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METHODOLOGY

The concurrent recording of electroencephalography and impedance cardiography: Effects on EEG

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Abstract

Three experiments were performed testing the effects of a variety of impedance cardiograph electrode types and recording arrangements on recorded electroencephalography (EEG) using either a monopolar single-ear reference or a physically linked ears reference. EEG was recorded either alone or concurrently with an impedance cardiograph. When the cardiograph was recorded using a spot electrode for the top current-inducing electrode, there was an overall decrease in power density of the EEG, and this effect was dependent on the location of the recording electrode. This effect was diminished when the top cardiograph spot electrode was replaced by a mylar-coated neck band electrode and EEG was recorded using a monopolar, single-ear reference. However, there tended to be an overall increase in log power density of the EEG in each frequency band below 60 Hz when less technologically advanced EEG amplifiers were used. This effect was diminished if the EEG was recorded using a physically linked ears reference. Recommendations for the concurrent recording of EEG and impedance cardiography are discussed.

Descriptors: Electroencephalography, Impedance cardiography, Human

The purpose of these experiments was to develop an adequate method for the simultaneous recording of electroencephalography (EEG) and impedance cardiography (IC). Both EEG and IC are commonly used in psychophysiological research. Valid and reliable methods for the recording of each have been well documented (for EEG, see Nunez, 1995; Pivik et al., 1993; for IC, see Sherwood et al., 1990). However, no study has been reported in which simultaneous measures of EEG and IC were recorded in a group of humans. Because the recording of EEG is extremely sensitive to external electrical noise and because IC requires the imposition of an external current, concurrent recording of EEG during IC measurement is fraught with complications.

EEG involves the recording of electrical field potentials that occur at the surface of the scalp and are related to underlying cortical neural activity. The EEG signal recorded is typically small in magnitude, ranging from around 0.1 to 100 μV, with frequencies ranging from DC to around 50 c/s. The signal recorded from a given site on the scalp is not the absolute activity at that site but rather the potential difference at that site with respect to another site (bipolar and common reference) or an averaged (average reference) or weighted average (source derivation) of all or a few other sites. Signals that are recorded at a given site are referred to as differential or out-of-phase signals and are the signals of interest in the EEG. The primary source of such signals is assumed to be biological and located within the brain. Signals that are common to both the "active" recording site and the reference are referred to as common mode or in-phase signals and can be a source of interference in the EEG. These signals can arise from other physiological sources (e.g., myocardial activity) or from external physical sources, such as neighboring electrical noise. A balanced or differential amplifier is able to discriminate between these two types of signals and amplifies the out-of-phase signal relative to the in-phase signal. The effectiveness of this discrimination is assessed by computing the ratio of differential signal to the common mode signal and is referred to as the common mode rejection ratio (CMRR). The higher the CMRR, the more effective the amplifier will be at rejecting external interference, all other things being equal.

A critical feature of the differential amplifier is that the amplifier will only be effective at rejecting external interference if the interference appears as a common-mode signal. In general, the closer an external noise source is to a given site relative to the reference site, the less the signal from that external source will appear as a common mode signal. In typical EEG recording situations, the reference site is close to the scalp (e.g., the ear lobe or tip of the nose) or on the scalp itself, as is the case for average or weighted average references, and the individual is usually isolated from external electrical noise sources either by shielding or by maintaining a considerable distance between the individual and the noise source. However, these precautions are not possible when a number of other electrical devices are used to monitor other aspects of the individual's physiology, as is the case during some surgical procedures or when IC is used.

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IC involves inducing an electrical field across an individual's chest and measuring changes in transthoracic electrical impedance. These changes in transthoracic impedance are inversely related to thoracic fluid volume and can then be used to derive an estimate of stroke volume (Kubicek, Karnegis, Patterson, Witsoe, & Mattson, 1966). Typically, a current is induced between two outer electrodes while two inner electrodes measure voltage changes across the thorax. The induced signal is typically low in current, in the milliampere or microampere range, and high in frequency, 20–200 kHz. Traditionally, the electrodes used for recording IC were mylar-coated bands that were place around the throat and chest (Sherwood et al., 1990). More recently, disposable spot or patch electrodes have been used in place of the bands, and a variety electrode placements have been used effectively in psychophysiological research (e.g., Sherwood, Royal, Hutchenson, & Turner, 1992). The spot electrodes have a number of advantages over the band electrodes. In particular, they are considerably easier to apply and are less invasive than the bands. However, the current distribution induced using spot electrodes may be less uniform than that induced using band electrodes.

