(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) World Intellectual Property Organization
International Bureau

(43) International Publication Date
13 October 2016 (13.10.2016)

WIPO | PCT

(10) International Publication Number
WO 2016/161484 A2

(51) International Patent Classification:
A61B 3/00 (2006.01)

(21) International Application Number:
PCT/AU2016/050263

(22) International Filing Date:
8 April 2016 (08.04.2016)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
2015901270 9 April 2015 (09.04.2015) AU

(71) Applicant: **SALUDA MEDICAL PTY LTD** [AU/AU]; Level 1, 407 Pacific Highway, Artarmon, New South Wales 2064 (AU).

(72) Inventor: **PARKER, John Louis**; Level 1, 407 Pacific Highway, Artarmon, New South Wales 2064 (AU).

(74) Agent: **MONKS IP**; PO Box 164, Blackheath, New South Wales 2785 (AU).

(81) Designated States *(unless otherwise indicated, for every kind of national protection available)*: AE, AG, AL, AM, AO, AT, AU, AZ, BA, BB, BG, BH, BN, BR, BW, BY, BZ, CA, CH, CL, CN, CO, CR, CU, CZ, DE, DK, DM, DO, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, GT, HN, HR, HU, ID, IL, IN, IR, IS, JP, KE, KG, KN, KP, KR, KZ, LA, LC, LK, LR, LS, LU, LY, MA, MD, ME, MG, MK, MN, MW, MX, MY, MZ, NA, NG, NI, NO, NZ, OM, PA, PE, PG, PH, PL, PT, QA, RO, RS, RU, RW, SA, SC, SD, SE, SG, SK, SL, SM, ST, SV, SY, TH, TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, ZA, ZM, ZW.

(84) Designated States *(unless otherwise indicated, for every kind of regional protection available)*: ARIPO (BW, GH, GM, KE, LR, LS, MW, MZ, NA, RW, SD, SL, ST, SZ, TZ, UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, RU, TJ, TM), European (AL, AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HR, HU, IE, IS, IT, LT, LU, LV, MC, MK, MT, NL, NO, PL, PT, RO, RS, SE, SI, SK, SM, TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, KM, ML, MR, NE, SN, TD, TG).

Published:

— *without international search report and to be republished upon receipt of that report (Rule 48.2(g))*

(54) Title: ELECTRODE TO NERVE DISTANCE ESTIMATION

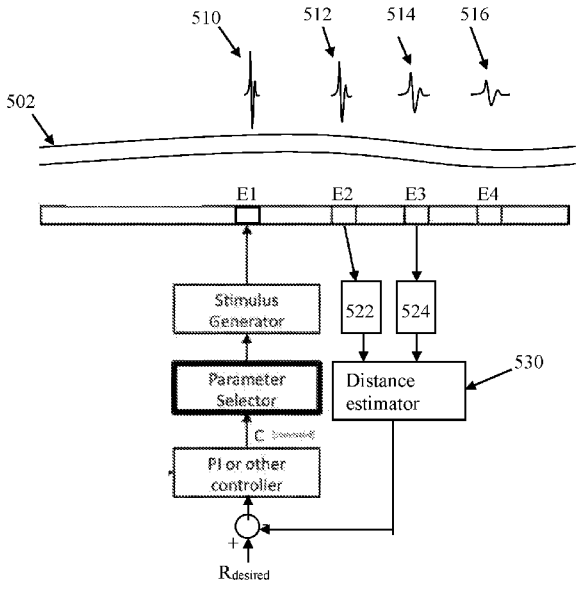

Figure 5

(57) Abstract: Estimating a nerve-to-electrode distance involves applying a stimulus from a stimulus electrode to a nerve. Neural measurements of at least one evoked compound action potential are obtained, and processed in order to estimate an originating state of stimulation exhibiting at least one characteristic defined by a single fibre size. A single fibre model is then applied to produce a measure of the nerve-to-electrode distance. Also provided for is estimation of a distribution of recruited fibres. Measurements of a compound action potential are obtained from sense electrodes spaced apart along a neural pathway. A conduction velocity of the compound action potential is determined from the latency between the measurements. From the conduction velocity a dominant recruited fibre diameter is determined. A rate of dispersion of the compound action potential between the sense electrodes is determined. From the rate of dispersion a distribution of diameters of the recruited fibre population is determined.

WO 2016/161484 A2

ELECTRODE TO NERVE DISTANCE ESTIMATION

Cross-Reference To Related Applications

[0001] This application claims the benefit of Australian Provisional Patent Application No. 2015901270 filed 9 April 2015, which is incorporated herein by reference.

Technical Field

[0002] The present invention relates to neurostimulation, and in particular relates to observing evoked compound action potentials caused by electrical stimuli, in order to estimate a distance, or a change in distance, between a nerve and an electrode being used to stimulate the nerve.

Background of the Invention

[0003] There are a range of situations in which it is desirable to apply neural stimuli in order to give rise to a compound action potential (CAP). For example, neuromodulation is used to treat a variety of disorders including chronic pain, Parkinson's disease, and migraine. A neuromodulation system applies an electrical pulse to tissue in order to generate a therapeutic effect. When used to relieve chronic pain, the electrical pulse is applied to the dorsal column (DC) of the spinal cord. Such a system typically comprises an implanted electrical pulse generator, and a power source such as a battery that may be rechargeable by transcutaneous inductive transfer. An electrode array is connected to the pulse generator, and is positioned in the dorsal epidural space above the dorsal column. An electrical pulse applied to the dorsal column by an electrode causes the depolarisation of neurons, and generation of propagating action potentials. The fibres being stimulated in this way inhibit the transmission of pain from that segment in the spinal cord to the brain. To sustain the pain relief effects, stimuli are applied substantially continuously, for example at 100 Hz.

[0004] Neuromodulation may also be used to stimulate efferent fibres, for example to induce motor functions. In general, the electrical stimulus generated in a neuromodulation system triggers a neural action potential which then has either an inhibitory or excitatory effect. Inhibitory effects can be used to modulate an undesired process such as the transmission of pain, or to cause a desired effect such as the contraction of a muscle.

[0005] For a number of reasons it is desirable to be able to determine the distance of a nerve fibre responding to electrical stimulation from the stimulating electrode. Conventionally, spinal cord stimulation (SCS) delivers stimulation to the dorsal column at a fixed current. When a subject moves or changes posture the distance between the spinal cord and the implanted

electrode array varies, resulting in an increase or decrease in the amount of current received by the dorsal columns. These changes in current result in changes to recruitment and paraesthesia, which can reduce the therapeutic effect of SCS and can create side effects including over-stimulation.

[0006] If a stimulus is of an amplitude and/or peak width and/or has other parameter settings which put it below the recruitment threshold, delivery of such a stimulus will fail to recruit any neural response. Thus, for effective and comfortable operation, it is necessary to maintain stimuli amplitude or delivered charge above the recruitment threshold. It is also necessary to apply stimuli which are below a comfort threshold, above which uncomfortable or painful percepts arise due to increasing recruitment of Aδ fibres which are thinly myelinated sensory nerve fibres associated with joint position, cold and pressure sensation. In almost all neuromodulation applications, a single class of fibre response is desired, but the stimulus waveforms employed can recruit action potentials on other classes of fibres which cause unwanted side effects, such as muscle contraction if motor fibres are recruited. The task of maintaining appropriate stimulus amplitude is made more difficult by electrode migration and/or postural changes of the implant recipient, either of which can significantly alter the neural recruitment arising from a given stimulus, depending on whether the stimulus is applied before or after the change in electrode position or user posture. Postural changes alone can cause a comfortable and effective stimulus regime to become either ineffectual or painful.

[0007] Another control problem, facing neuromodulation systems of all types, is achieving neural recruitment at a sufficient level required for therapeutic effect, but at minimal expenditure of energy. The power consumption of the stimulation paradigm has a direct effect on battery requirements which in turn affects the device's physical size and lifetime. For rechargeable systems, increased power consumption results in more frequent charging and, given that batteries only permit a limited number of charging cycles, ultimately this reduces the implanted lifetime of the device.

[0008] Any discussion of documents, acts, materials, devices, articles or the like which has been included in the present specification is solely for the purpose of providing a context for the present invention. It is not to be taken as an admission that any or all of these matters form part of the prior art base or were common general knowledge in the field relevant to the present invention as it existed before the priority date of each claim of this application.

[0009] Throughout this specification the word "comprise", or variations such as "comprises" or "comprising", will be understood to imply the inclusion of a stated element, integer or step, or group of elements, integers or steps, but not the exclusion of any other element, integer or step, or group of elements, integers or steps.

[0010] In this specification, a statement that an element may be “at least one of” a list of options is to be understood that the element may be any one of the listed options, or may be any combination of two or more of the listed options.

Summary of the Invention

[0011] According to a first aspect the present invention provides a method of estimating a nerve-to-electrode distance, the method comprising:

applying from a stimulus electrode to a nerve at least one stimulus having defined stimulus parameters;

obtaining a plurality of neural measurements of at least one compound action potential evoked by the at least one stimulus;

processing the plurality of neural measurements in order to estimate an originating state of stimulation, the originating state of stimulation exhibiting at least one observable characteristic defined by a single fibre size; and

applying a single fibre model to the estimated originating state of stimulation and the stimulus parameters, in order to produce a measure of the nerve-to-electrode distance.

[0012] According to a second aspect the present invention provides an implantable device for estimating a nerve-to-electrode distance, the device comprising:

at least one stimulus electrode and at least one sense electrode;

measurement circuitry for obtaining a neural measurement from the or each sense electrode; and

a processor configured to apply from the or each stimulus electrode to a nerve at least one stimulus having defined stimulus parameters, obtain from the measurement circuitry a plurality of neural measurements of at least one compound action potential evoked by the at least one stimulus, process the plurality of neural measurements in order to estimate an originating state of stimulation, the originating state of stimulation exhibiting at least one observable characteristic defined by a single fibre size; and apply a single fibre model to the estimated originating state of stimulation and the stimulus parameters, in order to produce a measure of the nerve-to-electrode distance.

[0013] The originating state of stimulation may be considered as a threshold condition of stimulation, at which a single fibre or a single fibre size dominates or defines the nature of the evoked neural response. The present invention recognises that by estimating the originating state of stimulation, it is possible to isolate at least one characteristic which is defined by a single fibre size. Knowledge of an evoked characteristic which is defined solely or largely by a single size of neural fibre in turn enables a single fibre model of recruitment to be applied, in order to estimate the nerve-to-electrode distance. The present invention thus operates to eliminate complicating effects arising from propagation of a compound action potential along a group of neural fibres of distinct size.

[0014] Any suitable single fibre model may be applied to the originating state of stimulation in order to produce the measure of nerve-to-electrode distance. The single fibre model may comprise a lookup table matching the observed characteristic and the stimulus parameters to a corresponding nerve-to-electrode distance.

[0015] The measure of the nerve-to electrode distance may comprise an absolute measure of distance, or a relative measure reflecting a change in distance from a previous time, or a measure of a rate of change of distance.

[0016] In some embodiments of the invention, the method may comprise the further step of adjusting a therapeutic stimulus regime in response to an observed change in the nerve-to-electrode distance.

[0017] In some embodiments of the invention, the method may be performed intra-operatively, as part of a surgical procedure, for example to progressively monitor a position of a structure bearing the electrodes relative to the nerve. In some embodiments the method may be conducted as part of a postoperative fitting procedure of a neurostimulator.

[0018] In some embodiments of the invention the originating state of stimulation may comprise an estimate of the ECAP peak width at the stimulus site. In such embodiments the applying comprises applying a single stimulus in order to evoke a single ECAP, and the plurality of neural measurements are obtained from at least two sense electrodes each at a unique distance away from the stimulus electrode. Such embodiments recognise that an ECAP comprises a compound response made up of contemporaneous action potentials evoked on a plurality of individual nerve fibres, and that each nerve fibre exhibits a conduction velocity which depends at least partly on the diameter of that fibre, so that the ECAP peak width widens at an

approximately linear rate as each individual action potential propagates away from the stimulus site at a unique velocity. Such embodiments preferably estimate an originating ECAP peak width by extrapolating the first and second ECAP measures back to the stimulus site, given that a distance from the stimulus electrode to the first and second sense electrodes is known. Such embodiments of the invention recognise that the originating ECAP peak width can be assumed to be dominated by that single fibre recruited at the stimulus site which had the broadest action potential peak width, typically comprising the recruited fibre of largest diameter as larger fibers are more excitable than smaller diameter fibers. Such embodiments further recognise that the nerve-to-electrode distance can in turn be estimated from the originating ECAP peak width because of the dependence of originating ECAP dispersion upon the fibre to electrode distance.

