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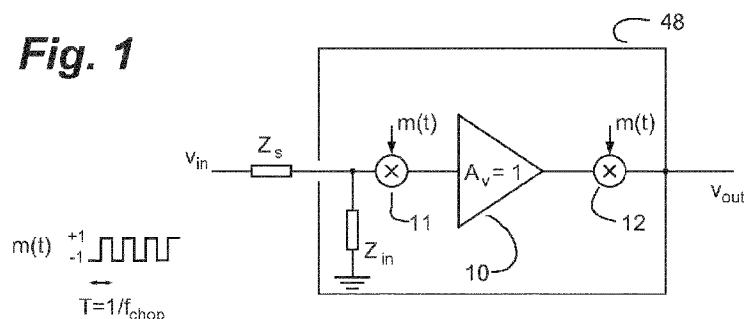
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(54) Title: AN ACTIVE ELECTRODE HAVING A CLOSED-LOOP UNIT-GAIN AMPLIFIER WITH CHOPPER MODULATION

Fig. 1

(57) Abstract: An active electrode has an electrode for sensing an electric potential and generating an input signal, and a shield placed near said electrode but being electric insulated from the electrode. An integrated amplifier (10) has an input connected to the at least one electrode for receiving the input signal, and providing a buffered path outputting a buffered output signal. The shield being connected to said output of the integrated amplifier to actively drive the electrical potential of said shield, thereby providing an active shielding of said electrode. The buffered path includes a first mixer (11) in front of the integrated amplifier for frequency shifting the input signal from a basic frequency range to a higher frequency range, and a second mixer (12) on the output of the integrated amplifier for frequency shifting the amplified signal from said higher frequency range back to said basic frequency range. The active electrode may be used for recording EEG signals.

AN ACTIVE ELECTRODE HAVING A CLOSED-LOOP UNIT-GAIN AMPLIFIER WITH CHOPPER MODULATION

The present invention relates to an electric field sensor. The invention, more particularly,
5 relates to electric field sensors for sensing an electric potential difference – e.g. a bio-
potential signal - and generating an input signal for signal processing. The sense
electrodes are guarded by an actively driven shield to minimize capacitive coupled noise
and minimize the capacitive load of the source signal. Also, the invention relates to a
sensor system for sensing an electric potential difference. Furthermore, the invention
10 relates to a method for sensing an electric potential difference in a sensor system.

Active electrodes are widely used for bio-potential recordings, especially for advanced
electrode technology like non-contact and dry-contact electrode. The purposes of
employing active electrode are shielding the electrode from external interferences, and
15 compensating the parasitic capacitances.

Compared with passive electrodes, active electrodes generally provide better immunity
against surrounding interferences, in particular capacitive coupled interferences as for
example Power Line Interference (PLI). Therefore, they are suitable for emerging
20 applications in wearable bio-potential recording. And nowadays, active electrode
combined with right-leg-driven technique is accepted as the main trend in the high-quality
bio-signal recording. In general, an active electrode should hold several features including
sufficient low input-referred noise, high input impedance and low bias current, low input
referred offset, low output impedance, high Common Mode Rejection Ratio (CMRR) and
25 Power Supply Rejection Ratio (PSRR), and for wearable devices a low power
consumption.

CMRR of a differential amplifier is the rejection of unwanted input signals common to
both input terminals, relative to the wanted difference signal. PSRR is the ratio between
30 the change in supply voltage in the op-amp and the equivalent (differential) output
voltage it produces. The output voltage will depend on the feedback circuit, and an ideal
instrumentation amplifier would have infinite PSRR.

Electrocardiography (ECG) is a transthoracic recording of the electrical activity of the
35 heart as a function of time. ECG signals are picked up by electrodes attached to the

surface of the skin and recorded by a device external to the body. Electroencephalography (EEG) is the recording of electrical activity along the scalp, and EEG signals is a measure of the voltage fluctuations resulting from ionic current flows due to neuronal activity within the brain. Ear-EEG is attractive as the electrodes can be arranged in the ear-canal
5 or around the ear in a device similar to a hearing aid.

Electrophysiological signals are normally weak in amplitude compared with surrounding interferences. Among regular physiology signals, ECG is relatively stronger typical with peak amplitudes in the $100\mu\text{V} - 1\text{mV}$ range. EEG is weaker, in the range of $10\mu\text{V} -$
10 $100\mu\text{V}$. For Ear-EEG peak amplitudes are typically in the range of $1\mu\text{V} - 10\mu\text{V}$, which is approximately 20dB below on-scalp EEG. However, the coupling interferences from surrounding could easily be in millivolt-level or even volt-level. Most of these interferences normally appear in common mode along with bio-signals. In principle, it's possible to pick up signals of interest clearly if electrodes and bio-amplifier are
15 completely differential, but in any practical amplifier, the CMRR is not infinite. Therefore, the noise immunity is of significant importance for bio-signal recording.

In practice it is not possible to design an ideal instrumentation amplifier, and thus any amplifier will be designed to provide a good trade-off between the ideal parameters.

20

The purpose of the invention is to provide an electric field sensor with improved key performance metrics relative to the ideal instrumental amplifier. By providing such an electric field sensor, EEG and ear-EEG sensors can be developed and designed for daily use e.g. for detecting Hypoglycemia. This may assist e.g. diabetes patients to maintain a
25 normal daily life.

An active electrode according to the invention comprises an electrode for sensing an electric potential and generating an input signal, a shield placed near said electrode and being electric insulated from said electrode, and an integrated amplifier having an input
30 connected to said at least one electrode for receiving the input signal, and providing a buffered path outputting a buffered output signal. The shield being connected to said output of the integrated amplifier to actively drive the electrical potential of said shield, thereby providing an active shielding of said electrode. The buffered path includes a first mixer in front of the integrated amplifier for frequency shifting the input signal from a

basic frequency range to a higher frequency range, and a second mixer on the output of the integrated amplifier for frequency shifting the amplified signal from said higher frequency range back to said basic frequency range.

- 5 The new technique proposed according to the invention provides a combination of the advantages from active electrodes designs with buffer and with chopper modulation amplifier, reaching a trade-off in key performances metrics.

10 Due to the finite input impedance of the amplifier, imbalance in electronic impedances leads to that part of the common mode signal will appear in differential mode on the input of the instrumentation amplifier. This is in particular the case in electrodes with high impedance as is the case for e.g. dry-contact and capacitive electrodes. Active shielding increases the input impedance, thereby increases the CMRR on the input of the amplifier, and thereby significantly diminishes this interference.

15 The active electrode design according to the invention provides several attractive advantages. Due to the unit-gain configuration of the amplifier configured as a chopped buffer, good shielding properties can be achieved. As a consequence, ultra-high input impedance is obtainable. The new active electrode design according to the invention
20 provides improved Common Mode Rejection Ratio (CCMR) and improved Power Supply Rejection Ratio (PSRR) between two buffer channels, which is very important for the noise immunity against surrounding interferences.

25 Chopper modulation shielded by the buffered output leverages the voltage-domain and the current-domain accuracies, reaching a good trade-off compared to conventional techniques using buffer and chopper amplifier. Besides, a significant benefit resulting from chopper modulation is the improved CMRR and PSRR between two buffer channels, which could be quite useful to enhance the noise immunity against surrounding interferences.

30 A chopper spike filter (CSF) in front of a subsequent differential amplifier could filter out the accompanying chopper spikes and ripples at the expense of extra amount of power. An active electrode with chopped buffer according to the invention will be highly useful within high-quality bio-recording systems.

A buffer amplifier provides electrical impedance transformation from one circuit to another. If the voltage is transferred unchanged (the voltage gain A_v is 1), the amplifier is a unity gain buffer. A unity gain amplifier (a buffer) is often implemented as a voltage follower as the output voltage follows or tracks the input voltage. Although the voltage gain of such a buffer amplifier may be (approximately) unity, it usually provides considerable current gain and thus power gain. A closed-loop unit-gain amplifier with chopper modulation according to the invention provides electrical impedance transformation, and acts as a voltage follower with a considerable current gain and thus power gain.