The signal generated between the two current-inducing electrodes of the IC is an external electrical signal that represents a potential noise source for EEG recording. If EEG is recorded during IC recording, the amplitude of the externally imposed current used for IC should be diminished to the extent that it is common to both inputs of the amplifiers used to record EEG. This criterion is met for an electrocardiogram recorded with the two inner electrodes of the IC because these electrodes are both within the electrical field produced by the two, current inducing electrodes. However, it is not met for the EEG. The effect of the IC signal on the EEG recorded at a given site on the head should be dependent on the relative distances among the IC current-inducing electrode, the EEG reference electrode, and the EEG scalp electrode. Specifically, the less common the IC signal is to both the EEG reference and the recording electrodes, the more the IC signal will appear as a differential signal, assuming a constant CMRR. Furthermore, if the differential signal is far outside the bandwidth of the EEG amplifier, the amplifier will no longer function as a true operational amplifier and the CMRR will approach 1, greatly diminishing the amplitude of the EEG.

Other aspects of the EEG operational amplifier, such as the quality of the input protection diodes, may also be important in its use in the presence of a high frequency external noise source. An ideal operational amplifier draws no current at its input diodes. However, in practice some current does flows in each diode. The input bias current is defined as the average of the input currents at the two input diodes (e.g., Johnson & Jayakumar, 1982). These input bias currents are usually very small, typically in the nanoampere to picoampere range. However, when the source impedance is high and/or a large external current source is present, even small differences in the quality of the input protection diodes may produce appreciable differences in input bias current (leakage current). In general, the larger the leakage current, the greater the potential for interference with the EEG signal, even for leakage current whose frequency is higher than that found in an EEG signal, because the lower harmonics of the current may appear in the EEG frequency range. Furthermore, the IC signal itself may not be stable, producing a distorted waveform with a wide frequency range, the leakage currents from which could spuriously contribute to signals in the EEG frequency range.

The type of EEG reference used may be able to minimize the effects of the leakage currents on the EEG. In particular, a physically linked ears reference may provide a low resistance shunt by which the leakage currents can travel away from the EEG scalp sites. A physically linked ears (or mastoids) reference is different from other traditional references because it involves shorting together two reference sites, thereby providing a low resistance pathway. Therefore, in the case of appreciable leakage currents, the physically linked ears reference may be more appropriate than an averaged ears reference (one in which the potentials of the two ears are mathematically averaged to provide one reference) or any other single site or averaged reference.

Given the potential problems associated with simultaneously recording EEG and IC, one might question the benefit of undertaking such an endeavor. Simultaneous measurements allow investigation of relations between cortical brain activity and cardiovascular reactivity in an unobtrusive, cost-efficient manner with reasonable resolution; both measures provide useful information in studies of emotion. In three experiments, we investigated possible methods for the simultaneous recording of EEG and IC. These experiments were performed on independent sample groups testing the effects of a variety of IC electrode types and configurations on recorded EEG using Grass model 12A5 amplifiers.

The purpose of the first experiment was to test the effect of two different arrangements of IC electrodes on EEG recorded using a monopolar single-ear computer rederived to average ears reference and Grass model 12A5-C amplifiers. The second experiment was designed to replicate the first experiment using Grass model 12A5-B amplifiers. In the third experiment, we tested the effect of a neck band IC electrode configuration on EEG recorded with model B amplifiers and two different EEG references: a monopolar single-ear computer rederived to average ears reference and a physically linked ears reference.

Method

Participants

In total, 30 undergraduates participated for either course credit or money: 11 women and 3 men in the first experiment and 8 men in both the second and third experiments (all independent samples).[1] All but one of the participants was right handed. The average height of the participants was 177.1 cm (SD = 7.01 cm), and the average weight was 73.1 kg (SD = 12.88 kg). The age of the participants was 18–23 years.