[0019] Such embodiments, which estimate the originating ECAP peak width or dispersion, may be particularly suitable in applications where the stimulus and sense electrodes are well aligned alongside a neural pathway, such as in the case of SCS.

[0020] The measure of ECAP peak width may comprise a half-height peak width, being a measure of a width of an ECAP peak as observed at an amplitude which is half the amplitude of the peak amplitude of the observed ECAP peak. Alternatively the measure of ECAP peak width may comprise a time between the N1 and P1 peaks of the observed response, and/or a time between the P1 and P2 peaks. Alternatively the measure of ECAP peak width may comprise a time between a zero crossing preceding the N1 peak and a zero crossing following the N1 peak.

[0021] The ECAP peak width may be measured or assessed by extracting frequency components of the neural measurements, for instance fast Fourier transform. Preferably the neural measurements are first windowed to exclude discontinuities or like stimulus effects and/or measurement effects. The frequency domain information of the respective neural measurements may then be used to extract a measure of the dispersion. For example a profile of the frequency domain spectrum of the neural measurements may be assessed for a roll-off or decay with frequency, whereby a faster roll-off of higher frequency components reflects a more dispersed ECAP peak, that is, a peak which is more dominated by lower frequency components. A slope or rate of decay of the frequency roll-off may then be determined for each neural measurement, and used to estimate an originating state of stimulation namely the frequency roll-off present in the evoked response at the site of stimulation. Such embodiments may be advantageous in measuring dispersion in noisy neural measurements, as a frequency roll-off can be averaged or fitted over a relatively wide spectral range. Such embodiments may further be advantageous in

enabling a measure of dispersion to be obtained without reliance on the amplitude of the ECAP, for example in embodiments where manual user feedback or automated feedback operates to control recruitment at a substantially constant level.

[0022] The measure of ECAP peak width may comprise a function of one or more such measures, or may comprise any measure which reflects dispersion of the ECAP over time.

[0023] In some embodiments, neural measurements may be obtained of both orthodromic and antidromic ECAPs, to permit an averaged or more robust estimate of the originating state of stimulation, and thus of the nerve-to-electrode distance estimate, to be obtained.

[0024] Additionally or alternatively, the originating state of stimulation may in some embodiments comprise a stimulus threshold such as the Rheobase. In such embodiments, a stimulus threshold is preferably determined at at least two differing stimulus pulse widths, from which the Rheobase can be calculated. The conduction velocity is preferably measured and used to determine a fibre diameter recruited at threshold. Fitted relationships of the modelled single fibre Rheobase to the electrode-to-nerve separation are then used to determine the separation. Such embodiments may be particularly advantageous in applications providing or permitting only one measurement electrode, as may occur in the brain which does not comprise a single longitudinal neural pathway.

[0025] The originating state of stimulation may in some embodiments be selectively explored in relation to a sub-population of fibres as defined by refractory period. Such embodiments recognise that the fibres within the population of recruited fibres may have different refractory periods. The originating state of stimulation may be estimated in relation to a specific sub-population of the fibres selected for different refractory periods, for example by applying a stimulus sequence comprising a first stimulus referred to as a masker stimulus which recruits all the fibres of interest, and then a short duration later applying a second stimulus referred to as a probe stimulus. The duration between the masker and probe stimuli is selected to be longer than the refractory period of some fibres, but shorter than the refractory period of other fibres. Consequently, the probe stimulus will recruit only those fibres having a short enough refractory period to have recovered from the masker stimuli and able to be recruited a second time by the probe stimulus. In such embodiments the neural measurements are then analysed specifically in relation to the portion of the observed measurement which corresponds with the response evoked by the probe stimulus.

[0026] Embodiments of the invention may thus be applied in neural stimulation applications where the separation between the responding fibres and the stimulating electrode varies often or even continuously with patient movement, whereby knowledge of the fibre-to-electrode distance or at least of incremental changes thereof, would be valuable. Other embodiments of the invention may be applied in relation to locating responding fibres three dimensionally in space in order to avoid or locate them during a surgical or imaging procedure for example. In another application it is desirable to be able to locate a target fibre and position an electrode array in optimal position relative to the fibre in order to achieve the most effective stimulation. Such embodiments may further comprise identifying a target nerve fascicle within a larger nerve bundle, at differing locations along the nerve bundle, in order to detect variation in position of the fascicle within the bundle.

[0027] In some embodiments, the ECAP measurements are further used to estimate the distribution of fiber diameters present in an ECAP. An indication of the distribution or spread of fiber diameters can provide a useful validation for computer models and may be used to inform device and algorithm design to improve outcomes for SCS.

[0028] Thus, according to a third aspect, the present invention provides a method of estimating a distribution of fibres recruited by a stimulus, the method comprising

obtaining from at least two sense electrodes spaced apart along a neural pathway respective measurements of a compound action potential propagating along the neural pathway;

determining a conduction velocity of the compound action potential from the latency between the measurements, and determining from the conduction velocity a dominant recruited fibre diameter;

determining a rate of dispersion of the compound action potential between the sense electrodes, and determining from the rate of dispersion a distribution of diameters of the recruited fibre population.

[0029] According to a fourth aspect, the present invention provides a device for estimating a distribution of fibres recruited by a stimulus, the device comprising

at least one stimulus electrode and at least two sense electrodes, configured to be spaced apart along a neural pathway;

measurement circuitry for obtaining a neural measurement from each sense electrode; and

a processor configured to obtain from the at least two sense electrodes respective measurements of a compound action potential propagating along the neural pathway, determine a

conduction velocity of the compound action potential from the latency between the measurements, determine from the conduction velocity a dominant recruited fibre diameter, determine a rate of dispersion of the compound action potential between the sense electrodes, and determine from the rate of dispersion a distribution of diameters of the recruited fibre population.

[0030] In embodiments of the third and fourth aspects of the invention, the rate of dispersion may be determined in any suitable manner described herein, including any one or more of the observed ECAP peak width, ECAP peak spacing, ECAP zero crossings, ECAP half-height peak width or ECAP spectral content.

[0031] The third and fourth aspects of the invention recognize that the overall distribution of fibre diameters, the distribution of fibre diameters recruited by a given stimulus, and/or the recruited fibres' conduction velocities may vary from one subject to the next. Moreover, some embodiments further recognize that variations in such characteristics may be correlated with the neurological condition which brought about the need for neurostimulation: for example, changes in conduction velocity and distribution of fibre diameters in dorsal columns have been recorded in mouse models of neuropathic pain as a result of central sensitization. Some embodiments of the third and fourth aspects of the present invention may thus further comprise treating the neurological condition by administering or modifying a therapy in a manner responsive to the determined distribution of diameters of the recruited fibre population, or responsive to a change in the determined distribution over time.

[0032] According to a further aspect the present invention provides a non-transitory computer readable medium for estimating a nerve-to-electrode distance, comprising instructions which, when executed by one or more processors, causes performance of the following:

applying from a stimulus electrode to a nerve at least one stimulus having defined stimulus parameters;

obtaining a plurality of neural measurements of at least one compound action potential evoked by the at least one stimulus;

processing the plurality of neural measurements in order to estimate an originating state of stimulation, the originating state of stimulation exhibiting at least one observable characteristic defined by a single fibre size; and

applying a single fibre model to the estimated originating state of stimulation and the stimulus parameters, in order to produce a measure of the nerve-to-electrode distance.

[0033] According to a further aspect the present invention provides a non-transitory computer readable medium for estimating a distribution of fibres recruited by a stimulus, comprising instructions which, when executed by one or more processors, causes performance of the following:

obtaining from at least two sense electrodes spaced apart along a neural pathway respective measurements of a compound action potential propagating along the neural pathway;

determining a conduction velocity of the compound action potential from the latency between the measurements, and determining from the conduction velocity a dominant recruited fibre diameter;

determining a rate of dispersion of the compound action potential between the sense electrodes, and determining from the rate of dispersion a distribution of diameters of the recruited fibre population.

Brief Description of the Drawings

[0034] An example of the invention will now be described with reference to the accompanying drawings, in which:

Figure 1 schematically illustrates an implanted spinal cord stimulator;

Figure 2 is a block diagram of the implanted neurostimulator;

Figure 3 is a schematic illustrating interaction of the implanted stimulator with a nerve;

Figure 4 illustrates the typical form of an electrically evoked compound action potential (ECAP) of a healthy subject;

Figure 5 illustrates stimulus of a nerve and dispersion and measurement of a response;

Figure 6 is a plot of computed single fibre action potentials, for 10 different fibre diameters;

Figure 7 illustrates a size distribution of a recruited fibre population;

Figure 8 is a plot of modelled single fibre action potentials, scaled in amplitude according to the distribution of Figure 7;

Figure 9 illustrates the synthetic compound action potential produced by summation of the scaled potentials of Figure 8, at various distances away from the stimulus location;

Figure 10 illustrates the calculated dispersion for a single fibre for a number of differing electrode-to-fibre separations;

Figure 11 illustrates the width at half height of the synthetic ECAPs as observed at increasing distance from the stimulus site;

Figure 12 is a plot of the slope of peak widths relative to four selected population distributions;

Figure 13 illustrates single fibre responses as superimposed at the stimulus site;

Figure 14 is a plot of the action potential peak width for single fibres of varying diameters;

Figure 15a is an overlaid plot of experimentally obtained measurements of a sheep ECAP obtained from spaced apart measurement electrodes, and Figure 15b is a plot of the peak to peak amplitude observed on each such measurement electrode in response to increasing stimulation;

Figure 16a is a plot of the sheep orthodromic responses' N1 peak width at half height as a function of the channel number, and Figure 16b is the corresponding plot for antidromic responses;

Figure 17 is a plot of ECAPs recorded from electrodes placed in the sheep epidural space;

Figure 18 is a plot of the width of the N1 peak plotted against the recording channel number, from the data of Figure 17;

Figures 19a and 19b are plots of ECAPs recorded from 24 electrodes placed in the epidural space of another sheep, in the orthodromic and antidromic direction respectively; and Figures 19c and 19d are plots of response amplitude and response peak width for the recordings of Figures 19a and 19b, respectively;

Figure 20 plots simulated ECAPs produced at varying nerve-to-electrode separation at the site of stimulation, together with experimental data transformed to the site of stimulation;

Figure 21 illustrates a best-fit fibre size distribution profile;

Figure 22 illustrates a synthetic ECAP modelled from the distribution profile of Figure 21, together with observed sheep ECAP profiles, at various distances from the stimulus site;

Figure 23 illustrates best fit distribution profiles determined for sheep ECAPs observed in response to four different stimulus current levels;

Figure 24 illustrates a strength-duration curve;

Figure 25 illustrates an alternative representation of the strength duration curve, for varying electrode-to-nerve separation;

Figure 26 is a plot of the Rheobase current against electrode to fibre separation;

Figure 27 is a plot of a Rheobase-to-height fitting constant against fibre diameter;

Figure 28 presents simulated plots of the relationship of increasing separation upon Rheobase;

Figure 29 schematically depicts Rheobase measurement; and

Figures 30a-30d illustrate another embodiment in which electrode-to-nerve separation is measured by extracting frequency components of neural measurements.

Description of the Preferred Embodiments

[0035] Figure 1 schematically illustrates an implanted spinal cord stimulator 100. Stimulator 100 comprises an electronics module 110 implanted at a suitable location in the patient's lower abdominal area or posterior superior gluteal region, and an electrode assembly 150 implanted within the epidural space and connected to the module 110 by a suitable lead. Numerous aspects of operation of implanted neural device 100 are reconfigurable by an external control device 192. Moreover, implanted neural device 100 serves a data gathering role, with gathered data being communicated to external device 192.

[0036] Figure 2 is a block diagram of the implanted neurostimulator 100. Module 110 contains a battery 112 and a telemetry module 114. In embodiments of the present invention, any suitable type of transcutaneous communication 190, such as infrared (IR), electromagnetic, capacitive and inductive transfer, may be used by telemetry module 114 to transfer power and/or data between an external device 192 and the electronics module 110.