According to a further aspect of the invention there is provided a sensor system for sensing an electric potential difference. The sensor system comprises at least one set of electrodes including a reference electrode providing an electric potential reference and a sensing electrode providing a measurement point for measuring an electrical potential relative to the reference electrode. The sensor system further comprises a differential amplifier receiving input from the sensing electrode and the reference electrode and generating an output signal representing the electrical potential difference between the sensing electrode and the reference electrode. At least one electrode of the at least one set of electrodes being an active electrode, comprising an electrode for sensing an electric potential and generating an input signal, a shield placed near the electrode, and the shield being electric insulated from the electrode and an integrated amplifier having an input connected to the at least one electrode for receiving the input signal, and providing a buffered path outputting a buffered output signal. The shield is connected to the output of the integrated amplifier to actively drive the electrical potential of the shield, thereby providing an active shielding of the electrode. The buffered path includes a first mixer in front of the integrated amplifier for frequency shifting the input signal from a basic frequency range to a higher frequency range, and a second mixer on the output of the integrated amplifier for frequency shifting the amplified signal from the higher frequency range back to the basic frequency range.

According to a further aspect of the invention, there is provided a method of for sensing an electric potential difference in a sensor system having at least one set of electrodes including an reference electrode providing an electric potential reference and a sensing

electrode providing a measurement point for measuring an electrical potential relative to the reference electrode. The method comprises shielding an electrode by placing a shield near but electric insulated from the electrode, generating an input signal by means of the electrode sensing an electric potential, amplifying the input signal received from the electrode in an integrated amplifier providing a buffered path outputting a buffered output signal, connecting the shield to the output of the integrated amplifier to actively drive the electrical potential of the shield, thereby providing an active shielding of the electrode, frequency shifting the input signal from a basic frequency range to a higher frequency range by means of a first mixer placed in front of the integrated amplifier, and frequency shifting the amplified signal from the higher frequency range back to the basic frequency range by means of a second mixer placed on the output of the integrated amplifier.

The invention will be described in further detail with reference to preferred aspects and the accompanying drawing, in which:

fig. 1 schematically illustrates an active electrode design according to an embodiment of the invention;

fig. 2 illustrates for a semiconductor based amplifier the noise spectrum composed by white and pink noise;

fig. 3 illustrates schematically a bio-potential monitoring system employing an active electrode design according to an embodiment of the invention;

fig. 4 illustrates schematically one embodiment of a chopped buffer based upon an unit-gain amplifier for use in an electric field sensor according to the invention;

fig. 5 illustrates a chopped buffer according to one embodiment of the invention;

fig. 6 illustrates the relative positions of the bio-signal and the noise in the frequency domain for the active electrode design shown in fig. 1;

fig. 7 illustrates one embodiment of a chopper switch used in the chopped buffer shown in fig. 5; and

fig. 8 illustrates one embodiment of a sensor system based upon two active sensors according to the invention;

- 5 fig. 9 illustrates one embodiment of an EEG device according to one aspect of the invention; and

fig. 10 shows the excess noise sources at the input of chopper amplifier.

10 DETAILED DESCRIPTION

Fig. 1 schematically illustrates an active electrode design according to an embodiment of the invention. A bio-potential signal $V_{in}(t)$ is sensed by a capacitive electrode (not shown) and fed to an input of an integrated amplifier 10. Impedance, Z_s , denotes the skin-electrode impedance. On the input of the integrated amplifier 10, the bio-potential signal
15 $V_{in}(t)$ is modulated with a modulation signal (chopper clock), $m(t)$ in a first mixer 11 in front of the integrated amplifier 10. The integrated amplifier 10 has a gain A_v equal to one, whereby the integrated amplifier 10 acts as a buffer, and by applying the same chopper modulation signal, $m(t)$, in a second mixer 12 on the output of the integrated amplifier 10, too, the integrated amplifier 10 and the two mixers 11 and 12 provide a
20 buffered path outputting an output signal $V_{out}(t)$.

The modulation signal $m(t)$ employed in the embodiment shown in fig. 1 is illustrated as a pulse-width modulated signal having a duty-cycle of 50%, and assumes a unity amplitude of +1 and -1. The chopper frequency, f_{chop} , is selected to ensure that flicker noise in the
25 low frequency range will be substantially eliminated. Impedance, Z_{in} , denotes the finite input impedance. The chopped buffer output $V_{out}(t)$ is used for driving the active shield placed near said electrode and being electric insulated from said electrode.

With reference to fig. 2, the noise spectrum composed by white and pink noise for a
30 semiconductor based amplifier is illustrated. The corner frequency f_c characterizes the border between the region dominated by the low-frequency flicker noise (pink noise) and thermal noise which is dominating as the higher frequency "flat-band" noise (white noise). Flicker noise occurs in most electronic devices, and provides a limitation on the

signal level a circuit may handle. This is illustrated in fig. 2 where $\log_{10}(f)$ is depicted on the x-axis, and the voltage squared is depicted on the y-axis.

In the current embodiment, the integrated amplifier is realized in a MOSFET transistor layout, and a corner frequency in the level of approximately 200 Hz has been observed. The corner frequency, f_{corner} , is the transition between the regions dominated by the low-frequency flicker noise and the higher frequency "flat-band" noise, respectively. Therefore the chopper frequency, f_{chop} , has to be chosen so well above the corner frequency, so the frequency shift introduced prior to the integrated amplifier is sufficient to escape the flicker noise region of the integrated amplifier. The modulation frequency providing the frequency shift is greater than the corner frequency, and according to the illustrated embodiment the chopper frequency f_{chop} has been chosen to be in the range from 200 Hz to 2 kHz. Preferably, chopper frequency f_{chop} is in the range from 400 Hz to 1 kHz. When the chopper frequency f_{chop} is higher, the power consumption will be adversely affected.

For an ear-EEG application the sense electrode will pick up a bio-potential signal $V_{\text{in}}(t)$ having amplitude at approximately $1\mu\text{V}$. The bio-potential signal $V_{\text{in}}(t)$ will in a first use situation have a spectral distribution in a basic frequency range between 0 and 35Hz – which is schematically illustrated in fig. 6a. Once modulated with the chopper signal $m(t)$ in the mixer 11, the bio-potential signal $V_{\text{in}}(t)$ will be shifted in frequency so it appears around the chopper frequency at e.g. 1 kHz as is illustrated in fig. 6b. The integrated amplifier will introduce flicker noise in the spectrum up to the corner frequency e.g. at 200 Hz, while the frequency range above the corner frequency – including the frequency shifted bio-potential signal - will only be affected by white, thermal noise. This is illustrated in fig. 6c.

In the mixer 12, the output from the integrated amplifier 10 is modulated with the chopper signal $m(t)$, where the bio-potential signal is brought back to the basic frequency range again, while the flicker noise of the amplifier is positioned around the chopper frequency. This is illustrated in fig. 6d. An appropriate low-pass filtering at a later signal processing stage will remove the flicker originated noise now present in the frequency range around the chopper frequency.

An active electrode design according to the invention may be designed with low input-referred noise, high input impedance and low bias current, low input referred offset, low output impedance, high CMRR and PSRR, and low-power consumption. The actual implementation of an active electrode may be optimized for different applications, for example implantable neural probe array and fabric-based use (dry-contact electrode).

Fig. 3 shows schematically a bio-potential monitoring system employing an active electrode design according to an embodiment of the invention. A plurality of electrodes is arranged in mesh 30 positioned on the scalp 35 of a user. In another embodiment, the electrodes may be provided on an earplug and data may be collected from the ear canal and processed in a battery driven data processor placed behind the ear. Electrodes 31 and 32 do each include a probe 34 being a capacitive sense electrode and an active shield electrode placed near but spaced apart from the capacitive sense electrode. The input signal picked up by the probes 34 is led to respective amplifiers 10, preferably arranged as a unity gain amplifier. The closed-loop unit-gain amplifiers 10 are connected between the sense electrode and the active shield electrode. With this arrangement, the parasitic capacitor of the sense electrode is effectively reduced, thereby increasing sensitivity.

The output from the closed-loop unit-gain amplifier 10 with chopper modulation is via a shielded cable 13, e.g. a coax cable, fed to a Variable-Gain Amplifier 14 varying the gain based on a control voltage, and further to an Analog-to-Digital Converter 15 converting the amplified V_{Bio} signal into a digital representation for further processing. The Variable-Gain Amplifier 14 is a differential amplifier. Shielding is preferred but not crucial/necessary between the front-end integrated circuit containing the closed-loop unit-gain amplifier 10, and the back-end integrated circuit containing the Variable-Gain Amplifier 14 and the Analog-to-Digital Converter 15.