Design

For the first and second experiments, EEG was recorded both without IC and concurrently with IC using two different IC electrode configurations. The two IC electrode configurations differed in the type and placement of the top current-generating electrode (I_1). In one condition (spot IC condition), a spot electrode was placed at the top of the neck behind and slightly below the right ear. In the other condition (neck-band IC condition), a mylar band electrode was placed around the top of the participant's neck. The two configurations were identical with respect to the remaining three IC spot electrodes. For the third experiment, a 2×2 repeated measures design was adopted. The first variable, IC condition, referred to whether EEG was recorded alone (control no-IC condition) or concurrently with IC using a neck-band electrode configuration (neck-band IC condition). The second variable, EEG reference, referred to the type of reference used for the EEG; a

[1]Only male participants were used in the second and third experiments for ease of application of the IC electrodes.

single ear reference (A1) was computer rederived to an average ears reference or a physically linked ears reference (linked ears). IC electrodes were in place for each of the conditions. The no-IC versus the IC conditions were manipulated in all three experiments simply by turning the IC-recording apparatus on and off. Participants were unaware of this manipulation.

Procedure

Two 60-s baseline samples of EEG were recorded from each participant under each of the conditions described above. The order of occurrence of the conditions was counterbalanced across participants. During each 60-s recording, the participants sat in a comfortable chair and were asked to close their eyes, relax, and minimize facial and body movement. Application of the electrodes took approximately 1 hr, and recording of the baselines took approximately 1 hr, for a total of 2 hr for each session.

Physiological Recording Apparatus and Techniques

EEG recording and data reduction. EEG was recorded from standard scalp leads of the 10/20 system plus some nonstandard locations (Electrode Position Nomenclature Committee, 1994). For the first experiment, the leads were PO3, T5, T3, C3, CP5, CP3, FC7, FC3, F7, F3, FP1, and the homologous sites on the right hemisphere, plus Pz, Cz, and Fz. For the second and third experiments, the leads were F3, F4, F7, F8, FC3, FC4, FC7, FC8, FZ, AF3, and AF4. Activity was also recorded from the right ear electrode (A2) in each experiment. All leads were referenced either to the left ear electrode (A1) and then computer rederived to an average ears reference or to physically linked ears. Grass model 12A5-C amplifiers were used in the first experiment, and Grass model 12A5-B amplifiers were used in the second and third experiments.

EEG was recorded with a modified electrode cap manufactured by Electro-Cap. Electrode impedances were kept below 5,000 Ω, and differences in impedance between the two ear electrodes were kept below 200 Ω. Horizontal and vertical electrooculogram (EOG) was recorded from the external cathi and the supra- to suborbit of the right eye.

EEG and EOG were amplified with a Grass model 12A5 Neurodata System with a bandpass of 1–300 Hz and with the 60-Hz notch filter in. All signals were passed through active anti-aliasing low-pass filters set at 210 Hz with a 36 dB/octave roll-off. The output of the filters was digitized on line at 500 Hz with an 80386 PC clone equipped with a 12-bit A/D board and signal acquisition software. Data were stored on magnetic tape cartridges.

All EEG data were edited for artifacts on a high-resolution graphic monitor, which displayed 16 simultaneous channels. Any period with gross movement artifact was edited out. EEG recordings from each of the scalp leads during all artifact-free periods were analyzed. Epochs of EEG 2 s in duration were then extracted through a Hamming window. A fast Hartley transform (FHT) was applied to each 2-s epoch of EEG, with epochs overlapping by 50%. The FHT output was then converted to power density (μV^2/Hz) in each of six bands: delta (1–4 Hz), theta (4–8 Hz), alpha (8–13 Hz), beta-1 (13–20 Hz), beta-2 (20–30 Hz), and gamma (36–58 Hz) by summing activity across all the 1-Hz bins within a band and dividing by the number of bins in that band. All power density values were then log transformed to normalize their distributions.