[0037] Module controller 116 has an associated memory 118 storing patient settings 120, control programs 122 and the like. Controller 116 controls a pulse generator 124 to generate stimuli in the form of current pulses in accordance with the patient settings 120 and control programs 122. Electrode selection module 126 switches the generated pulses to the appropriate electrode(s) of electrode array 150, for delivery of the current pulse to the tissue surrounding the selected electrode(s). Measurement circuitry 128 is configured to capture measurements of neural responses sensed at sense electrode(s) of the electrode array as selected by electrode selection module 126.

[0038] Figure 3 is a schematic illustrating interaction of the implanted stimulator 100 with a nerve 180, in this case the spinal cord however alternative embodiments may be positioned adjacent any desired neural tissue including a peripheral nerve, visceral nerve, parasympathetic nerve or a brain structure. Electrode selection module 126 selects a stimulation electrode 2 of electrode array 150 to deliver an electrical current pulse to surrounding tissue including nerve 180, and also selects a return electrode 4 of the array 150 for stimulus current recovery to maintain a zero net charge transfer.

[0039] Delivery of an appropriate stimulus to the nerve 180 evokes a neural response comprising a compound action potential which will propagate along the nerve 180 as illustrated, for therapeutic purposes which in the case of a spinal cord stimulator for chronic pain might be to create paraesthesia at a desired location. To this end the stimulus electrodes are used to deliver

stimuli at 30 Hz. To fit the device, a clinician applies stimuli which produce a sensation that is experienced by the user as a paraesthesia. When the paraesthesia is in a location and of a size which is congruent with the area of the user's body affected by pain, the clinician nominates that configuration for ongoing use.

[0040] The device 100 is further configured to sense the existence and intensity of compound action potentials (CAPs) propagating along nerve 180, whether such CAPs are evoked by the stimulus from electrodes 2 and 4, or otherwise evoked. To this end, any electrodes of the array 150 may be selected by the electrode selection module 126 to serve as measurement electrode 6 and measurement reference electrode 8. Signals sensed by the measurement electrodes 6 and 8 are passed to measurement circuitry 128, which for example may operate in accordance with the teachings of International Patent Application Publication No. WO2012155183 by the present applicant, the content of which is incorporated herein by reference.

[0041] The present invention recognises that the amplitude and morphology of an ECAP measurement depends on a number of factors, including the quantity of recruited fibres contributing to the compound response, the conduction velocity or diameter of each recruited fibre, the separation of the electrode from the fibres in both the radial direction and the axial direction relative to an axis of the fibre, and the separation of the measurement electrode(s) from the stimulus electrode(s).

[0042] Here we present methods to determine the separation of fibres from stimulation electrodes based on measurement of ECAPs. There are a number of techniques which can be used to eliminate variables in order to isolate the nerve-to-electrode distance *d*.

[0043] A first such technique is to estimate characteristics of the ECAP response as it existed when first evoked directly under or adjacent to the stimulation electrode 2. In this way, the effect of propagation of the response can be eliminated, allowing an estimation of the separation from the threshold current and conduction velocity of the fibre. Thus, the present embodiment of the invention recognises that the ECAP response as it first existed directly adjacent the stimulus electrode 2 is one type of an originating state of stimulation which can be useful in estimating *d*.

[0044] Figure 4 illustrates the typical form of an electrically evoked compound action potential (ECAP) of a healthy subject. The shape of the compound action potential shown in Figure 4 is predictable because it is a result of the ion currents produced by the ensemble of axons generating action potentials in response to stimulation. The action potentials generated

among a large number of fibres sum to form a compound action potential (CAP). The CAP is the sum of responses from a large number of single fibre action potentials. The CAP recorded is the result of a large number of different fibres depolarising. The propagation velocity of the action potential on each fibre is determined largely by the diameter of that fibre. The CAP generated from the firing of a group of similar fibres is measured as a positive peak potential P1, then a negative peak N1, followed by a second positive peak P2. This is caused by the region of activation passing the recording electrode as the action potentials propagate along the individual fibres. An observed CAP signal will typically have a maximum amplitude in the range of microvolts.

[0045] The CAP profile takes a typical form and can be characterised by any suitable parameter(s) of which some are indicated in Figure 4. Depending on the polarity of recording, a normal recorded profile may take an inverse form to that shown in Figure 4, i.e. having two negative peaks N1 and N2, and one positive peak P1.

[0046] In this embodiment, electrical stimuli are delivered to the spinal cord 502 by one or more stimulus electrodes denoted E1 in Figure 5. A desired degree of recruitment, $R_{desired}$, is input by the user or by a setting made by a clinician when fitting the device or by any other suitable means for defining desired recruitment. $R_{desired}$ is processed by a controller and selector and passed to a stimulus generator which generates a stimulus to be delivered to the neural tissue by electrode E1. As will be appreciated, while only a single stimulus electrode E1 is shown in Figure 5, a bipolar, monopolar or tripolar stimulus may be applied in conjunction with other stimulus electrodes, not shown. At the stimulus site adjacent to E1 within the spinal cord 202, a neural response 510 is evoked by the stimulus.

[0047] The neural response evoked by the stimulus at E1 is a compound response comprising the individual responses evoked in a number of fibres, and takes a form shown at 510. The evoked response 510 propagates along the recruited fibres within the spinal cord 502 away from the stimulus site adjacent to E1, and in so doing the form or morphology of the compound response alters or decays. Without intending to be limited by theory, the decay in the neural response as it travels is at least in part due to a spreading of the compound response along the spinal cord 502 resulting from each recruited fibre having a conduction velocity which differs from the conduction velocity of other recruited fibres. The alteration or decay in the morphology of the observed neural response as it travels is also in part due to a spreading of the compound response across the cross section of the spinal cord 502 due to the variation in depth of the

recruited fibres within the cord 502 at different positions along the cord. At a time t_2 the compound response passes sense electrode E2 and is recorded as having an amplitude and duration indicated at 512, which differs from the form of the response at 510 in that response 512 is of reduced amplitude and greater width or duration. At a later time t_3, after undergoing further spreading and decay, the compound response passes sense electrode E3 and is recorded as having an amplitude and duration indicated at 514. Observed response 514 is of lesser amplitude but greater duration then observed response 512. Similarly, at a later time t_4, after undergoing further spreading and decay, the compound response passes electrode E4 and is recorded as having a further decreased amplitude and increased duration as indicated at 516. Observed response 516 is of lesser amplitude but greater duration then observed response 514.

[0048] It is to be appreciated that the form of each observed response, as shown at 510, 512, 514 and 516, is illustrative. The decay and spreading observed in any neural response will depend at least upon the characteristics of the fibre population actually recruited by the stimulus, the neurophysiology of the subject, and the distance of the electrodes from the fibres.

[0049] In accordance with the present invention, electrodes E2 and E3 are used to obtain a first measurement 512 and a second measurement 514 of the neural response evoked by the stimulus, via measurement circuitry 522, 524 respectively. The evoked CAP measurements in this embodiment are made by use of the neural response measurement techniques set out in International Patent Publication No. WO2012/155183, with two data channels recording simultaneous data from the two electrodes E2 and E3.

[0050] An improved knowledge of the electrophysiological response may lead to explanations of the large variability which is observed in outcomes from SCS and may provide valuable insight into electrode and device design, and improved stimulation algorithms.

[0051] Without intending to be limited by theory, it is noted that the total potential electric field external to and produced from a single nerve fibre, including fast Na, persistent Na and slow potassium channels and myelin properties, can be modelled by:

$$\varphi(t) = \frac{1}{4\pi\rho}\sum_{n=-\infty}^{\infty}\frac{I_m(t-x_n)/v}{\sqrt{h^2+(x-x_n)^2}}$$

where

h is the distance of the measurement electrode from the fibre

x_n is the x co-ordinate of each node of Ranvier

I_m is the current produced by each node

t is time

v is the conduction velocity of the fibre.

[0052] For very small h, the field amplitude is inversely proportional to h, as the field is dominated by a single node of Ranvier. As the electrode is moved away the amplitude decreases and the relationship changes to a power law as the measurement electrode is influenced by the fields produced by more nodes. The shape of the action potential also changes with distance to the measuring electrode. The action current I_m is weighted and summed at the measurement electrode, but with different delays for each of the nodes x_n. The weights change because of the increase in distance from the node to the electrode. This looks like a filter (I/r).

[0053] Several suitable models exist for assessing single fibre behaviour, such as models based on Hodgkin Huxley cable models, and any such single fibre model may be used in embodiments of the present invention. With suitably chosen parameters for the ion channel gating functions, Figure 6 is a plot of computed single fibre action potentials, for 10 different fibre diameters from 19 μm to 8.7 μm. Specifically, plot 602 shows the computed single fibre action potential for a 19 μm diameter fibre, and plot 604 shows the computed single fibre action potential for a 8.7 μm diameter fibre. The large diameter fibres conduct at the highest velocity and the smaller diameter fibres have progressively longer latency and are progressively smaller in size. To calculate a compound action potential requires an estimate of the number of fibres for each size of fibre. The compound action potential is then simply the sum of contributions from all those fibres. That is, the ECAP consists of the contributions of the electrical activity from all the recruited fibers, where the response from a single fiber is referred to as the single fiber action potential (SFAP).

[0054] Calculations were made with the modelled measurement electrode positioned from 35mm to 84mm away from the stimulation electrode along the neural pathway, at increments of 7mm. Both the measurement and sense electrodes are modelled as being located directly above, and separated by a nominated distance (h) from, the modelled fibre. A population distribution was generated as a function of fibre diameter, as shown in Figure 7, in which the Y axis is an arbitrary scale plotted against the diameter of the fibre. The population distribution of Figure 7 comprises fibres as small as 15 μm, up to 21 μm, in the relative proportions shown.

[0055] Figure 8 is a plot of each of the single fibre potentials, scaled by the population distribution of Figure 7 for the modelled fibre diameters, at a single location. The electrode-to-nerve distance was modelled as 6 mm, and Figure 9 illustrates the synthetic compound action

potential calculated at multiple electrode locations by summation of the single fibre responses at each electrode location such as those shown in Figure 8 for a single location. In particular, Figure 9 shows such a synthetic compound action potential as observed at each of the electrodes positioned along the nerve and 6 mm away from the nerve, and at a distance of 35mm to 84mm away from the stimulus site, respectively. Specifically, plot 902 shows the computed compound action potential as observed at the electrode 35 mm away from the stimulus site, and plot 904 shows the computed compound action potential as observed at the electrode 84 mm away from the stimulus site, with interposed plots shown for respective interposed electrodes. As can be seen in Figure 9, the simulated compound action potential decays (reduces in amplitude) and disperses (widens) as it travels away from the stimulus site.

[0056] A convenient measure of the dispersion of the ECAP is to measure width of the N1 peak at half height, as indicated at 410 in Figure 4. The observed dispersion is related to the separation of the measurement electrode from the fibre, whereby for a given single action potential or compound action potential a narrower dispersion is observed at a smaller electrode-to-fibre separation and a wider dispersion is observed at a larger separation. This is a result of the previously discussed effect that the closer the electrode is to the fibre, the more the signal observed by the sense electrode is dominated by the nodes of Ranvier that are closest to the electrode. As the electrode-to-fibre distance increases, the signal present at the sense electrode becomes influenced by more nodes of Ranvier positioned along the fibre, dispersing the observed response. This effect sums over many fibres and thus also occurs when measuring contributions from many fibres as is the case in ECAP measurement. However the observed dispersion also depends on the contribution from fibres of different diameter, which each conduct at different velocities. The present embodiment recognises that the relative influence of the fibre population distribution, on one hand, and the relative influence of the height above the cord, on the other hand, can be separated.

[0057] Figure 10 illustrates the calculated dispersion for a single fibre, being the width 410 of the N1 peak at half height, for a number of differing electrode-to-fibre separations. The relationship is linear and varies by a factor of 4 over the range of separations calculated. However in practice recruitment of a single fibre by device 100 is impossible and the contribution of multiple recruited fibres, of varying diameter, must be taken into account for any practical observations. To this end, synthetic ECAPS were generated for a number of differing fibre populations, for two cases: measurement electrodes positioned at 3mm above the axon, and at 6mm above the axon. Figure 11 illustrates the width at half height of the resulting synthetic

ECAPs as observed at channels (electrodes) 5 through 16 at increasing distance from the stimulus site.