In the following there is provided a technical description of the active electrode using a chopped buffer according to the invention. Fig 4 illustrates that there exist several major parasitic-capacitance contributors in an active electrode concept. A shield 44 is placed near (and substantially in parallel with) said electrode 43, and the shield 44 is electric insulated from said electrode 43. There is an electric insulator (not shown) between the electrode 43 and the shield 44. This arrangement will cause a capacitive coupling between the electrode 43 and the shield 44. The electrode 43 is connected to the integrated circuit

via input pads 43a, and a capacitive parasitic coupling there between may be observed. The shield 44 is via input pads 48a connected to a shield 48 enclosing the integrated circuit, and also here there will be a capacitive parasitic coupling. With shielding by a buffer, electrode parasitic capacitances 40 at the sensing electrode and parasitic
5 capacitances 41 caused by capacitive couplings between the input pads 48a and 43a can be compensated.

The active electrode concept shown in fig. 4 illustrates that the amplifier is implemented as MOSFET transistors on a substrate (the integrated circuit). The amplifier is connected
10 to a power supply 46 and ground 45 via respective contact pads 46a and 45a, and has an output terminal 47. The output terminal 47 of the integrated amplifier is connected to the shield 48, while the contact pads 45a and 46a are electrical isolated therefrom. The shield 48 is connected to the output terminal 47 of the integrated amplifier to actively drive the electrical potential of the shield 48, thereby providing an active shielding of the electrode
15 43. Some capacitances are difficult to compensate, because shielding to their bottom node cannot be applied. This count for the parasitic capacitance 42a between the input pad 49a and the substrate, the parasitic capacitance between the transistor gate and the substrate – the gate-to-substrate capacitance 42b, the parasitic capacitance referred between gate and source of the transistor – the gate-to-source capacitance 42c, and the parasitic capacitance
20 between gate and drain of the transistor – the gate-to-drain capacitance 42d. In these circumstances, the objective is to design the circuitry such that the value of the capacitors will be as small as possible.

As shown in fig. 5, a choppered buffer according to the invention is implemented based
25 on a closed-loop unit-gain amplifier 10 according to the illustrated embodiment. An input transistor pair, M_1 and M_2 , of the closed-loop unit-gain amplifier 10 is minimized in size in order to reduce the input parasitic capacitances and thereby get high impedance. Flicker noise of input transistors M_1 and M_2 is a non-dominant noise source due to the employed chopper modulation.

30

A constant current is maintained through the input transistor pair, M_1 and M_2 , through the use of the current source 50 formed by a transistor M_N and a voltage source V_{battery} . The voltage source V_{battery} may according to one embodiment be a coin-cell battery of the type used for hearing aids having nominal supply voltage being approximately 1.2 V. A bias,

V_{bp} , is applied to the gate of the MOSFET transistor M_N controlling the current from the voltage source $V_{battery}$ fed to the source of the input transistor pair, M_1 and M_2 .

By maintaining a constant current through the input transistor pair, M_1 and M_2 , and
 5 applying the negative feedback of the unity gain configuration, the gate-to-source capacitance 42c (fig. 4) and gate-to-substrate capacitance 42b (fig. 4) to the sensor input are minimized.

The illustrated embodiment for implementing the closed-loop unit-gain amplifier 10 with
 10 chopper modulation according to the invention employs three chopper switches $CHOP_1$, $CHOP_2$, and $CHOP_3$. The sizes of chopper switches $CHOP_1$, $CHOP_2$, and $CHOP_3$ are optimized for speed and noise, and in this topology, the chopper switches $CHOP_2$ and $CHOP_3$ are arranged inside of the closed-loop unit-gain amplifier 10. Hereby, by using the inherent differential nodes, no extra differential nodes will be required. Furthermore,
 15 this will not limit the bandwidth of the closed-loop unit-gain amplifier 10 with chopper modulation.

The input chopper switch $CHOP_1$ receives the sensed bio-potential signal V_{in} as a first input signal, and the output signal V_{out} from the closed-loop unit-gain amplifier 10 via a
 20 feedback branch 53 as a second input signal. The chopper switch $CHOP_1$ operates at 1 kHz chopper frequency. The chopper signal alternates between +1 and -1 at 50% duty cycle. The bio-potential signal V_{in} has a low bandwidth (normally between 0-40 Hz), but the chopper frequency shall be above the corner, f_{corner} (fig. 2). Choosing the chopper frequency to be too high, will adversely affect the power consumption of the overall
 25 electrode assembly.

The gates of the input transistor pair, M_1 and M_2 , receive respective outputs from the input chopper switch, $CHOP_1$. The constant current from the current source is passed through the input transistor pair, M_1 and M_2 via respective source terminals, and the
 30 drains of the input transistor pair, M_1 and M_2 , are connected to respective terminals on the second chopper switch, $CHOP_2$.

The two outputs from the chopper switch, $CHOP_2$, are connected to respective source terminals of a MOSFET transistor pair, M_3 and M_4 . The transistor pair, M_3 and M_4 , forms

a source follower (common-drain amplifier) being a Field Effect Transistor amplifier topology, typically used as a voltage buffer.

5 MOSFET transistors M_5 , M_6 , M_7 and M_8 form a cascoded current mirror circuit, which would be recognized by a person skilled in the art as being a standard component in an operational amplifier. The cascoded current mirror circuit is a two-stage amplifier composed of a transconductance amplifier followed by a current buffer. The third chopper switch $CHOP_3$ is arranged in between the two stages of the cascoded current mirror circuit. The cascoded current mirror circuit improves input-output isolation as there is no
10 direct coupling from the output to input.

Three MOSFET Transistors M_{NC} , M_9 and M_{10} are arranged as an extra source follower. The three MOSFET Transistors M_{NC} , M_9 and M_{10} are connected to the voltage source $V_{battery}$ and operated as a level shifter providing lower dc bias to the transistors M_3 and M_4
15 forming the source follower.

Fig. 7 illustrates one embodiment of the chopper switch $CHOP_1$, $CHOP_2$, $CHOP_3$ used in the chopped buffer topography shown in fig. 5. The chopper switch has a pair of input terminals 80 and a pair of output terminals 81. The chopper switch is shielded by bulks
20 by means of a shield 82, and includes four transistor switches 84, 85, 86, and 87 – all controlled by a clock signal, Clk. For the transistor switches 85 and 87, the clock signal, Clk is received via respective inverters 88 and 89, whereby the transistor switches 84 and 86 closes when the clock signal is high, and the transistor switches 85 and 87 closes when the clock signal is low. The inverters 88 and 89 are NOT gates implementing logical
25 negation. Thereby, the four transistor switches 84, 85, 86, and 87 ensures that a first terminal of the pair of input terminals 80 alternately connected to a first and a second terminal of the output of the pair of output terminals 81. The second terminal of the pair of input terminals 80 alternately connected to the second and the first terminal of the output of the pair of output terminals 81.

30

By connecting the bulks or the shield 82 to the buffer output 90, all the transistor switches 84, 85, 86, and 87 are shielded to eliminate the body effects and extra current flow through the bulk nodes. Therefore the bias current will not significantly change when input common-mode voltage varies. In addition, the chopping clock is bootstrapped to

keep the overdrive voltage and thereby the “ON” resistance of the transistor switches 84, 85, 86, and 87 substantially constant. In such a way, the current noise and the thermal noise are insensitive to input common-mode voltage. A bootstrap circuitry 83 clock signal and the output buffer signal via the shield 82 and delivers on the output the chopper clock
 5 signal, Clk, used to open and close the transistor switches 84, 85, 86, and 87. The bootstrap circuitry 83 deliberately intends to alter the input impedance.

As illustrated in fig. 10, the noise issues may be separated into two categories: voltage-domain noise and current-domain noise. Fig. 10 show the excess noise sources at the
 10 input of chopper amplifier as shown in fig. 5. A man skilled in the art will understand that the problems caused by current noise are highly depending on the value of source impedance. Since the skin-electrode impedance is relatively large, and subject to large variations, in the dry-contact acquisition analog front-end, both two noises deserve our concerns.

15

Voltage-domain noise

Fig. 10 shows the excess noise sources at the input of chopper amplifier. R_{sx} denotes the skin-electrode resistance. The main voltage noise contributors 60 include the input transistor pair, M_1 and M_2 , and the final transistor pair M_7 and M_8 in the cascoded
 20 current mirror circuit, due to the high voltage gain from the these MOSFETs gates. The voltage noise may be expressed as a result of the flicker noise and the thermal noise, and the offset and flicker noise is theoretically removed from low frequency signal band, due to the chopper modulation. The dominant component of the residue noise is thermal noise, V_{noise} , of MOSFETs and may be expressed as follows:

25

$$V_{noise} \approx \sqrt{(1 + g_{m7}/g_{m1})(8kT/g_{m1})BW} \quad (1)$$

where BW denotes the bandwidth of interest, g_{mi} the transconductance of MOSFETs M_i , k is Boltzmann's constant, and T is the absolute temperature of the component. Within the
 30 bandwidth between 0.5-100Hz, the integrated noise is approximately $0.29\mu V_{rms}$ when the chopper frequency is selected to be 1 kHz and the transconductance of the dominating MOSFETs has been optimized in order to minimizing the thermal noise.