IC recording. Measures of cardiac performance were obtained using a computerized impedance cardiograph (CIC-1000), software version 5.14 (SORBA Medical Systems, Milwaukee, WI). The CIC-1000 generates a 0.5-mA, 50-kHz oscillating current between the two outer current-generating electrodes. Pregelled low-impedance ultrastable disposable ICE-1106 spot electrodes were used either alone (spot IC condition) or in combination with an adhesive aluminum-coated mylar band electrode (neck-band IC condition). The top current-generating electrode was placed as described above, and the three remaining spot electrodes were placed according to the manufacturer's recommendation (for details, see Miles & Gotshall, 1989).

Results[2]

Preliminary analyses showed no difference in the pattern of results among six EEG frequency bands (delta, theta, alpha, beta-1, beta-2, gamma); therefore, only results in the alpha frequency band (8–13 Hz) are reported for conciseness. A Cardiograph (control-no IC, spot, neck-band) × Region[3] × Hemisphere (right, left), repeated-measures analysis of variance (ANOVA) was first performed with log power density as the dependent measure for the first two studies.[4] Only the effects with the cardiograph variable are discussed because this was the primary variable of interest. All of the possible effects with cardiograph were significant in both experiments (see Table 1). A similar 3 (cardiograph) × 3 (region: Pz, Cz, and Fz) repeated-measures ANOVA was also performed for the midline sites. Again, all of the possible effects with cardiograph were significant for both studies. The three-way interaction in each experiment was further investigated with separate variance t tests comparing the control no-cardiograph condition with the spot and neck-band cardiograph conditions at each site on the head, $\alpha = .05$.

The spot condition had significantly less power than the control condition at all sites in the first experiment (see Table 2). This decrease in log alpha power density was greatest at sites closest to the top current-generating IC electrode and farthest from the EEG ear reference electrode (see Figure 1). This same pattern was observed in the second experiment; the spot condition was associated with significantly less log alpha power than the control condition at all sites except F7 (see Table 3). The neck-band and the control conditions did not differ at any site in the first experiment (see Table 2). The neck-band condition was associated with more log alpha power than was the control condition at all sites in the second experiment (see Table 3). However, this increase was significant only at F7.

A 2 (cardiograph: control-no IC, neck-band IC) × 2 (EEG reference: A1, linked ears) × 5 (region) × 2 (hemisphere) repeated-measures ANOVA was first performed with log alpha power density as the dependent measure for the third study. The effects that were of primary interest—the cardiograph main effect, the Cardiograph × EEG Reference interaction, and the four-way interaction—were all significant but small effects, accounting for 6%, 7%, and 0.4% of the variance, respectively (see Table 4). Preliminary analyses indicated no differences between the two references

[2]The IC data were not analyzed in any of the experiments because the purpose of these studies was to investigate any possible effects of recording IC on recorded EEG.

[3]For the first experiment there were 11 regions: parieto-occipital (PO3–4), anterior temporal (T3–4), posterior temporal (T5–6), central (C3–4), medial centroparietal (CP3–4), lateral centroparietal (CP5–6), medial frontocentral (FC3–4), lateral frontocentral (FC7–8), medial frontal (F3–4), lateral frontal (F7–8), and frontal (FP1–2) poles. For the second and third experiments, there were five regions: FC3–4, FC7–8, F3–4, F7–8, and anterior frontal (AF3–4).

[4]Huynh–Feldt correction factors were applied to all analyses.

Table 1. *Condition Effects for the Cardiograph × Region × Hemisphere ANOVA for the First and Second Experiments*

Effect	F	p	ϵ	η^2
Experiment 1				
Cardiograph (C)	1674.52	.0001	0.9247	.5865
C × Region (R)	24.64	.0001	0.3437	.0023
C × Hemisphere (H)	942.69	.0001	0.8166	.0321
C × R × H	173.50	.0001	0.3101	.0097
Experiment 2				
Cardiograph	138.02	.0001	0.8607	.5684
C × R	10.85	.0002	0.2207	.0054
C × H	52.46	.0001	0.5382	.0178
C × R × H	45.84	.0001	0.3530	.0108