[0058] As can be seen in Figure 11, the plot of dispersion of the ECAP varies considerably with changes in the recruited fibre population, even for unchanged electrode-to-nerve separation. For synthetic observations 1110, which all relate to an electrode-to-nerve separation of 6mm, the variation in recruited fibre population can give dispersion as little as 50 μs between channel 5 and 16 in the case of observation 1112, or as large as 100 μs between channel 5 and 16 in the case of observation 1114. The modelled fibre population distributions at each height comprise a distribution width of 6, 7, 8 or 9 μm, whereby a distribution with an increased number of smaller fibres, and having fibres of a smaller diameter, gives rise to an increased slope of the width at half height of the N1 peak in relation to the propagation distance. Similar variation can be seen in the synthetic observations 1120, which all relate to an electrode-to-nerve separation of 3 mm, for the same four selected fibre distributions.

[0059] Figure 11 shows that the width at half depth of the N1 peak, and thus the CAP dispersion, has a linear dependence on the propagation distance, increasing with propagation distance due to impact of the smaller diameter fibres travelling at slower speeds and increasing the width or dispersion of the peak.

[0060] Figure 12 is a plot of the slope, in ms per channel, of the peak widths relative to the four selected population distributions from the same data as Figure 11. As can be seen, as the population distributions are widened by the addition of smaller fibres, the resulting dispersion increases.

[0061] Thus dispersion alone cannot be used to determine electrode to nerve separation because the recruited fibre population's size distribution is an unknown. However, referring again to Figure 11, the present embodiment recognises that a line or curve fitted to the observed ECAP peak widths and extrapolated to the stimulus site (channel "0" in Figure 11), gives a value which is substantially independent of the propagation dispersion effects. In particular, all of the curves 1110 meet the y-axis of Figure 11 at substantially a first point around 125 μs, while all of the curves 1120 meet the y-axis at substantially a second point around 60 μs. Accordingly, in this embodiment the intercept of the lines 1102 or 1104 with the y-axis at channel "0" is taken as the originating state of stimulation. Importantly, propagation dispersion is effectively eliminated by determining the y-intercept. The present embodiment recognises that the y-intercept value of the lines varies with electrode–to-nerve separation. Moreover, the y-intercept can be obtained in

practice as simply as by applying one stimulus and obtaining as little as two measurements of the ECAP at spaced-apart sense electrodes, as a line can be fitted to two such data points to estimate a y-intercept. Other embodiments may obtain many sense electrode measurements of a single ECAP in order to improve accuracy, or may determine the y-intercept by any suitable means.

[0062] It is further to be noted that the y-intercept value of the lines 1110 and 1120 is impossible to measure directly in practice, as the stimulus applied at the stimulus site is many orders of magnitude larger than the response evoked.

[0063] Accordingly, in some embodiments changes in the y-intercept may be used to indicate relative changes in the electrode-to-nerve distance *d*, even if the absolute value of *d* is not known.

[0064] However, other embodiments further provide for an estimation of the absolute value of the distance *d*, as follows. These embodiments are based on the recognition that the ECAP peak width at Channel 0 (being the stimulus location) is dominated by the width of the single fibre action potential of the largest recruited fibre contributing to the response. For a given nerve, the largest recruited fibre is typically the most easily recruited and can thus be assumed to have been recruited if any ECAP at all is evoked. The action potential peak width of the largest recruited fibre is a constant, but will be observed as a broader peak with increasing fibre to electrode distance *d*. Thus, the peak width of the observed response at channel 0 is dependent on the separation *d* but is substantially independent of the population distribution of the fibres recruited, at least for the range of populations simulated in figure 11.

[0065] Figure 13 shows the action potentials calculated for individual fibres over a range of fibre diameters from 18 micron (SFAP 1302) to 23 micron (SFAP 1304), with intervening fibres' action potentials shown but not labelled. The sum of the SFAPs produces the larger CAP 1306. As shown in Figure 13, the larger diameter fibres produce the largest SFAP responses. When mapped to the stimulus site, the smaller diameter fibres at Channel 0 contribute responses to the compound response 1306 which are enveloped by, or do not significantly affect some key characteristics of, the single fibre response 1304 of the largest contributing fibre diameter.

[0066] Figure 14 is a plot of the single fibre action potential peak width, for single fibres of varying diameters in the range 11 - 23 μm, as indicated by square data points. The circular data point 1402 indicates the synthetic compound action potential peak width occurring at channel 0 for the population of fibres shown in Figure 7. As can be seen from 1402, the 20 μm fibre was

the most abundant in this population distribution and the peak width of the ECAP at 1402 was $1.3x10^{-4}$ s whereas the width of a single 20 μm fibre response was $1.27 x10^{-4}$ s, which is a 2% error in the approximation. The error gets worse with wider distributions of fibre diameters in the fibre population. With a range of fibres from 14μm to 23μm in diameter the width at half depth is 4% larger than the value for the SFAP. The net effect is that the present technique will slightly overestimate the separation of electrode from the responding fibre when the relation evident in figure 14 is used to calculate the corresponding distance.

[0067] To verify the above theoretical approach, animal (sheep) experiments were conducted by epidural implantation of a 24 channel linear electrode array with electrode spacing of 7mm. Current sources were configured to produce tripolar stimulation with a central cathode (channel 2) and anodes on each side (channels 1 & 3). Evoked responses were recorded on electrodes 4 to 24. Figure 15a is an overlay of all of the recordings from channels 4 to 24 in response to stimulation at 0.7 mA. Figure 15b is a plot of the peak to peak amplitude observed on each electrode 4 to 24 in response to increasing stimulation from 0.4 mA to 1.0 mA.

[0068] The ECAP peak width, defined here as the width at half height of the observed N1 peak, was determined on all channels, at various stimulation current levels. Figure 16a is a plot of the sheep orthodromic responses' N1 peak width at half height as a function of the channel number for stimulation currents of 1 mA (squares), 0.9 mA (circles) and 0.8 mA (triangles). Figure 16b is the corresponding plot for the antidromic responses' N1 peak width at half height as a function of the channel number, for stimulation currents of 1.106 mA (squares), 0.996 mA (circles) and 0.801 mA (triangles). Raw neural response measurement data was interpolated in order to remove sampling quantisation effects and improve the estimates of peak width. In each plot the straight line is from least squares fit of the data for all the measurements averaged for each channel. Figure 16 shows that the relationship of the width at half height of the responses with the channel number is substantially independent of the stimulation current. Thus the approach of extrapolating such data to the stimulus channel location (channel 2) is robust to variations in stimulation current and/or to movement induced changes in the recruitment efficacy of a given current.

[0069] In Figure 16 the width at half height (HH) of the responses in the orthodromic direction have a slope with channel number of $5.2x10^{-6}$, and in the antidromic direction the slope is $5.5 x10^{-6}$, which demonstrates that the fibres which are responding in both antidromic and orthodromic directions have similar distributions of fibre diameters.

[0070] In Figure 16a the stimulus channel is channel 2 so that the originating state of stimulation of interest in this embodiment is the peak width at channel 2. Extrapolating the channel 5-20 orthodromic data back to the site of channel 2 gives an estimated channel 2 peak width of 0.00014s (140 μs). In Figure 16b the stimulus channel is channel 20 so that the originating state of stimulation of interest in this embodiment is the peak width at channel 20. Extrapolating the channel 2-17 antidromic data back to the site of channel 20 gives an estimated channel 20 peak width of 0.000125s (125 μs). Taking the average of the orthodromic estimate and the antidromic estimate, and comparing to Figure 10, allows the average electrode-to-fibre distance along the array to be estimated at about 5mm.

[0071] In another experiment ECAPs were recorded from electrodes placed in the sheep epidural space for a stimulation current of 1mA 40 μs pulse width biphasic stimuli. The wave form measured on a single electrode has a duration of less than 1.5ms and the recordings on electrodes which are a short distance from the stimulation electrode are truncated by the blanking period of the amplifier and presence of the stimulus current. Figure 17 shows the obtained recordings. Figure 18 is a plot of the width of the N1 peak plotted against the recording channel number, from the data of Figure 17. As shown by fitted line 1802, the width of the N1 peak is linear with the propagation distance across the first 4 electrodes (channels 3-6). As shown by fitted line 1804 the width of the N1 peak is also linear for the next four electrodes (channels 7-10) albeit with a different, smaller, slope. A corresponding fitted line could be fitted to the final few electrodes, channels 12-14. The y-intercept of lines 1802 and 1804 is, notably, the same: 0.12 ms (120 μs). This represents the width of the ECAP if it could be recorded under the stimulating electrode. Some embodiments of the invention may thus fit a plurality of lines to the ECAP width or dispersion measurements, being one line fitted to each subset of electrodes positioned within each respective vertebral segment. Such embodiments reflect the fact that discontinuities in dispersion appear with propagation distance, accompanied by a change in the slope of the dispersion towards smaller slopes, due to the removal of smaller diameter slower conducting fibres from the recruited population as such fibres terminate at each crossing between vertebral segments. In such embodiments the plurality of fitted lines may each be extrapolated to the stimulus location to estimate the originating peak width, and/or the fitting of such lines may be constrained by a requirement that each line must intersect all others at the stimulus channel, to thereby improve the robustness of the multi-line fitting estimate of the originating state of stimulation, as compared to a single line fitting. In Figure 18 it can be seen by visual inspection that channel 11 most likely is adjacent a vertebral segment crossing in a region where only a subset of that segment's terminating fibres have in fact terminated, so that channel 11 is not

clearly grouped with either channels 7-10 nor channels 12-14. Some embodiments may seek to identify and discard such vertebral segment crossing data points when fitting lines 1802, 1804, etc.

[0072] To further study this effect a 24 channel electrode was implanted in another sheep and antidromic and orthodromic responses were measured, with results shown in Figure 19. Stimulation was tripolar, biphasic. For Figure 19c stimulation was delivered from a cathode on electrode 2 and anodes on channel 1 and 3, and recording electrodes from 5 to 20. For Figure 19d stimulation was delivered from a cathode on electrode 20 and anodes on channel 19 and 21, and recording electrodes from 2 to 17. Figures 19c and 19d show that the discontinuities in the dispersion which appear with the propagation distance arise in both the orthodromic and antidromic conduction directions, consistent with the neuroanatomy of the spinal cord. The present technique may thus be used to assess electrode height not only when stimuli are delivered at the caudal end of the array, but also when stimuli are delivered at the rostral end of the array. When stimuli are delivered from part way along the array, recording electrodes positioned both caudally and rostrally of the stimulus electrode(s) may be used to provide both orthodromic and antidromic estimates of ECAP dispersion to give a combined estimate of the originating ECAP peak width and the electrode height. In the study of Figure 19 there were 4-5 electrodes spanning a single vertebral segment and recordings were made across 4 vertebral segments. It is evident when comparing the upper and lower portions of Figure 19c, and of Figure 19d, that both the dip or cornerpoint in the amplitudes and the change in slope of the dispersion plot correspond with the fibres crossing from one verterbral segment to the other. The data points indicated by triangles, squares and circles in the lower plots of Figure 19c and 19d reflect dispersion data obtained in response to stimuli of differing amplitude. Fitted lines 1902, 1904, 1906 and 1908 correspond to each vertebral segment and, despite having different slope, each have the same value at the stimulus electrode, channel 2. The multiple lines fitted to the antidromic data of Figure 19d also have a shared intercept at the stimulus electrode on channel 20, around 0.11 ms, allowing response peak width at the site of the channel 20 stimulus to be robustly estimated, and in turn allowing relative or absolute electrode-to-nerve separation to be measured as discussed elsewhere herein.

[0073] Thus, the above approach allows an absolute value of the electrode-to-fibre distance to be estimated solely from electrical ECAP measurements.

[0074] To further test the validity of the 5mm separation estimate obtained above in relation to Figure 16, the theoretical "channel 0" responses evoked by an electrode positioned at 2mm, 5mm, 8mm and from a nerve fibre were simulated. In Figure 20, the continuous curve 2002 is the simulated response at 2mm separation, continuous curve 2004 is the simulated response at 5mm separation, and 2006 is the simulated response at 8mm separation. The experimental data points shown in Figure 20 are produced by taking the experimental data of Figure 15a, time scaling each channel's observed response to have a peak width equal to the experimentally determined "channel 0" peak width, removing the pre-response latency at that electrode as defined by conduction velocity and electrode distance from the stimulus site by temporally aligning the N1 peak of each response, and normalising the N1 amplitude. As can be seen in Figure 20, the experimentally observed responses when scaled in this manner (a) take substantially the same profile as each other, and (b) coincide very closely with the simulated channel 0 response 2004 which is evoked by the simulated electrode when at a 5mm spacing from the fibre, thereby verifying the estimate of 5mm produced above.