From equation (1) it is seen that the noise voltage V_{noise} is proportional with the reciprocal value of the square root of the transistor transconductance. It has been found that current consumption of $8\mu\text{A}$ and noise of $29\text{nV}/\sqrt{\text{Hz}}$ as acceptable in terms of power and noise budget of the system, which has been indicated in fig. 5 by operating the closed-loop unit-gain amplifier 10 through the use of the current source 50 providing $8\mu\text{A}$.

Current-domain noise

Bias current 61 gives rise to offset across the source impedance. For bio-signal sensor amplifiers, the major sources of bias current include leakage in ESD protection circuitry, gate leakage of input MOSFETs and base current of bipolar junction transistors, chopping activities as well as PCB leakage. The dominant contributors include leakage of ESD protection circuitry and current flow caused by periodic chopping activities.

The leakage of ESD protection circuitry is highly dependent on the ESD techniques and circuitry properties. Therefore it exists in all the amplifiers and it's hard to be completely avoided. Periodic chopping activities give rise to dynamic current flow through the chopper switches and switch-capacitor resistance. By definition, the bias current is the average current over a relatively long time at the input node. For the CMOS chopper amplifiers, such kind of current could be the dominant bias current source over the others.

It has been observed that the excess noise normally can be regarded as negligible in amplifiers with low-source-impedance, for instance, $10\text{k}\Omega\sim 20\text{k}\Omega$ of wet electrodes. However for high-source-impedance, for instance, several hundred kilo ohm to several mega ohm of dry-contact electrode, the imperfections like dc offset and corresponding output noise will be regarded as problematic.

By applying appropriate design optimization strategies, all the switches may advantageously be shielded to eliminate the body effects and extra current flow through the bulk nodes. Therefore the bias current will not significantly change when input common-mode voltage varies. In addition, the chopping clock is bootstrapped to keep the overdrive voltage and thereby the 'on' resistance of the switches approximately a constant. In such a way, the current noise and the thermal noise are insensitive to variation in the input common-mode voltage.

By ensuring a well optimization of the choppered buffer according to one embodiment of the invention, the bias current is quite low. The chopper switches are naturally shielded by the buffer and there exist no significant potential differences between sources and drains as well as bulks in the switches. Therefore there is no current path in the chopper.

- 5 The current noise for the choppered buffer according to one embodiment of the invention has been observed in the level about $0.3\text{fA}/\sqrt{\text{Hz}}$. With a $1\text{M}\Omega$ resistor connected, the excess noise density contribution would be $0.3\text{nV}/\sqrt{\text{Hz}}$.

- The Common-Mode Rejection Ratio (CMRR) of a differential amplifier is the rejection
10 by the device of unwanted input signals common to both input leads, relative to the wanted difference signal. A high CMRR is required when a differential signal must be amplified in the presence of a disturbing common-mode input. Power Supply Rejection Ratio (PSRR) is defined as the ratio of the change in supply voltage in the op-amp to the equivalent (differential) output voltage it produces. The output voltage will depend on the
15 feedback circuit. Chopper modulation has been found to not only reduce the noise but also to contribute to the improvement of the CMRR and the PSRR.

- The Common-Mode Rejection (CMR) of the amplifier without chopper modulation has been observed to be -73.3dB and with chopper modulation the CMR is improved to $-$
20 107.9dB , almost 35dB enhancement in CMRR. Furthermore, the Power Supply Rejection (PSR) has been improved with chopper modulation from -48dB to -97.3dB , an almost 49dB enhancement in PSRR. This has been observed with a capacitance load of 10pF at the output node for a frequency band (wanted input signal) below 100Hz .

- 25 According to the invention there is provided a new choppered buffer employed in the active electrode design. Compared with conventional and state-of-art designs, an active electrode with choppered buffer exhibits several attractive advantages. Thanks to the unit-gain configuration, a well shielding property can be permitted. As a consequence, ultra-high input impedance is obtainable, and thereby high CMRR of input impedance network
30 could be realized. Chopper modulation shielded by the buffered output leverages the voltage-domain and the current-domain accuracies, reaching a good trade-off compared to conventional techniques using buffer and chopper amplifier. Besides, a significant benefit resulting from chopper modulation is the improved CMRR and PSRR between two buffer channels, which could be quite useful to enhance the noise immunity against surrounding

interferences. The subsequent differential amplifier could filter out the accompanying chopper spikes and ripples at expense of extra amount of power. The active electrode with choppered buffer is very suitable for use in high-quality bio-recording systems.

5 Fig. 8 shows an embodiment of a sensor system based upon two active electrodes according to the invention. The shown sensor system includes a front-end module 84 being connected to a back-end module 86 via a set of wires 80. The front-end module 84 includes in the illustrated embodiment a pair active electrodes 43 with choppered buffer as described with reference to fig. 1. The choppered buffer is based upon the integrated
10 amplifier 10 has a gain, A_v , equal to one, and two mixers 11 and 12 applying the same chopper modulation signal, $m(t)$. Furthermore the output terminal 47 of the choppered buffer is connected to the shield 48 enclosing the integrated circuit. Preferably, the active electrodes operate in differential mode – which means that one of the electrodes acts as reference.

15

The back-end module 86 has a choppered instrumentation amplifier based upon an integrated amplifier 82 having a gain for amplifying the bio-potential signal from the active electrodes, and two mixers 81 and 83 applying the same chopper modulation signal, $n(t)$. The chopper modulation signal, $n(t)$ is applied in order to avoid amplifying
20 the flicker noise in the integrated amplifier 82.

The chopper clock signal $n(t)$ is preferably a square-wave signal that contains odd harmonics at f_{ch} , $3f_{ch}$, $5f_{ch}$, and as most of the energy of the chopper ripple is located at the first harmonics, f_{ch} , the higher harmonics may be eliminated by applying a Chopper
25 Spike Filter (CSF) 85 providing a low pass or band pass filtering effect. The Chopper Spike Filter 85 includes a sample and hold circuit provided by a switch and a capacitor, where the switch is driven by sampling pulses. The Chopper Spike Filter 85 removes glitch caused by the chopper switches. Two branches fed from the output of choppered instrumentation amplifier 81-83, but with reverse polarity, has been included in order to
30 generate a fully differential output is fed a Programmable Gain Amplifier (PGA) 87 and an Analog-to-Digital Converter (ADC) 88, from where the signal is supplied to a not-shown microcontroller for processing.

The chopped instrumentation amplifier 82 may in one embodiment be provided in front-end module 84 and thereby included within the active shielding. Then the number of thin wires 80 connecting the front-end module 84 to a back-end module 86 may be reduced from four to two (shielded) wires. These wires carries supply voltage, ground, clock and signal and may in a specific embodiment have a length of 10 mm.

Fig. 9 shows an ear EEG device 115 according to one aspect of the invention. The ear EEG device 115 that can be worn inside the ear of a person to be monitored, e.g. for detecting Hypoglycemia, e.g. like a per se known In-The-Canal (ITC) hearing aid. Furthermore, the device will allow healthcare personal to remote monitor or record EEGs for several days at a time. Healthcare personal would then be allowed to monitor patients who have regularly recurring problems like seizures or micro-sleep. The ear EEG device 115 will not interfere with normal life, because the ear EEG device 115 has an acoustic vent 116 so the wearer will be able to hear. After a while, the wearer forgets that he wears the ear EEG device 115. The ear EEG device 115 is on its outer surface provided with two active electrodes 117 according to the invention. Internally the ear EEG device 115 contains an electronic module 118.

The ear EEG device 115 is formed to fit into the external auditory canal 111 of the wearer, and defines a cavity in the external auditory canal 111 together with the tympanic membrane 110, and the cavity is opened by means of the acoustic vent 116 extending through the entire length of the ear EEG device 115. Preferably the ear EEG device 115 does not extend beyond the pinna 112.

The electronic module 118 is shown schematically in enlarged view in the dotted box 118. The electronic module 118 includes a power supply 120 based upon a standard hearing aid battery for powering the electronics. The two electrodes 117 provided on the surface of the ear EEG device 115 pick up a potential and delivers the data via a module 125 operating as electrode frontend and Analog to Digital Converter (ADC) to a digital signal processor 124. Details of the electrode frontend and ADC module 125 has been explained with reference to fig. 8. The digital signal processor 124 receives the amplified and digitized signal for processing. According to one embodiment, the digital signal processor 124 analyses the EEG signal picked up for detecting hypoglycemia by

monitoring the brain wave frequency, and if the brain wave frequency falls beyond a predefined interval, this may indicate that a medical emergency may arise. Hypoglycemia is a medical emergency that involves an abnormally diminished content of glucose in the blood. Upon detection of abnormal brain wave activities, the digital signal processor 124
5 communicates these findings to a device operating controller 122.