Note: ϵ = Huynh–Feldt correction epsilon; η^2 = proportion of variance accounted for by a given effect.

for the control conditions, so the A1 and linked-ears reference control conditions were averaged to produce one control condition for further comparisons. The four-way interaction was then further investigated with separate variance t tests comparing the averaged control condition, the neck-band + A1 reference and the neck-band + physically linked ears conditions at each EEG site on the head, α =.05. The neck-band + A1 reference condition had sig-

Table 2. *Mean (SD) Alpha Log Power Density ($\mu V^2/Hz$) Under Three Cardiograph Conditions for the First Experiment*

	Cardiograph condition		
EEG site	Control-no IC	Spot IC	Band IC
PO3	2.33 (1.28)	1.12 (1.39)	2.28 (1.29)
PO4	2.37 (1.34)	−0.16 (1.37)	2.32 (1.38)
T5	1.73 (1.26)	0.67 (1.40)	1.66 (1.24)
T6	1.87 (1.40)	−1.02 (1.44)	1.83 (1.43)
P3	2.03 (1.23)	0.44 (1.30)	1.98 (1.24)
P4	2.06 (1.27)	−0.21 (1.31)	1.93 (1.44)
CP3	1.96 (1.25)	0.23 (1.57)	1.85 (1.18)
CP4	1.92 (1.15)	−0.24 (1.24)	1.49 (1.17)
CP5	1.53 (1.20)	−0.88 (1.26)	1.51 (1.28)
T3	0.83 (1.04)	−0.87 (1.26)	0.82 (1.01)
T4	0.98 (1.11)	−1.70 (1.38)	0.94 (1.11)
C3	1.58 (1.07)	−0.14 (1.35)	1.46 (1.17)
C4	1.64 (1.13)	−0.26 (1.22)	1.58 (1.16)
FC7	1.19 (1.03)	−0.19 (1.15)	1.17 (1.06)
FC8	1.22 (1.08)	−1.04 (1.01)	1.19 (1.11)
FC3	1.75 (1.04)	−0.19 (1.13)	1.67 (1.12)
FC4	1.74 (1.05)	−0.42 (1.06)	1.72 (1.07)
F7	0.85 (1.05)	−0.23 (1.07)	0.80 (1.06)
F8	0.86 (1.06)	−1.57 (0.99)	0.83 (1.06)
F3	1.45 (1.07)	−0.11 (1.09)	1.43 (1.09)
F4	1.49 (1.09)	−0.62 (1.08)	1.44 (1.10)
PZ	2.17 (1.23)	0.32 (1.46)	2.00 (1.37)
CZ	1.91 (1.02)	−0.19 (1.19)	1.78 (1.14)
FZ	1.62 (1.06)	−0.30 (1.06)	1.60 (1.09)