[0075] Another embodiment of the invention further recognises that Figures 6-14 reveal a means by which the size distribution of fibres recruited by a single stimulus may be estimated from ECAP measurements. Referring to Figure 11 and 12, it can be seen that for a given fibre population distribution, the observed slope of the ECAP peak width depends on electrode-to-fibre distance, the slope being larger when the distance d is smaller.

[0076] A further variable which affects the dispersion, or growth in peak width, is the conduction velocity of the recruited fibre. However the conduction velocity can be determined from the latency of the measured responses as is visible in Figure 15a in which the velocity of the N1 peak is $116ms^{-1}$. The SFAP which conducts at this velocity in the model has a diameter of 21μm, which is consistent with the diameter versus conduction velocity slope of 5.4 which has been reported.

[0077] Thus, the conduction velocity observed in Figure 15a enables a determination to be made as to the diameter of the most abundant fibre in the recruited fibre population, in this case 21μm. Next, the slope of the width at half height observed in Figure 16a and 16b enables the relationship shown in Figure 12 to be used to estimate the distribution of recruited fibres. Finally, a profile of the distribution of recruited fibres may be produced. This involves taking a nominal distribution profile, and summing the individual fibre contributions to a simulated ECAP using a chosen single fibre model, and fitting the simulated ECAP to an observed ECAP

by trial and error adjustment of the nominal profile, until a best fit is found. In the case of the sheep data of Figure 15a, the best fit fibre distribution profile determined in this manner is shown in Figure 21. Figure 22 comprises plots 2202 of the simulated ECAP resulting from single fibre modelling and summation of a recruited fibre distribution having the profile shown in Figure 21, together with plots 2204 of the actual observed sheep ECAP data, as observed at electrodes at distances from 35 mm to 84 mm away from the stimulus site. As can be seen, the best fit fibre distribution profile of Figure 21 when simulated results in a close fit of the simulated ECAP 2204 to the observed sheep ECAP 2202, and such fitting thus enables the fibre distribution profile to be estimated. Any suitable fitting technique, such as a pointwise least squares fitting, may be applied to determine which nominal fibre distribution profile gives the best fit to an observed ECAP.

[0078] Accordingly, some embodiments of the invention may additionally or alternatively seek to use response dispersion to estimate the recruited fibre population's dominant fibre size, and also the width of the distribution of fibre sizes recruited.

[0079] Thus the response of the sheep spinal cord to SCS demonstrates a consistent increasing distribution of fiber velocities with increasing current. These techniques are also applicable to use in humans, where detailed understanding of the electrophysiological response of the spinal cord to electrical stimulation, and the distribution of fiber diameters in chronic pain sufferers, may lead to better diagnostic and patient programming outcomes.

[0080] The effect of increasing the stimulation current, as observed in Figure 15b, on the profile of the best-fit distribution was also determined. The best fit distribution profiles determined for each current level are shown in Figure 23. Figure 23 shows that the population of responding fibres varies with the changes in current, and in particular as the current is increased the proportion of smaller diameter (18-19 μm) fibres contributing to the ECAP is expected to increase.

[0081] Moreover, once the fibre distribution characteristics are known, including the dominant fibre size recruited (in Figure 23 being 20-21 μm), distribution width (17-23 μm) and distribution profile (as shown in Figure 23), single fibre modelling and summation even enables the number of fibres of each size which are being recruited to be estimated, by amplitude comparison to the observed ECAP. Thus, as shown in the y-axis of Figure 23 it can be determined for the sheep data that at peak current about 300 18 μm fibres were recruited and about 550 21 μm fibres were recruited, for example. Similarly, the total number of fibres

recruited is simply the integral of the population distribution curve and for the sheep spinal cord the total number of recruited fibres is 2080 fibres at 1.0mA, 1780 fibres at 0.9 mA, 1440 fibres at 0.8mA and 545 fibres at 0.7mA.

[0082] In yet another embodiment, a technique which can be used to estimate the nerve-to-electrode distance *d* involves probing the Rheobase by delivering appropriate stimuli and measuring the neural responses thereto.

[0083] A plot of the threshold current required to evoke a response against the pulse width is a strength duration curve as shown in Figure 24. The Rheobase current is defined as the maximum current at infinite pulse width which doesn't evoke a response. The present embodiment recognises that the Rheobase current depends on the separation of the electrode from the nerve, so that the curve of Figure 24 shifts towards the origin for a small separation, and shifts away from the origin (up and to the right in Figure 24) for larger separations.

[0084] Fig 25 comprises an alternative representation of the strength duration curve of Figure 24, by plotting applied charge against pulse width, for a 10 micron diameter fibre. Further, this figure includes a plot at a number of electrode separations from the fibre, namely a plot 2502 for a separation of 3 mm, 2504 for 4 mm, 2506 for 5 mm, 2508 for 6 mm, 2510 for 7 mm and 2512 for 8 mm. In this representation of the strength duration curve, each curve is substantially linear and the slope is equal to the Rheobase current.

[0085] Figure 25 allows a plot, shown in Fig 26, to be derived which represents the relationship of the Rheobase current R to the separation *h* from the fibre. In this instance of a simulated single fibre diameter, the fitted relationship is R = B*h*^A , where A and B are empirically fitted constants. Here A = 0.5385 and B = 10e3.485.

[0086] This process by which Figure 26 was obtained for a single fibre model of a fibre diameter of 10 microns was then repeated for multiple fibre diameters D, namely D = [7,8,9,10,11,12] microns. For these single fibre sizes, the fitting constants A were respectively calculated as taking the values A = [0.55,0.537,0.532,0.5385,0.53,0.51]. This indicates that A is approximately constant with changing fibre diameter at least throughout this range, the average value of A being 0.5329.

[0087] On the other hand, the fitting constants B were respectively calculated as taking values B = [3.399,3.445,3.472,3.485,3.493,3.507]. This indicates that B is monotonic increasing with

increasing fibre diameter, at least in this fibre diameter range. Figure 27 plots the values of B against fibre diameter, and also shows a straight-line-fit to the data points, having the equation B = 0.01991 D + 3.278, where D is the diameter of the fibre.

[0088] Thus, this embodiment applies stimuli of varying pulse width from a first stimulus electrode to determine at least two points on the strength-duration curve, as it exists for the unknown separation *h*. From two such points, the Rheobase R for the first-recruited fibre can be calculated in respect of the first stimulus electrode. Because fibre diameter D is unknown, the Rheobase R alone does not yield *h*.

[0089] Figure 28 shows simulated plots of the relationship of increasing separation upon the expected Rheobase value. As expected, Rheobase generally increases as electrode-to-nerve distance increases. However, for each single fibre diameter, the relationship follows a curve and not linear, and moreover at any given separation the Rheobase value of one fibre diameter differs from a differing fibre diameter. As the observed response is a compound action potential, at any given separation the first-recruited fibre will always be the same, being the largest most proximal fibre to the stimulus electrode. That is, the separation vs Rheobase relationship for the observable compound response will be the same as the curve 2802 of the largest most easily recruited fibre.

[0090] The present embodiment thus further provides for determining a conduction velocity of the evoked response at threshold, as conduction velocity is well related to fibre diameter. For example two sense electrodes spaced apart along a neural pathway may record a time of arrival of an evoked response in order to determine the conduction velocity V. The recruited fibre diameter D can then be determined by the empirically determined relationship D = V / X, where X is typically ascribed a value around 5.4 -6. Knowing D, B can be deduced from Figure 27. Now knowing R, B and A, the equation R = Bh^A can be solved to give the electrode to fibre separation h, as desired. In this embodiment the originating state of stimulation is thus the Rheobase, for which an observable characteristic is defined by a single fibre size as demonstrated by reference to Figure 28.

[0091] In other embodiments, rather than a straight line fit, a curve may be fitted to the data points of Figure 27 to improve B estimation, and such embodiments are within the scope of the present invention.

[0092] There are a number of ways to measure the Rheobase dynamically and in real time such as during SCS. As described previously the Rheobase current can be estimated from the slope of the charge duration curve. The slope estimation requires at least two points along this curve, and to obtain these two points requires estimation of the threshold of response for two different stimulus durations. The threshold measurement can be made in a number of ways, and a simple way to make this measurement, schematically depicted in Figure 29, is to measure the slope of the amplitude of the ECAP with respect to stimulation current, to estimate a threshold current. Determination of the threshold at two pulse widths provides the data necessary to compute the Rheobase current. Thus in this method four stimuli are required, but at least two of the stimuli can be controlled to have the same charge as a required therapeutic level, and thus provide therapeutic stimuli. The other two stimuli required to complete the Rheobase measurement can then be lower in amplitude and thus not uncomfortable and may even be below a perception threshold.

[0093] It is to be appreciated that the stimuli sequence could be applied continuously and the Rheobase calculated continuously and averaged over time, or further signal processing techniques applied to improve the SNR of this measure. The conduction velocity needs only be measured infrequently and for many applications can be measured only once, or on rare occasions, to provide the remaining constant.

[0094] Figures 30a-30d illustrate another embodiment of the invention, in which ECAP dispersion, and electrode-to-nerve separation, are measured or assessed by extracting frequency components of the neural measurements, by fast Fourier transform. The neural measurements are first windowed to exclude discontinuities or like stimulus effects and/or measurement effects. The frequency domain information of the respective neural measurements may then be used to extract a measure of the dispersion. Figure 30a plots simulations of the ECAP of a 12 μm diameter fibre measured 35mm away from the stimulating site and at separations of 2,3,4,5 and 6 mm, respectively, between the electrode and the fibre. Once again, both the amplitude and the dispersion of the observed response changes with separation, with larger separations producing smaller more dispersed responses. Figure 30b shows the Fourier spectrum of the data from the first figure (Haming window). The present embodiment operates by noting that the decay in the frequency response of each observed ECAP at frequencies higher than the peak is linear. It is also noted that the peak amplitude of each curve in Figure 30b, and the spectral spread of each curve, reflects the sharpness of the observed response, and either or both such measures may thus be used as a measure of the inverse of response dispersion. Figure 30c shows the slope of the

decay of frequency contributions to each respective observed response, at frequencies higher than the peak of the respective curve. The present embodiment further notes that the slope of the decay of each curve is proportional to the separation of the fibre from the electrode, whereby a steeper decay slope corresponds to greater dispersion and thus greater separation. Figure 30d is a plot of decay slope against separation, indicating the monotonic nature of this relationship in the separations observed, and for example Figure 30d may be reflected in a lookup table whereby the spectral decay slope observed in a given response may be used to look up electrode-to-nerve separation. Such embodiments may be advantageous in measuring dispersion in noisy neural measurements, as a frequency roll-off can be averaged or fitted over a relatively wide spectral range. Such embodiments may further be advantageous in enabling a measure of dispersion to be obtained without reliance on the amplitude of the ECAP, for example in embodiments where manual user feedback or automated feedback operates to control recruitment at a substantially constant level.

[0095] The stimulation electrodes and sensing electrodes in one embodiment are an array of electrodes. The stimulus location and the measurement location could be changed from one measurement to the next, such as by being scanned across the array with electronic switching means, and the Rheobase / distance computed in real time and from this a two dimensional picture of the underlying neural active elements and their location with respect to the electrodes of the array could be determined.

[0096] An image so produced could in turn be used to guide a surgical procedure, such as the removal of tissue with little or no response such as is performed in DREZ lesion surgery, or detection and removal of aberrantly responding tissue such as the removal of brain lesions responsible for focal origin epilepsy.

[0097] The geometry of the sensing stimulating electrodes need not be planar but may be circumferential to a neural structure such as those employed in cuff electrodes. Electrodes spaced around the circumference of a major nerve, for instance the vagal nerve could use the techniques described above to provide estimates as to the locations of individual fascicles within the nerve bundle. It is highly desirable to be able to address individual fascicles with stimulation and a knowledge of the fascicle geometry and arrangement could, via current steering or other means, provide selective stimulation.

[0098] The fascicles in major nerves do not run a linear course through the nerve. For example, examination of serial cross sections of the nerve at different positions along the nerve

would reveal that individual fascicles at the centre of the bundle in one section could be found at the edges in another section. This observation, combined with the herein described techniques to map the separation of electrode to active tissue, could be used to choose effective electrodes or be used to appropriately place a cuff electrode on a nerve during surgery.

[0099] It will be appreciated by persons skilled in the art that numerous variations and/or modifications may be made to the invention as shown in the specific embodiments without departing from the spirit or scope of the invention as broadly described. The present embodiments are, therefore, to be considered in all respects as illustrative and not limiting or restrictive.

