models describing power line interference in biopotential measurements cannot explain interference present in

measurement systems that use shielded electrode leads and amplifiers with high CMRR and IMRR. Displacement currents coupled into the patient body can certainly explain some interference because they can produce a differential-mode voltage. However, shielding the patient by a grounded metal foil does not completely eliminate power line interference. Because a patient shield also shields electrodes, there must be some additional interference-coupling channel responsible for the remaining interference. Fig. 1 considers that additional channel to be the imbalance in power supply transformers, modeled in Fig. 2.

In common biopotential recordings (unshielded patient), unshielded electrodes account for most of the power line interference, which increases for large electrode impedance imbalance. Shielding electrodes reduce that interference, and active electrodes keep input impedance high.

The model in Fig. 1 applies to other measurements in volume conductors using electrodes such as impedance measurements in soil or in large samples. Internal interference can affect any measuring instrument with either differential or single-ended input.

In addition to a high input impedance and a large CMRR and IMRR, high-quality instruments using shielded power-supply transformers with central tap need the secondary to have balanced voltage and stray capacitance to ground.

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