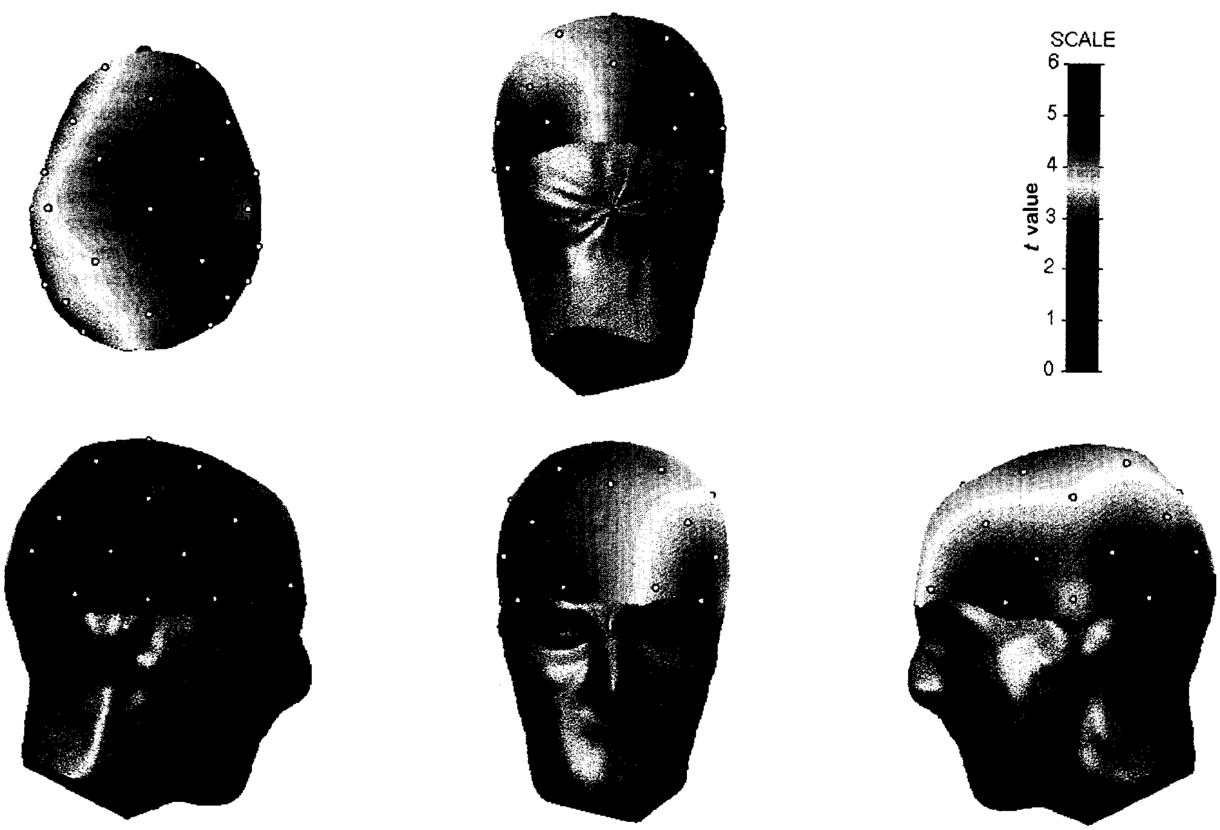

Figure 1. Topographic map of the control minus spot-IC alpha power density change score t values for Experiment 1. Maps were created by calculating the relevant t value for each site and then using a spline interpolation to map the resulting t scores onto a normative head model. Larger positive t values are reflected in the red end of the color scale and denote lesser alpha power density in the spot-IC condition compared with the control condition.

Table 3. *Mean (SD) Alpha Log Power Density ($\mu V^2/Hz$) Under Three Cardiograph Conditions for the Second Experiment*

	Cardiograph condition		
EEG site	Control-no IC	Spot IC	Band IC
FC7	1.33 (0.44)	0.83 (0.76)	1.77 (0.32)
FC8	1.27 (0.38)	0.05 (1.12)	1.49 (0.35)
FC3	1.74 (0.42)	0.70 (0.78)	1.90 (0.52)
FC4	1.71 (0.44)	0.64 (0.86)	1.96 (0.36)
F7	0.94 (0.33)	0.49 (0.78)	1.55 (0.29)
F8	0.94 (0.37)	−0.60 (1.24)	1.23 (0.39)
F3	1.46 (0.39)	0.47 (0.91)	1.85 (0.34)
F4	1.48 (0.40)	0.19 (0.84)	1.64 (0.47)
AF1	1.23 (0.39)	0.34 (0.75)	1.68 (0.32)
AF2	1.26 (0.38)	0.04 (0.91)	1.59 (0.36)
FZ	1.57 (0.40)	0.19 (1.00)	1.89 (0.36)

Table 5. *Mean (SD) Alpha Log Power Density ($\mu V^2/Hz$) Under Three Reference Conditions for the Third Experiment*