CLAIMS:

1. A method of estimating a nerve-to-electrode distance, the method comprising:

applying from a stimulus electrode to a nerve at least one stimulus having defined stimulus parameters;

obtaining a plurality of neural measurements of at least one compound action potential evoked by the at least one stimulus;

processing the plurality of neural measurements in order to estimate an originating state of stimulation, the originating state of stimulation exhibiting at least one observable characteristic defined by a single fibre size; and

applying a single fibre model to the estimated originating state of stimulation and the stimulus parameters, in order to produce a measure of the nerve-to-electrode distance.

2. The method of claim 1 wherein the measure of the nerve-to electrode distance comprises an absolute measure of distance.

3. The method of claim 1 wherein the measure of the nerve-to electrode distance comprises a relative measure reflecting a change in distance from a previous time.

4. The method of any one of claims 1 to 4 comprising the further step of adjusting a therapeutic stimulus regime in response to an observed change in the nerve-to-electrode distance.

5. The method of any one of claims 1 to 4 wherein the single fibre model comprises a lookup table matching the observed characteristic and the stimulus parameters to a corresponding nerve-to-electrode distance.

6. The method of any one of claims 1 to 5 performed intra-operatively, as part of a surgical procedure.

7. The method of claim 6 performed to progressively monitor a position of a structure bearing the electrodes relative to the nerve.

8. The method of claim 6 or claim 7 performed to image fibres of the nerve.

9. The method of any one of claims 1 to 5 performed as part of a postoperative fitting procedure of a neurostimulator.

10. The method of any one of claims 1 to 9 wherein the originating state of stimulation comprises an estimate of the ECAP peak width at the stimulus site.

11. The method of claim 10, wherein the applying comprises applying a single stimulus in order to evoke a single ECAP, and wherein the plurality of neural measurements are obtained from at least two sense electrodes each at a unique distance away from the stimulus electrode.

12. The method of claim 10 or claim 11, further comprising estimating an originating ECAP peak width by extrapolating the first and second ECAP measures back to the stimulus site.

13. The method of claim 12, further comprising taking the originating ECAP peak width as being dominated by a single fibre of largest diameter, and determining the nerve-to-electrode distance from the relationship of the single fibre originating peak width to nerve-to-electrode distance.

14. The method of any one of claims 10 to 13 wherein the measure of ECAP peak width comprises a half-height peak width, being a measure of a width of an ECAP peak as observed at an amplitude which is half the amplitude of the peak amplitude of the observed ECAP peak.

15. The method of any one of claims 10 to 14 wherein the measure of ECAP peak width comprises a time between peaks of the observed response.

16. The method of any one of claims 10 to 15 wherein the measure of ECAP peak width comprises a time between a first zero crossing and a second zero crossing of the neural measurement.

17. The method of any one of claims 1 to 16 wherein neural measurements are obtained of both orthodromic and antidromic ECAPs, to improve an estimate of the originating state of stimulation.

18. The method of any one of claims 1 to 17 wherein the originating state of stimulation comprises the Rheobase.

19. The method of claim 18 wherein a stimulus threshold is determined at at least two differing stimulus pulse widths, and wherein the Rheobase is calculated from the stimulus thresholds.

20. The method of claim 18 or claim 19 wherein the conduction velocity is measured and used to determine a fibre diameter recruited at threshold.

21. The method of claim 20 wherein a fitted relationships of the modelled single fibre diameter Rheobase to the electrode-to-nerve separation is used to determine the separation.

22. An implantable device for estimating a nerve-to-electrode distance, the device comprising:

at least one stimulus electrode and at least one sense electrode;

measurement circuitry for obtaining a neural measurement from the or each sense electrode; and

a processor configured to apply from the or each stimulus electrode to a nerve at least one stimulus having defined stimulus parameters, obtain from the measurement circuitry a plurality of neural measurements of at least one compound action potential evoked by the at least one stimulus, process the plurality of neural measurements in order to estimate an originating state of stimulation, the originating state of stimulation exhibiting at least one observable characteristic defined by a single fibre size; and apply a single fibre model to the estimated originating state of

stimulation and the stimulus parameters, in order to produce a measure of the nerve-to-electrode distance.

23. A method of estimating a distribution of fibres recruited by a stimulus, the method comprising

obtaining from at least two sense electrodes spaced apart along a neural pathway respective measurements of a compound action potential propagating along the neural pathway;

determining a conduction velocity of the compound action potential from the latency between the measurements, and determining from the conduction velocity a dominant recruited fibre diameter;

determining a rate of dispersion of the compound action potential between the sense electrodes, and determining from the rate of dispersion a distribution of diameters of the recruited fibre population.

24. A device for estimating a distribution of fibres recruited by a stimulus, the device comprising

at least one stimulus electrode and at least two sense electrodes, configured to be spaced apart along a neural pathway;

measurement circuitry for obtaining a neural measurement from each sense electrode; and

a processor configured to obtain from the at least two sense electrodes respective measurements of a compound action potential propagating along the neural pathway, determine a conduction velocity of the compound action potential from the latency between the measurements, determine from the conduction velocity a dominant recruited fibre diameter, determine a rate of dispersion of the compound action potential between the sense electrodes, and determine from the rate of dispersion a distribution of diameters of the recruited fibre population.

25. A non-transitory computer readable medium for estimating a nerve-to-electrode distance, comprising instructions which, when executed by one or more processors, causes performance of the following:

applying from a stimulus electrode to a nerve at least one stimulus having defined stimulus parameters;

obtaining a plurality of neural measurements of at least one compound action potential evoked by the at least one stimulus;

processing the plurality of neural measurements in order to estimate an originating state of stimulation, the originating state of stimulation exhibiting at least one observable characteristic defined by a single fibre size; and

applying a single fibre model to the estimated originating state of stimulation and the stimulus parameters, in order to produce a measure of the nerve-to-electrode distance.

26. A non-transitory computer readable medium for estimating a distribution of fibres recruited by a stimulus, comprising instructions which, when executed by one or more processors, causes performance of the following:

obtaining from at least two sense electrodes spaced apart along a neural pathway respective measurements of a compound action potential propagating along the neural pathway;

determining a conduction velocity of the compound action potential from the latency between the measurements, and determining from the conduction velocity a dominant recruited fibre diameter;

determining a rate of dispersion of the compound action potential between the sense electrodes, and determining from the rate of dispersion a distribution of diameters of the recruited fibre population.