The device operating controller 122 is responsible for several operations and has an audio front-end module 123 including a microphone and a speaker. With the microphone, the device operating controller 122 is able to pick up audio samples and classify the current
10 sound environment. Furthermore, the device operating controller 122 may have access to real time clock information – either from an internal clock module or from a personal communication device (e.g. a smartphone) accessible via a radio module 121. The personal communication device and the radio module 121 may establish a wireless communication link by means of a short range communication standard, such as the
15 Bluetooth™ Low Energy standard. The device operating controller 122 adjusts the predefined interval for normal the brain wave activity in dependence to the real time clock information and the sound environment classification. With the speaker, the device operating controller 122 is able to alert the wearer of the ear EEG device 115 that medical emergency may arise and that precautionary actions have to be taken.

20

The number of electrodes has so far been identified as a pair of active electrodes operating in differential mode. However two or more active electrodes may be acting as sensing electrodes for measuring the electric potential difference relative to an active electrode acting as a common reference electrode. The electrodes will operate in a
25 unipolar lead mode.

The ear EEG device 115 may in a further embodiment operate as a hearing aid if the processor is provided with a gain for alleviating a hearing loss of the wearer. The ear EEG device 115 may advantageously be integrated into an In-The-Canal (ITC) hearing aid, a
30 Receiver-In-Canal (RIC) hearing aid or another type of hearing aid.

CLAIMS

1. An active electrode, comprising:
 - 5 - an electrode for sensing an electric potential and generating an input signal;
 - a shield placed near said electrode, and said shield being electric insulated from said electrode;
 - an integrated amplifier having an input connected to said at least one electrode for receiving the input signal, and providing a buffered path outputting a buffered
 - 10 output signal;
 - said shield being connected to said output of the integrated amplifier to actively drive the electrical potential of said shield, thereby providing an active shielding of said electrode; and
 - said buffered path includes a first mixer in front of the integrated amplifier for
 - 15 frequency shifting the input signal from a basic frequency range to a higher frequency range, and a second mixer on the output of the integrated amplifier for frequency shifting the amplified signal from said higher frequency range back to said basic frequency range.
- 20 2. The active electrode according to claim 1, wherein said frequency shift is adapted to avoid the flicker noise region of the integrated amplifier.
3. The active electrode according to claim 1, wherein a corner frequency defines the transition between low-frequency flicker noise and higher frequency white noise, and
- 25 wherein the modulation frequency providing the frequency shift is greater than the corner frequency.
4. The active electrode according to claim 3, wherein the modulation frequency is lower than four times the corner frequency.
- 30 5. A sensor system for sensing an electric potential difference, comprising:
 - at least one set of electrodes including
 - a reference electrode providing an electric potential reference;

- a sensing electrode providing a measurement point for measuring an electrical potential relative to said reference electrode; and
 - a differential amplifier receiving input from said sensing electrode and said reference electrode and generating an output signal representing the electrical potential difference between said sensing electrode and said reference electrode;
- 5 - wherein at least one electrode of said least one set of electrodes being an active electrode, comprising:
 - an electrode for sensing an electric potential and generating an input signal;
 - 10 ○ a shield placed near said electrode, and said shield being electric insulated from said electrode;
 - an integrated amplifier having an input connected to said at least one electrode for receiving the input signal, and providing a buffered path outputting a buffered output signal;
 - 15 ○ said shield being connected to said output of the integrated amplifier to actively drive the electrical potential of said shield, thereby providing an active shielding of said electrode; and
 - said buffered path includes a first mixer in front of the integrated amplifier for frequency shifting the input signal from a basic frequency range to a higher frequency range, and a second mixer on the output of the integrated amplifier for frequency shifting the amplified signal from said higher frequency range back to said basic frequency range.
- 20 6. The sensor system according to claim 5, wherein said least one set of electrodes is a pair of electrodes both being active electrodes.
- 25 7. The sensor system according to claim 5, wherein said at least one set of electrodes is provided as a plurality of pairs of active electrodes.
- 30 8. The sensor system according to claim 5 in which a plurality of sensing electrodes are measuring the electric potential difference relative to a common reference electrode.
- 9. The sensor system according to claim 8, wherein said plurality of sensing electrodes being active electrodes.

10. The sensor system according to claim 5, wherein sensing electrodes are measuring the electric potential difference relative to a linear combination of the other electrodes.
- 5 11. The sensor system according to claim 5, wherein said at least one set of electrodes being adapted to sense Electrocardiography (ECG) signals or Electroencephalography (EEG) signals.
12. A method of for sensing an electric potential difference in a sensor system having at
10 least one set of electrodes including an reference electrode providing an electric potential reference and a sensing electrode providing a measurement point for measuring an electrical potential relative to said reference electrode, wherein the method comprising:
- 15 - shielding an electrode by placing a shield near but electric insulated from said electrode;
 - generating an input signal by means of said electrode sensing an electric potential;
 - amplifying the input signal received from said electrode in an integrated amplifier providing a buffered path outputting a buffered output signal;
 - 20 - connecting said shield to said output of the integrated amplifier to actively drive the electrical potential of said shield, thereby providing an active shielding of said electrode;
 - frequency shifting the input signal from a basic frequency range to a higher frequency range by means of a first mixer placed in front of the integrated amplifier; and
 - 25 - frequency shifting the amplified signal from said higher frequency range back to said basic frequency range by means of a second mixer placed on the output of the integrated amplifier.
13. The method according to claim 12, and used in a sensor system for sensing an electric
30 potential difference, said sensor system includes an reference electrode providing an electric potential reference, and a sensing electrode providing a measurement point for measuring an electrical potential relative to said reference electrode, and at least one of the reference electrode and sensing electrode being an active electrode, said method comprising:

- receiving in a differential amplifier input from said sensing electrode and said reference electrode, and
 - generating in said differential amplifier an output signal representing the electrical potential difference between said sensing electrode and said reference electrode.
- 5

14. The method according to claim 12, wherein said integrated amplifier generates an output signal replicating the input signal fed to integrated amplifier.

- 10 15. The method according to claim 12, wherein said frequency shifting by means of a first mixer is adapted to avoid the flicker noise region of the integrated amplifier.