	Cardiograph × EEG Reference Condition		
EEG site	Control-no IC	A1 reference	Linked ears reference
FC7	1.19 (1.03)	1.77 (0.73)	1.13 (0.89)
FC8	1.22 (1.09)	1.28 (0.67)	1.17 (0.76)
FC3	1.75 (1.04)	1.88 (0.79)	1.59 (0.88)
FC4	1.74 (1.05)	1.76 (0.77)	1.60 (0.85)
F7	0.85 (1.05)	1.54 (0.66)	0.79 (0.78)
F8	0.86 (1.06)	1.01 (0.64)	0.86 (0.65)
F3	1.45 (1.07)	1.72 (0.74)	1.29 (0.83)
F4	1.49 (1.09)	1.54 (0.74)	1.36 (0.80)
AF1	0.96 (0.73)	1.57 (0.69)	1.09 (0.78)
AF2	1.23 (1.21)	1.41 (0.72)	1.16 (0.76)
FZ	1.37 (0.83)	1.68 (0.75)	1.41 (0.82)

nificantly more power than did either of the other two conditions in all comparisons (see Table 5). This effect was largest at sites closest to the EEG ear reference. This pattern is opposite that seen in the first and second experiments when I_1 was a spot electrode and similar to that seen when I_1 was a neck band electrode in the second experiment. The neck-band + physically linked ears condition did not differ significantly from the control condition in any of the comparisons.

Discussion

Concurrent recording of EEG and IC using four spot electrodes for the IC with a 50-kHz signal and Grass model 12A5-B and -C amplifiers for the EEG with a monopolar single-ear reference produced a significant decrease in the log power density of the recorded EEG. Although data for only the alpha band are presented here, the analyses indicated that this effect is present in all frequency bands (1–60 Hz). This decrease was greatest the closer the EEG site was to the top current-generating IC spot electrode. No effect on EEG was found when I_1 was a band electrode placed around the top of the neck with model C amplifiers (computer rederived linked ears reference) or with model B amplifiers if a physically linked ears reference was used. However, the pattern was reversed if I_1 was a mylar band electrode and EEG was recorded using model B amplifiers with a computer rederived linked ears reference.

Table 4. *Condition Effects for the Cardiograph × EEG Reference × Region × Hemisphere ANOVA for the Third Experiment*

Effect	F	p	ϵ	η^2
Cardiograph (C)	13.24	.0014		.06
C × EEG reference (E)	14.85	.0008		.07
C × Region (R)	27.25	.0001	0.615	.11
C × Hemisphere (H)	34.36	.0001		.007
C × E × R	14.96	.0001	0.590	.002
C × E × H	56.11	.0001		.01
C × R × H	26.60	.0001	0.586	.002
C × E × R × H	58.07	.0001	0.4584	.004

Note: ϵ = Huynh–Feldt correction epsilon; η^2 = proportion of variance accounted for by a given effect.

The Grass model 12A5-B and 12A5-C amplifiers are very similar in the input stage with the exception that the 12A5-C amplifiers contain slightly more advanced components. Both amplifiers use Burr–Brown ICI-2111 components for the input amplifiers, which have a CMRR of 80–90 dB, guaranteed for noise sources up to 20 kHz. The open loop gain of an ideal operational amplifier is infinite at all input frequencies. However, the gain for an actual operational amplifier is near infinity at zero frequency but drops off as the input frequency increases. This drop-off in gain as a positive function of input frequency is termed *roll-off.* For a differential amplifier, this decrease in gain translates into a decrease in the CMRR. A signal with a frequency far outside the band width of a given operational amplifier that is presented as differential input will decrease the gain of the amplifier and the CMRR will approach 1. Therefore, the decrease in amplitude of the EEG signal when a 50-kHz signal was introduced as a differential signal close to the head was most likely due to a substantial decrease in the CMRR of the amplifiers. However, the key finding was that this problem was diminished, or even eliminated, if the noise source was presented in such a fashion as to make it a common mode signal (a uniform band-generating electrode around the neck). In such a case, the high-frequency signal common at both inputs would be canceled out and the gain of the amplifier would not be affected.

In theory, any IC electrode configuration that produces a uniform signal across the head should also diminish the effect of the IC signal on the EEG. In fact, we found a similar pattern of results with two other IC configurations; a standard tetrapolar band configuration and a four spot electrode configuration with I_1 centered at the top of the neck (the other three electrodes positioned as described above).[5] However, the top band plus three spot electrode configuration has advantages over these two configurations. It is less invasive than the tetrapolar band configuration and is more consistent in producing a uniform signal across the head than the four spot electrode configuration. We established the validity of the IC signal using such a configuration by comparing it with a standard tetrapolar band configuration and a four-spot configuration (Dalton, Latta, & Davidson, 1996).