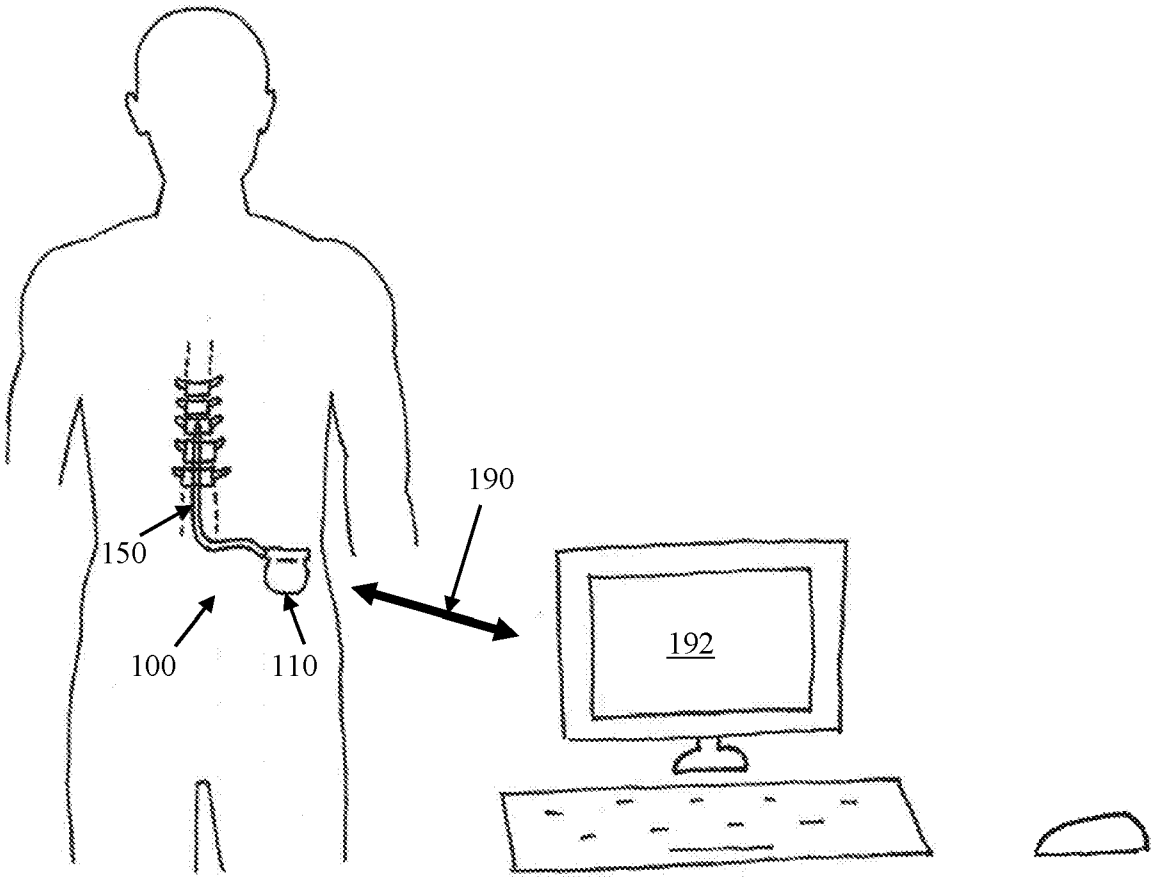

Figure 1

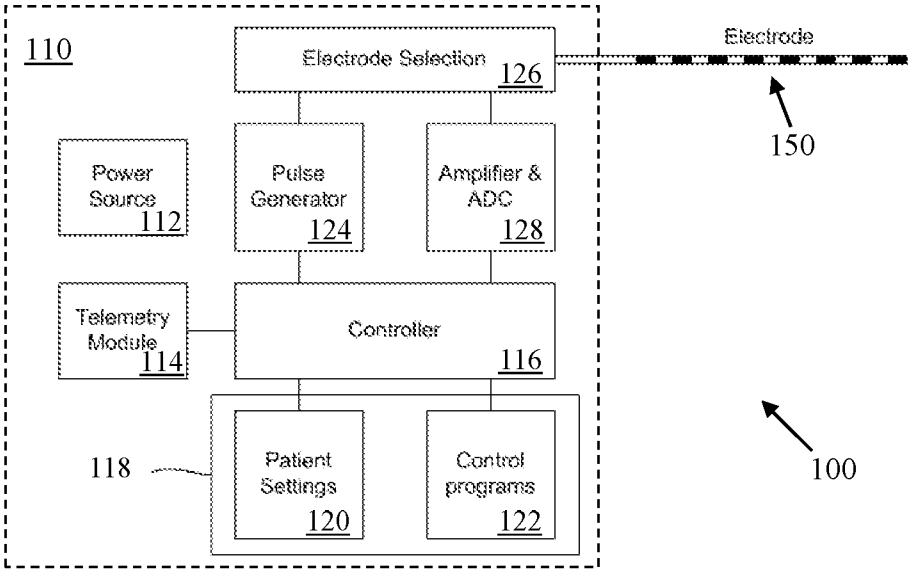

Figure 2

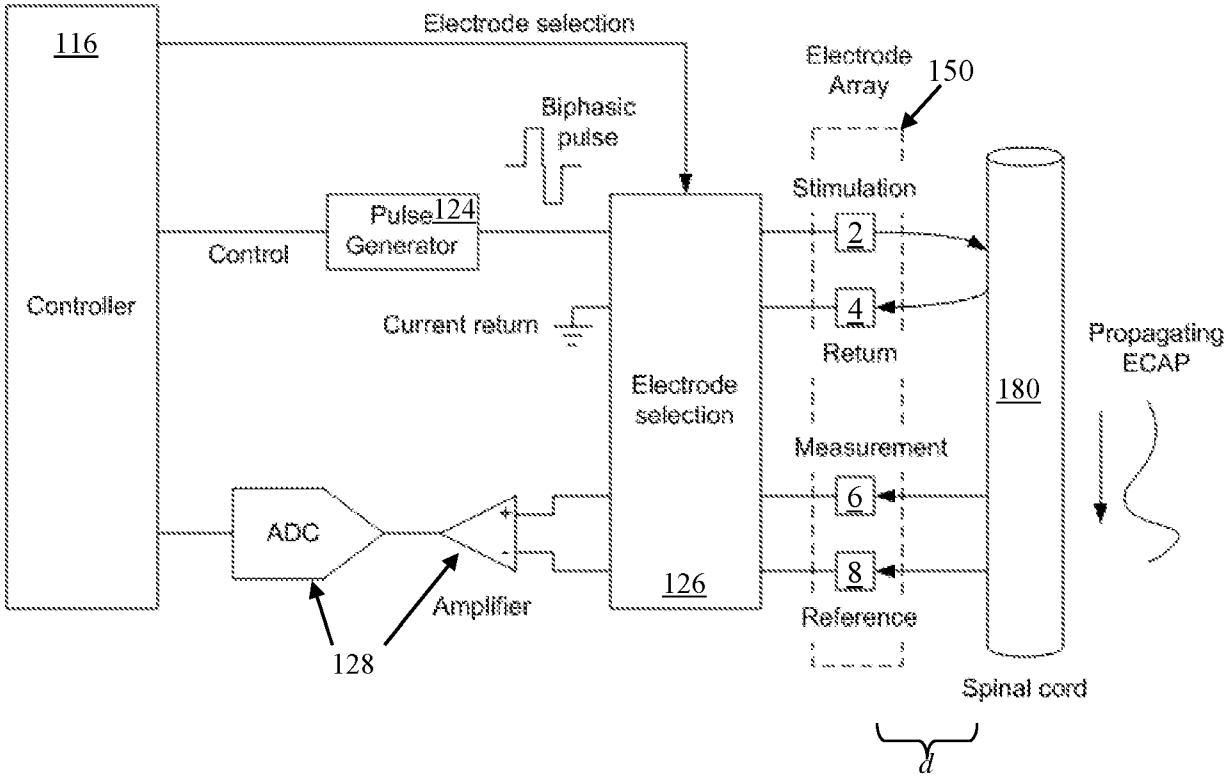

Figure 3

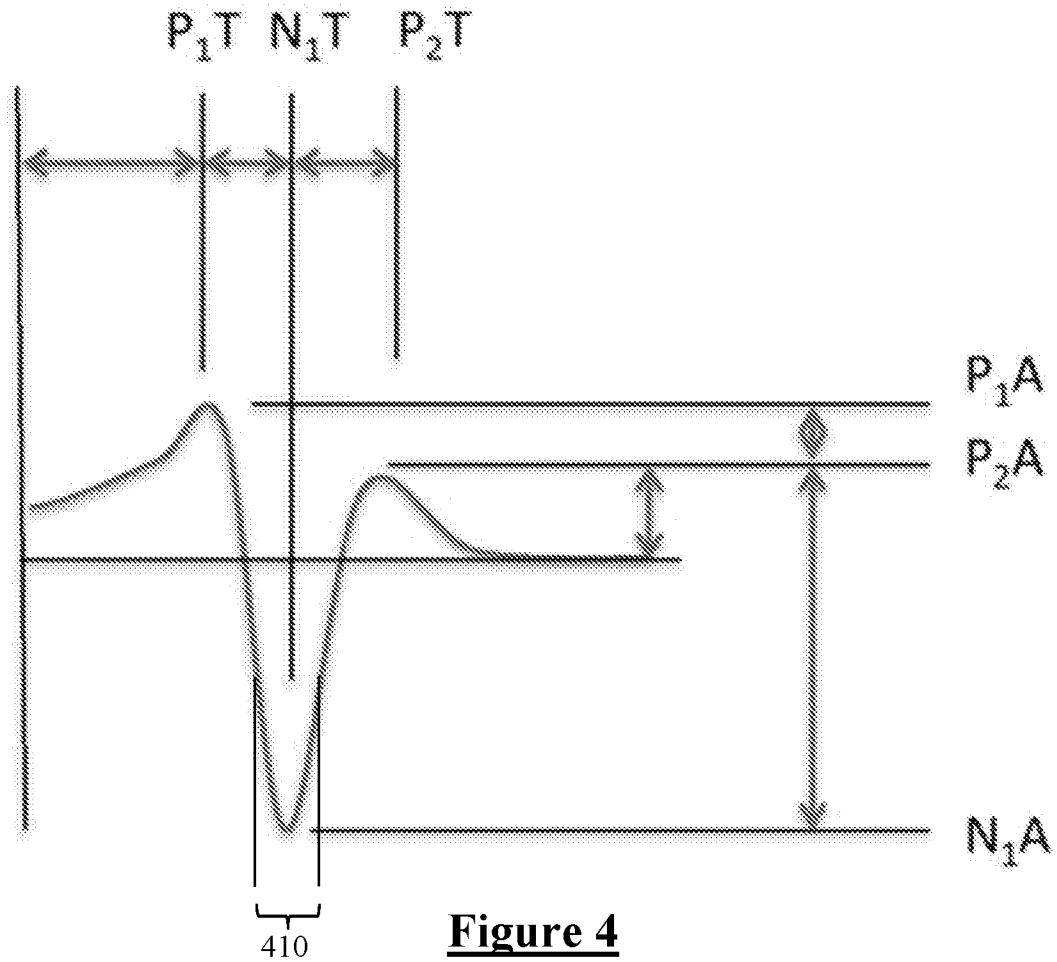

Figure 4

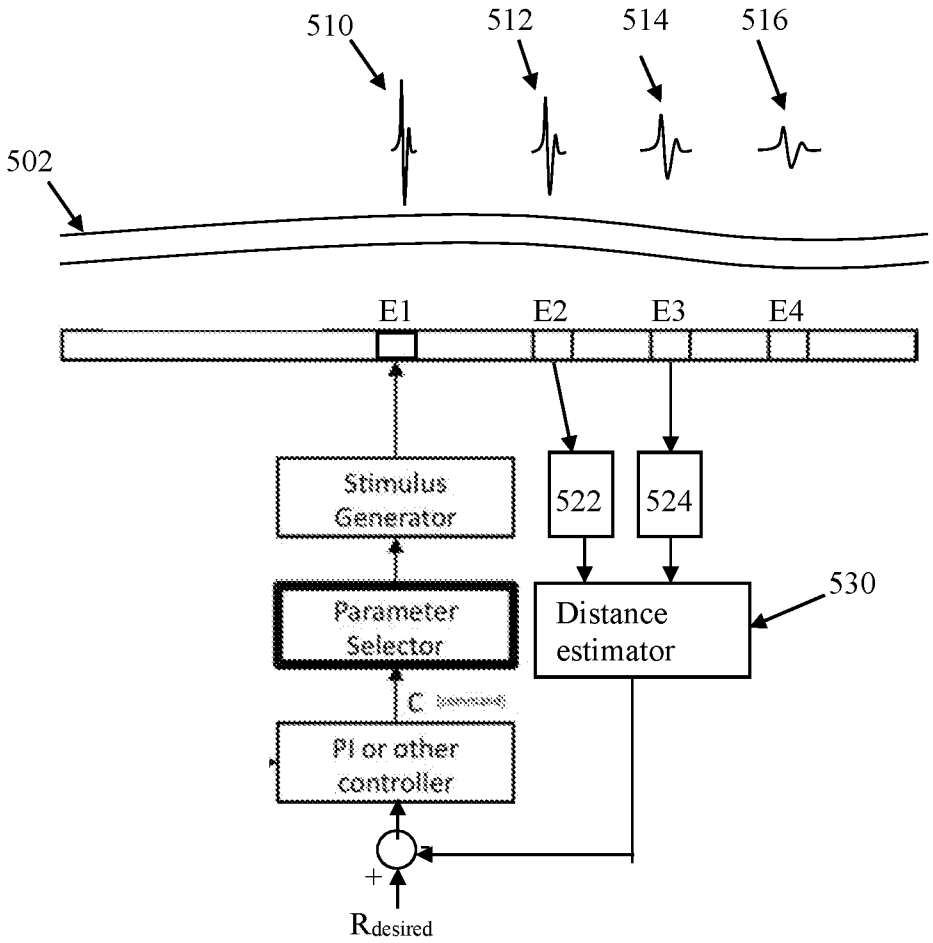

Figure 5

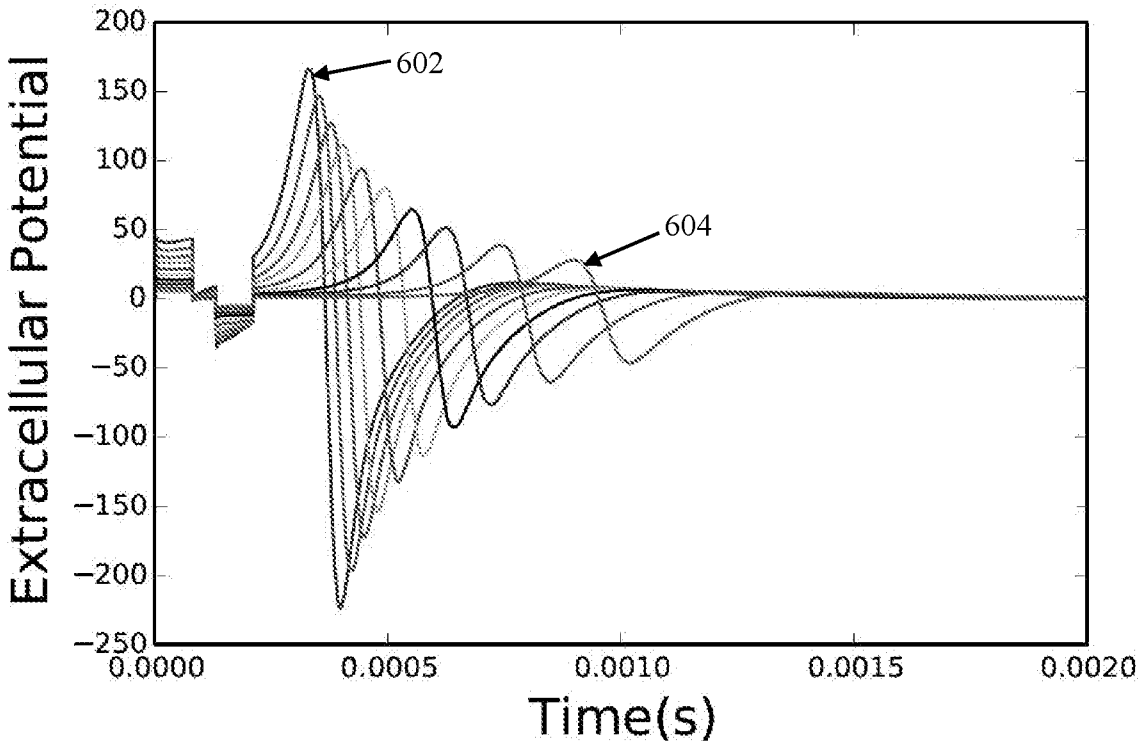

Figure 6

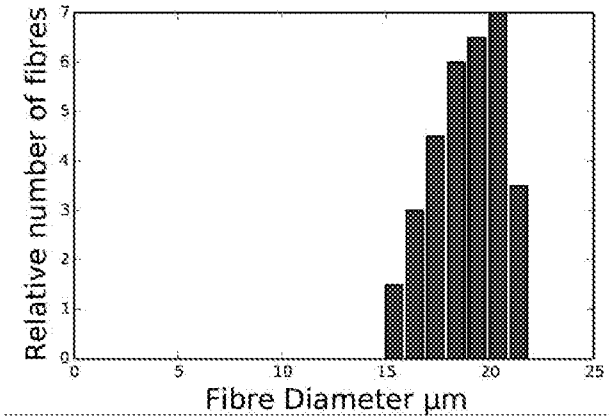

Figure 7

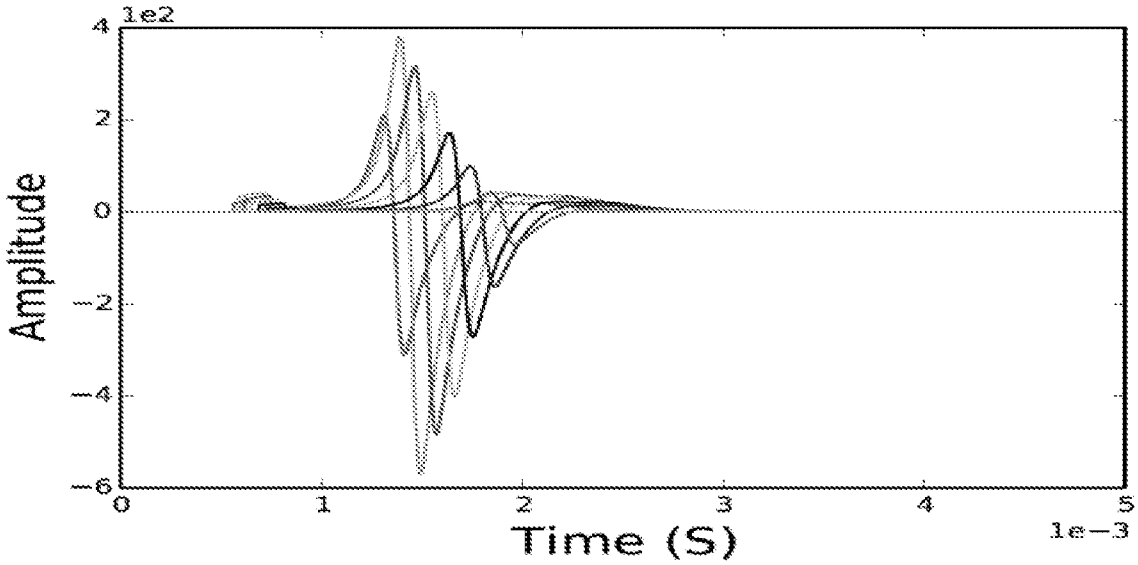