Fig. 1

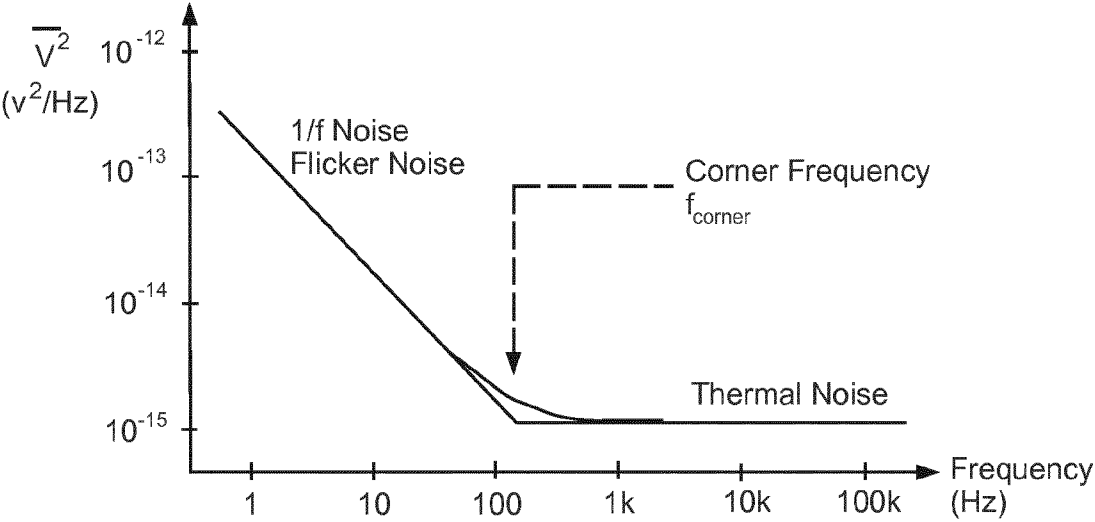
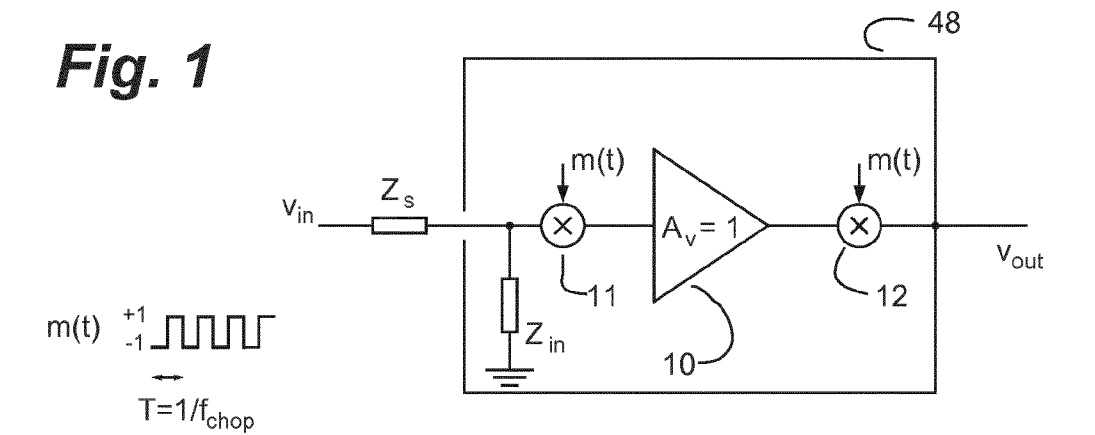
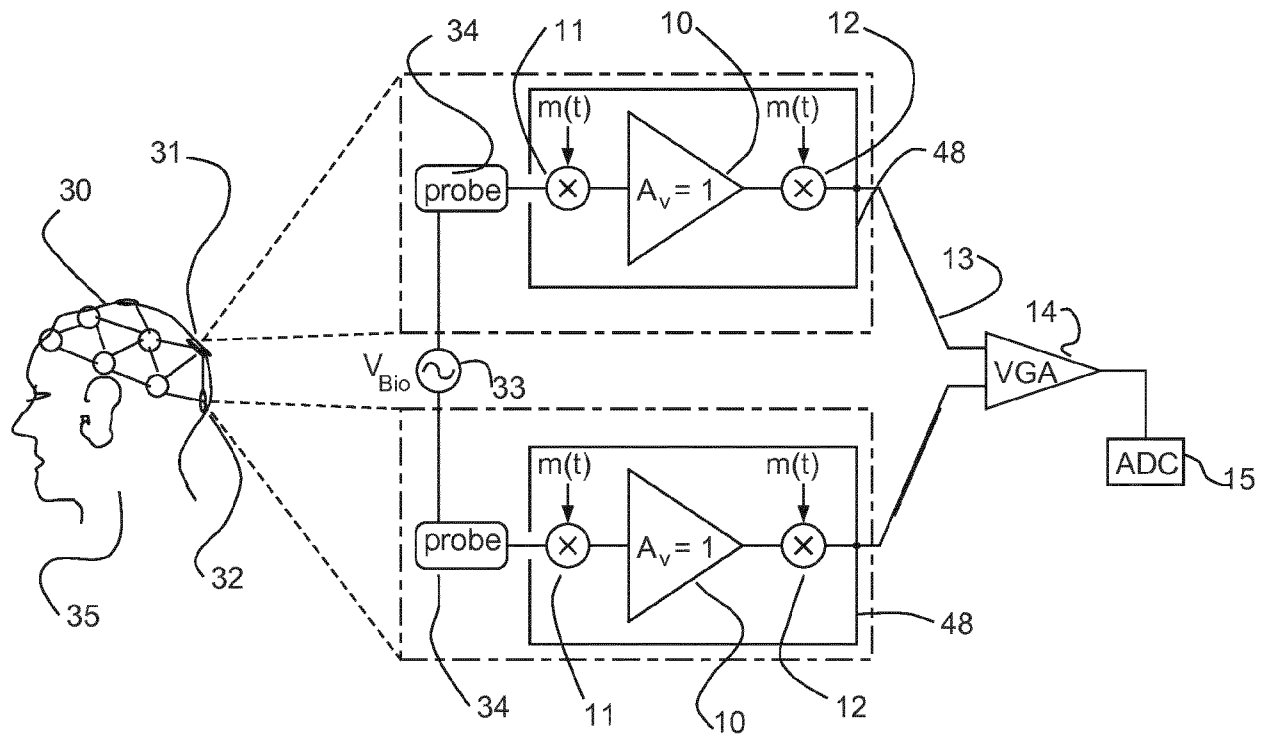
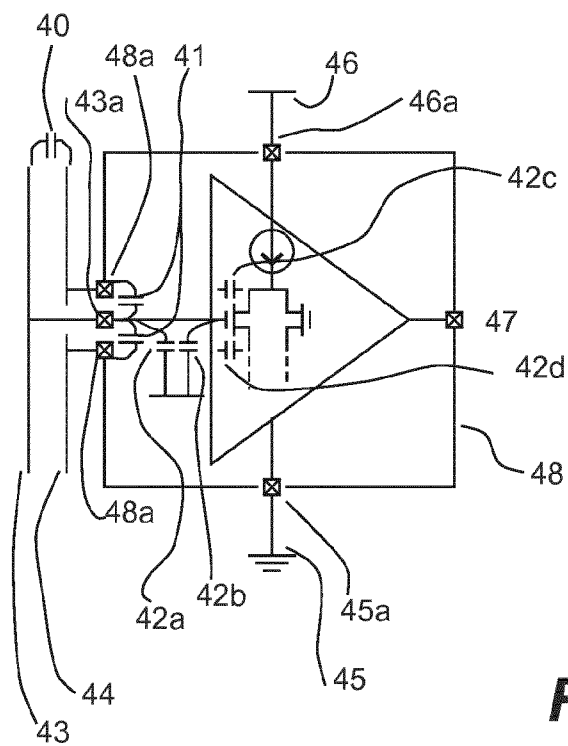


Fig. 2

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**Fig. 3****Fig. 4**

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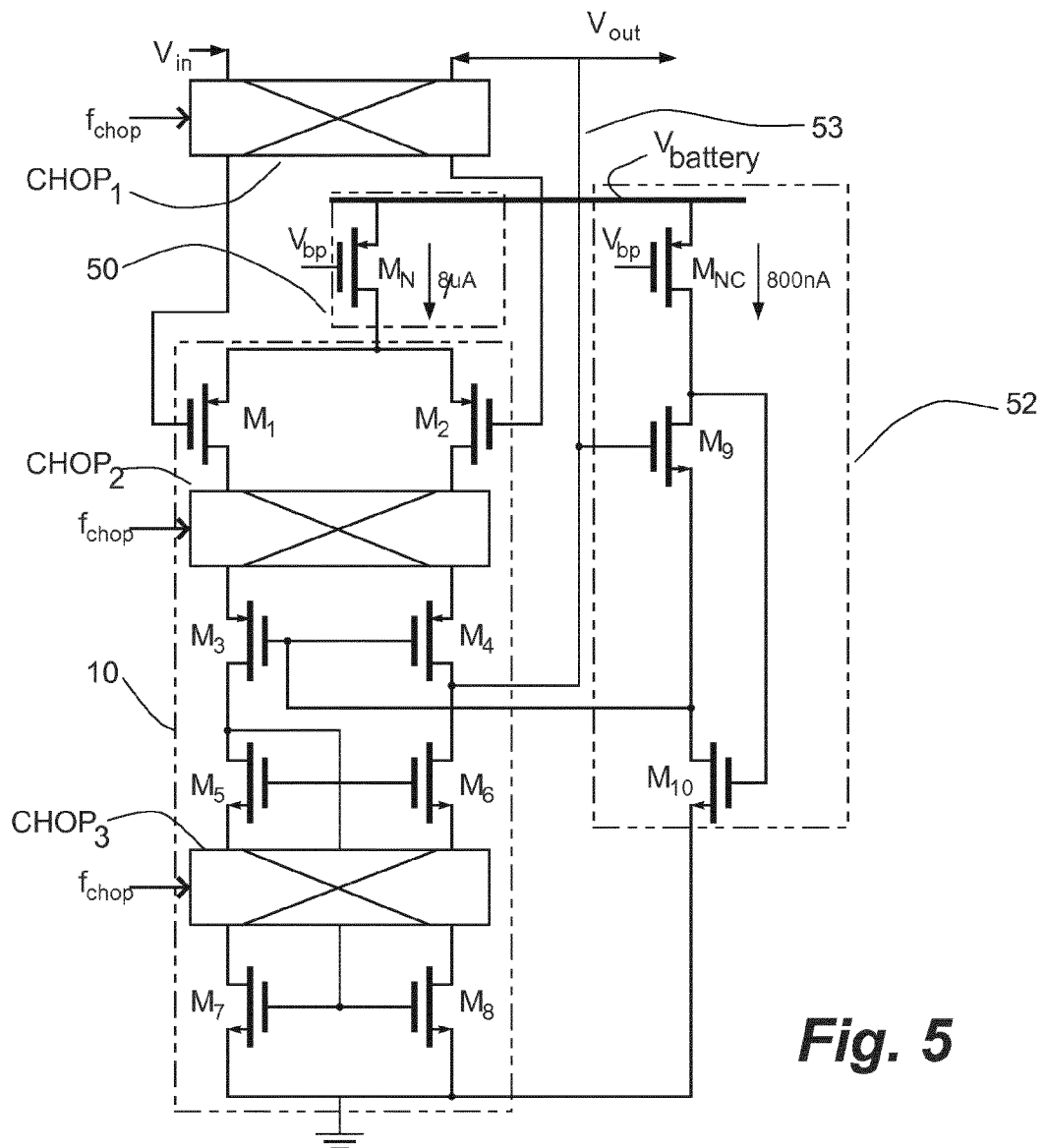