[5] The effect of the IC current on the recorded EEG under these conditions was identical to the effect under the combination condition and did not add to the interpretation of the present findings; therefore, these data were not included.

Other aspects of the EEG amplifiers also affected how the noise source was processed. The major difference between the model B and model C amplifiers is that the model C amplifiers have more sophisticated input protection diodes with extremely low leakage current (in the picoampere range). The input protection diodes in the B amplifiers are also low leakage current diodes but not quite as efficient (in the nanoampere range), so a balancing circuit is included to help balance any offset generated by leakage currents. This difference is small enough that it did not affect the way the amplifiers responded to signals in the typical EEG range when IC was not simultaneously recorded; however, this difference was appreciable when the 50-kHz IC signal was introduced as a uniform noise source close to the head. Spurious lower frequency harmonics from the 50-kHz signal may have leaked into the B amplifiers and positively biased the EEG amplitude. The amplitude of this effect increased as the distance between the EEG reference and head site decreased. These input currents themselves would be expected to be larger the smaller the input impedance of the source, which would be expected to be the case the closer the EEG site is to the reference. This effect was greatly diminished, if not eliminated, when a physically linked ears reference was used with the model B amplifiers probably because physically linking the ears provides a low resistance pathway by which the "leakage" currents may be shunted away from the scalp sites.

We expect these findings to be replicated using a wide frequency range of noise sources and amplifier characteristics. The most commonly used IC, the Minnesota 304B, utilizes a 4-mA 100-kHz signal. The BoMed NCCOM3-R7 (another frequently used IC) uses a 2.5-mA 70-kHz signal. In theory, a similar pattern of results would be expected using these units with the possible added difference that the effect may be larger with increases in noise source frequency. However, how well the noise is rejected from the EEG will be primarily dependent on characteristics of the EEG amplifiers.

A number of EEG amplifiers other than Grass are commercially available, such SAI Bioamps (James Long Co., NY), SynAmps (Neuroscan, VA), and Sensorium EPA-4 and -5 (Sensorium, VT). These amplifiers have CMRRs ranging from 100 to 126 dB guaranteed for up to 100 Hz and tend to have more advanced components than the Grass model 12A5 amplifiers. Again, in theory the same pattern of results would be expected using these amplifiers with the exception that the effects may be reduced with higher CMRRs and more sophisticated input diodes. In fact, higher CMRRs may offset any increases in the effect due to higher frequency noise sources and may actually eliminate the effect altogether. However, we recently tested a set of SAI BioAmps using the same procedure outlined above and found similar results using a 50-kHz noise source. Because the BioAmp's CMRR is state of the art (around 120 dB at 100 Hz), a CMRR of $\leq$120 dB probably is not sufficient to eliminate the effect of the 50-kHz noise signal on recorded EEG.

EEG and IC can be recorded simultaneously with little if any effect on the EEG if the IC signal is uniformly distributed across the head using a band electrode for I_1 and if any possible leakage currents are canceled out using a physically linked ears reference. This methodology should be successful for the concurrent recording of EEG and IC using other EEG amplifiers and other IC units. However, it is recommended that similar procedures be used to test for any possible effects of the IC signal on the EEG before such recordings are done. In particular, baseline recordings of EEG with and without an IC noise source should first be compared. If the amplitude of the EEG is significantly diminished, a deficiency in the CMRR of the EEG amplifier is indicated. In such cases, a mylar band electrode used around the top of the neck for I_1 should make the IC noise source more uniform across the head and should reduce its effect on the EEG. Alternatively, if the amplitude of the EEG is significantly increased, then low frequency components of the current from the noise source are most likely "leaking" into the EEG because of inefficient input protection diodes. In such a case, using a physically linked ears reference for the EEG should diminish the effect of the leakage current. If neither of these solutions is adequate, another alternative would be to introduce at the front of the EEG amplifiers filters specifically designed to work for the noise source frequency.

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