Fig. 5

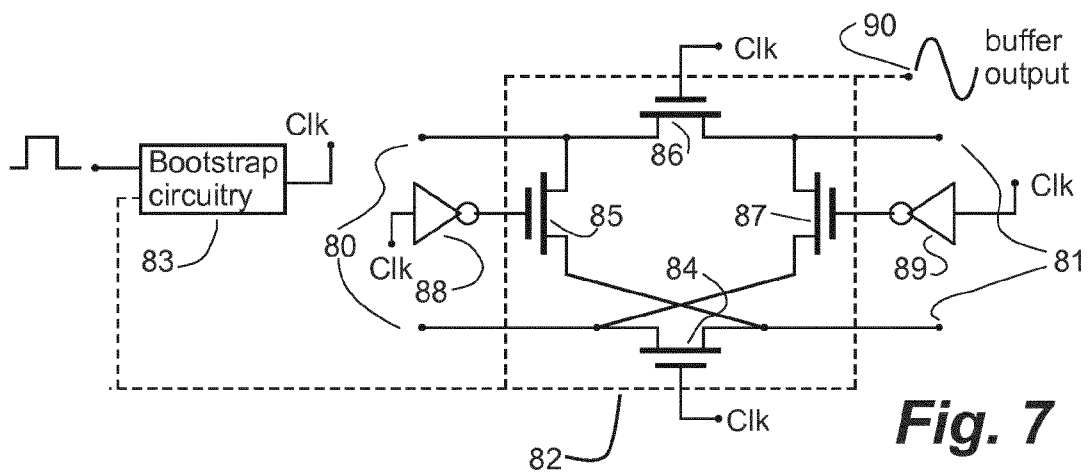
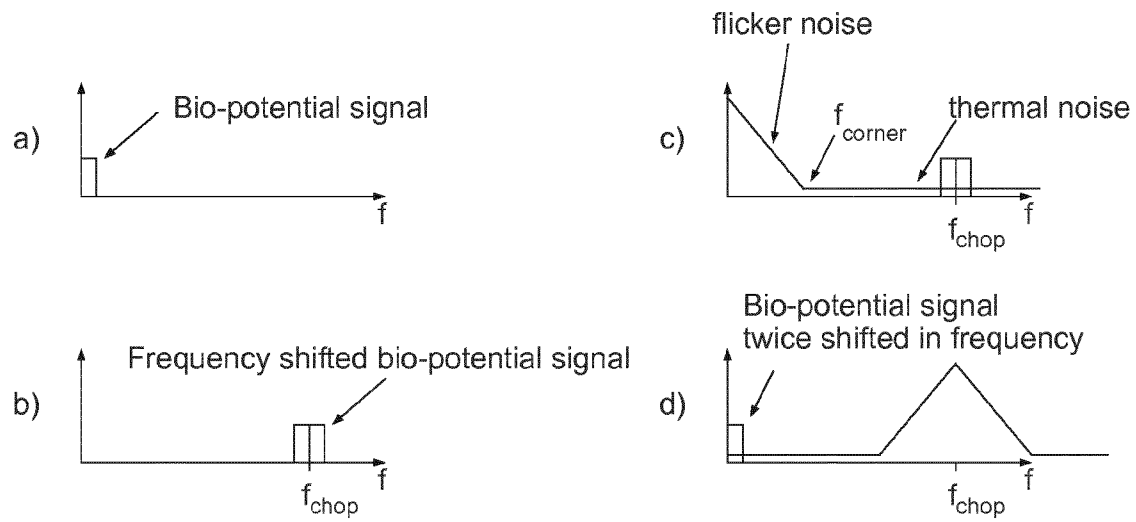
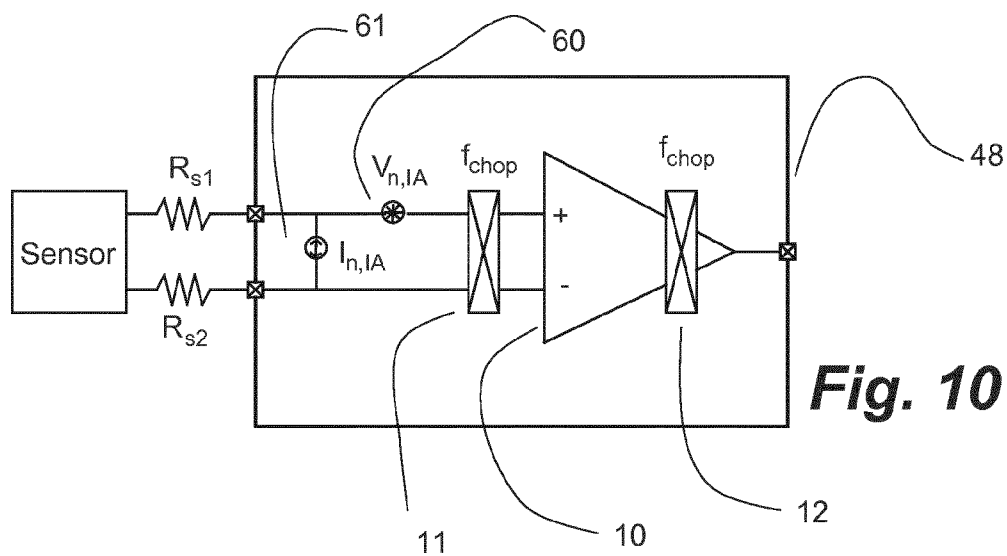
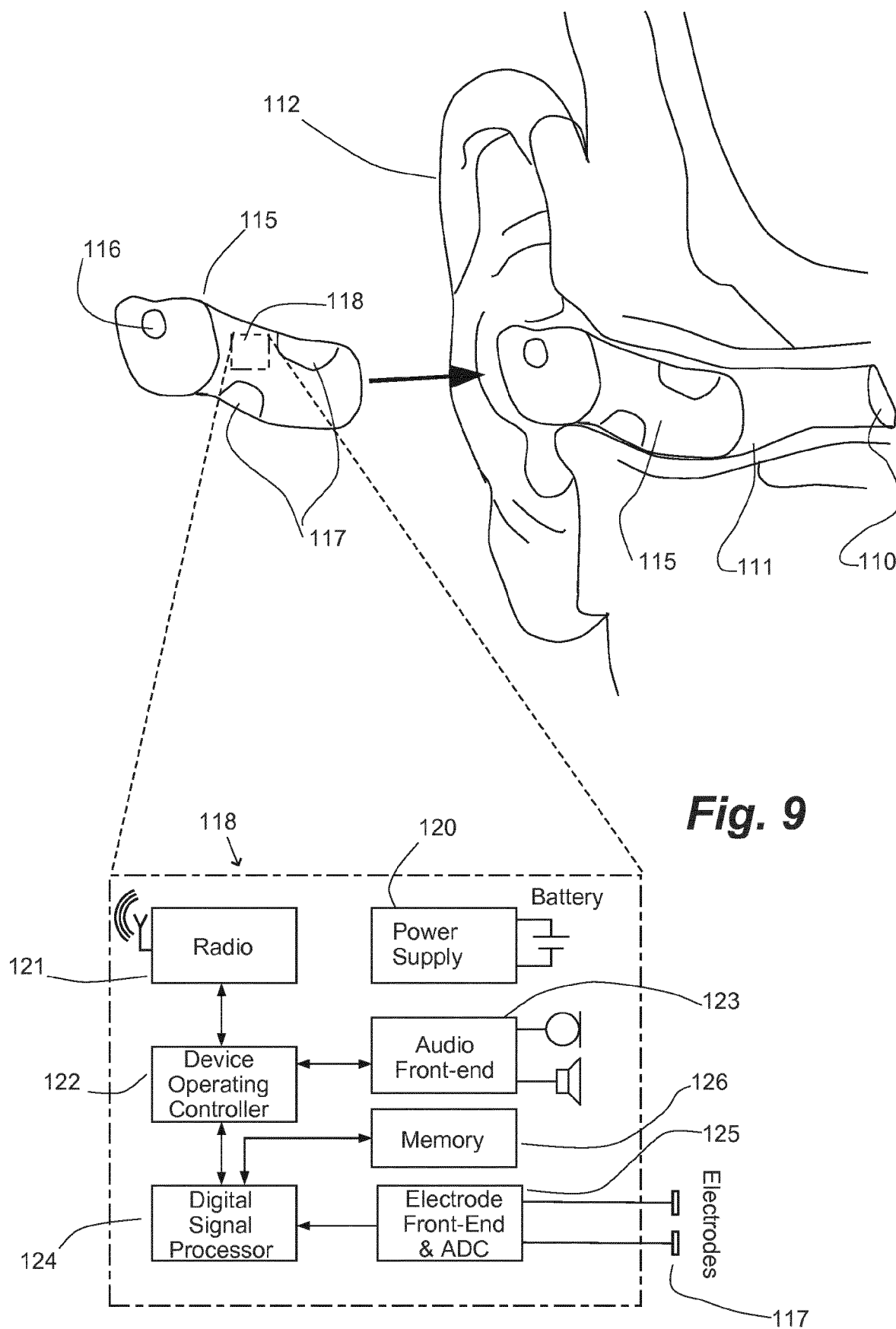


Fig. 7

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**Fig. 6****Fig. 10**

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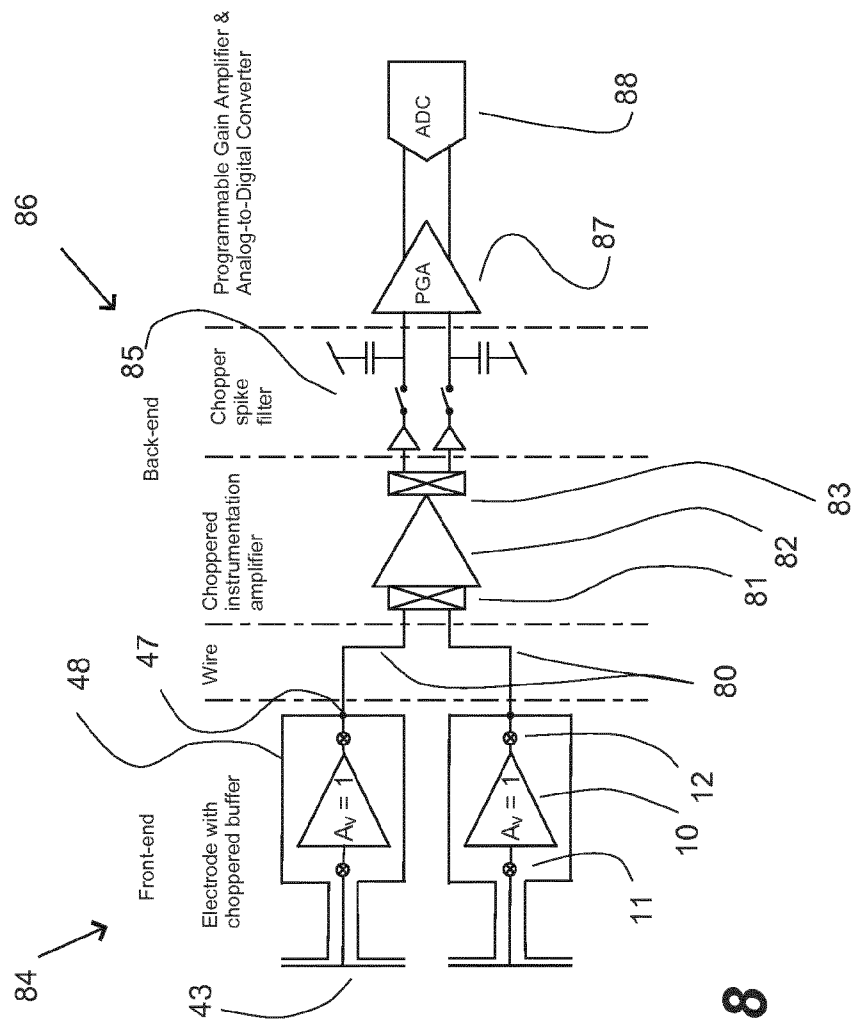


Fig. 8

INTERNATIONAL SEARCH REPORT

International application No
PCT/EP2014/078651

A. CLASSIFICATION OF SUBJECT MATTER
INV. A61B5/0408 H03F3/393 A61B5/0428 A61B5/0448 H03F3/45
ADD.

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
A61B H03F

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

EPO-Internal, WPI Data

C. DOCUMENTS CONSIDERED TO BE RELEVANT

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Y	WO 2013/026481 A1 (WIDEX AS [DK]; KILSGAARD SOREN [DK]; RANK MIKE LIND [DK]; UNGSTRUP MIC) 28 February 2013 (2013-02-28) abstract; figures 1,7-10 page 12, line 3 - page 14, line 18 ----- -/--	1-15



Further documents are listed in the continuation of Box C.



See patent family annex.

* Special categories of cited documents :

"A" document defining the general state of the art which is not considered to be of particular relevance

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"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art

"&" document member of the same patent family

Date of the actual completion of the international search

25 August 2015

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16/09/2015

Name and mailing address of the ISA/

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INTERNATIONAL SEARCH REPORT

International application No
PCT/EP2014/078651

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT		
Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
Y	ENZ C C ET AL: "CIRCUIT TECHNIQUES FOR REDUCING THE EFFECTS OF OP-AMP IMPERFECTIONS: AUTOZEROING, CORRELATED DOUBLES SAMPLING, AND CHOPPER STABILIZATION", PROCEEDINGS OF THE IEEE, IEEE. NEW YORK, US, vol. 84, no. 11, 1 November 1996 (1996-11-01), pages 1584-1614, XP001122266, ISSN: 0018-9219, DOI: 10.1109/5.542410 page 1590 - page 1594; figures 9-11 -----	1-15
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Information on patent family members

International application No

PCT/EP2014/078651

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专利名称(译)	具有带斩波调制的闭环单位增益放大器的有源电极		
公开(公告)号	KR102015716B1	公开(公告)日	2019-08-28
申请号	KR1020177019483	申请日	2014-12-19
[标]申请(专利权)人(译)	T&W工程公司		
申请(专利权)人(译)	티앤더블유엔지니어링에이/에스		
当前申请(专利权)人(译)	티앤더블유엔지니어링에이/에스		
发明人	키드모세 프레벤 주 시웅 킬스카아르드 소렌 리 키앙		
IPC分类号	A61B5/0478 A61B5/00 A61B5/04 H03F3/393 H03F3/45		
CPC分类号	A61B5/0478 A61B5/04004 A61B5/6803 A61B5/6817 H03F3/393 H03F3/45179 H03F3/45183 H03F3/45475 H03F2200/261 H03F2200/271 H03F2203/45306 A61B5/0408 A61B2562/0209 H03F1/26 H03F1/523 H03G3/20 H04R25/353 H04R25/505 H04R2225/43 H04R2225/67		
代理人(译)	Gimtaehong Gimjinhoe		
审查员(译)	蔡，宋 - 洙		
其他公开文献	KR1020170095333A		
外部链接	Espacenet		

摘要(译)

有源电极具有用于感测电势并产生输入信号的电极，以及布置在电极附近但是与电极电绝缘的屏蔽。集成放大器10的输入耦合到至少一个电极，用于提供缓冲路径以接收输入信号并输出缓冲的输出信号。屏蔽耦合到集成放大器的输出，以主动驱动屏蔽的电位，从而提供电极的有源屏蔽。缓冲路径包括在集成放大器前面的第一混频器11，用于将输入信号从基本频率范围移频到较高的频率范围，并且将放大的信号从较高频率范围移频回基本频率的范围。集成放大器输出上的第二个混频器。有源电极可用于记录脑电信号。