Figure 8

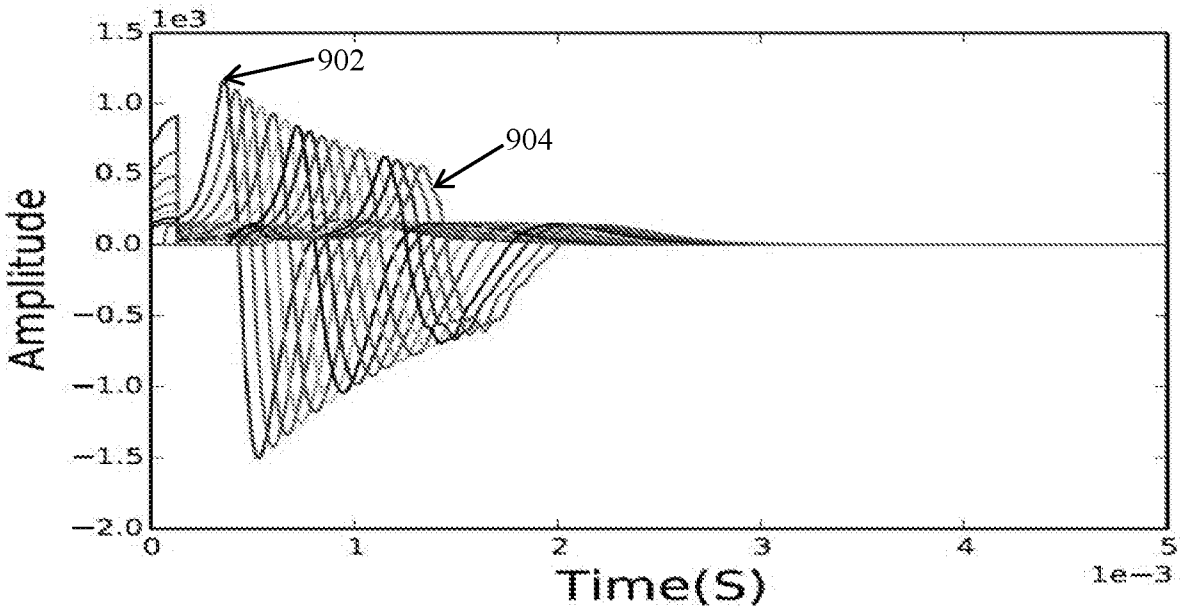

Figure 9

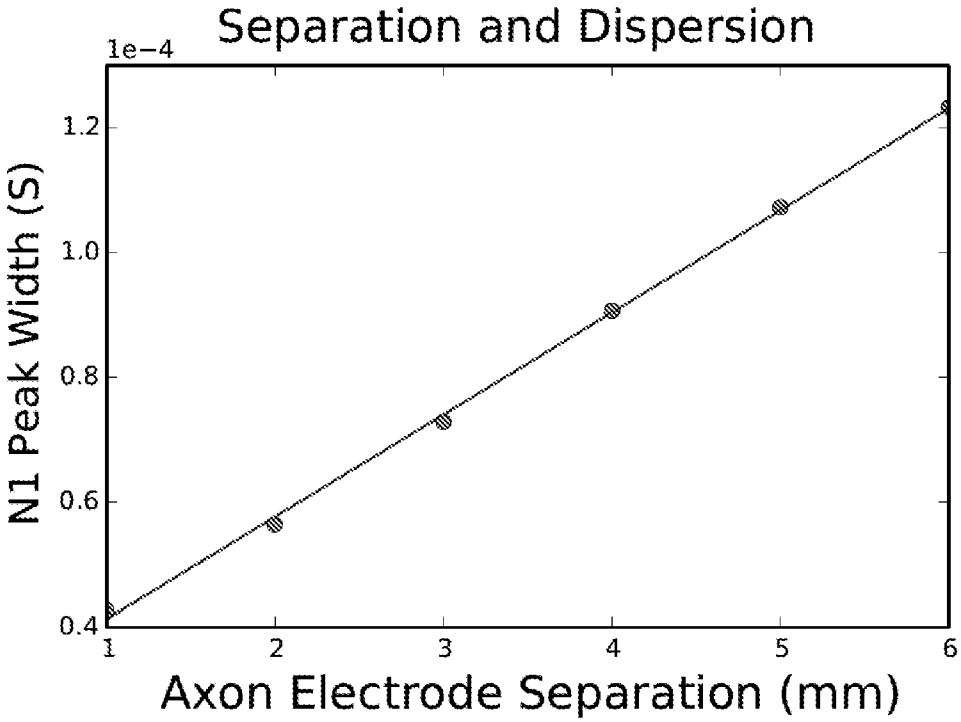

Figure 10

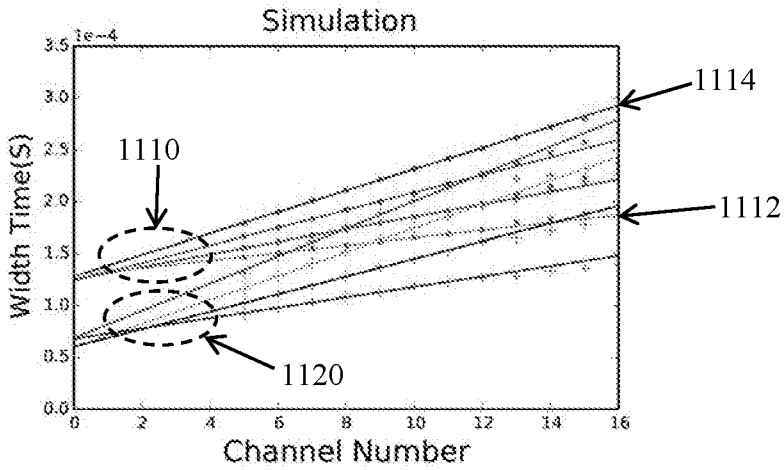

Figure 11

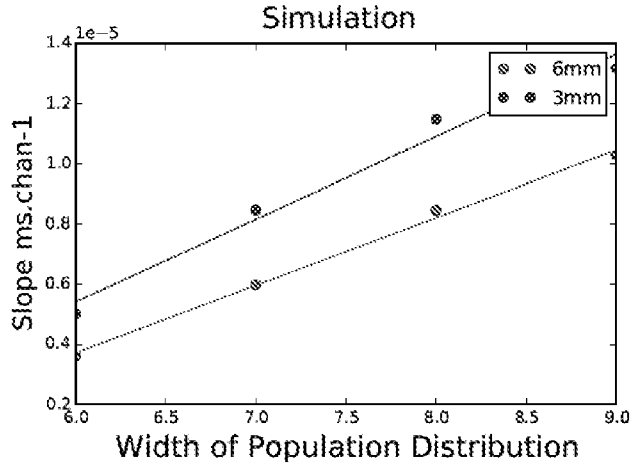

Figure 12

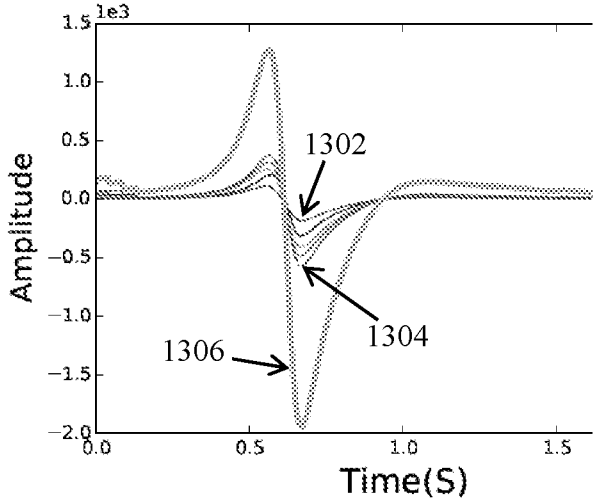

Figure 13

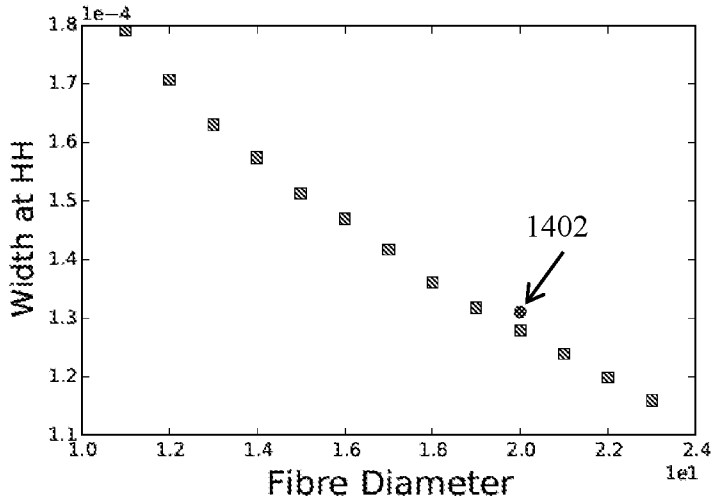

Figure 14

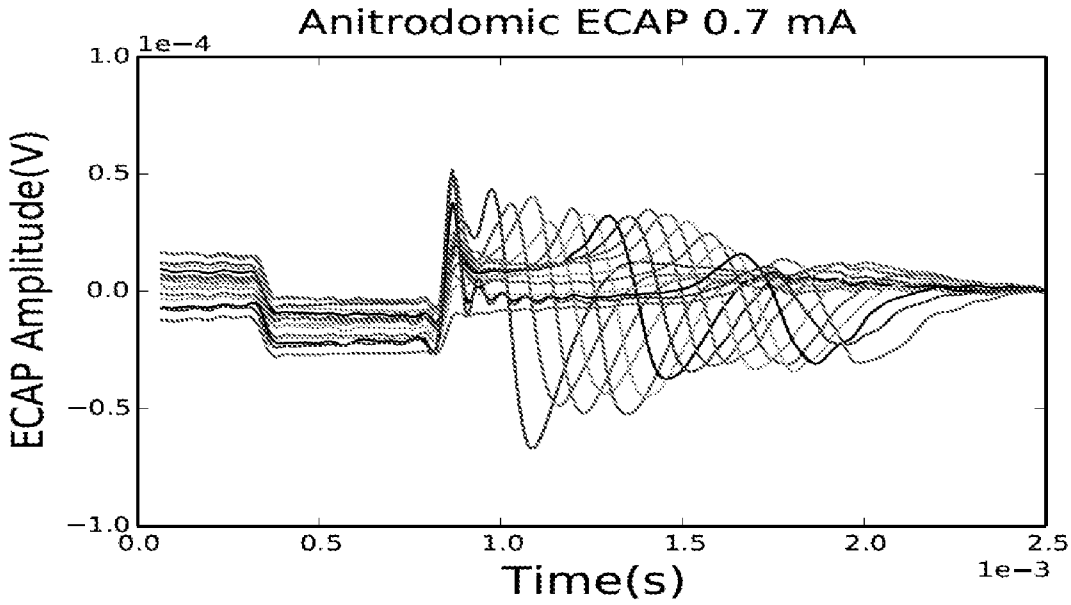

Figure 15a

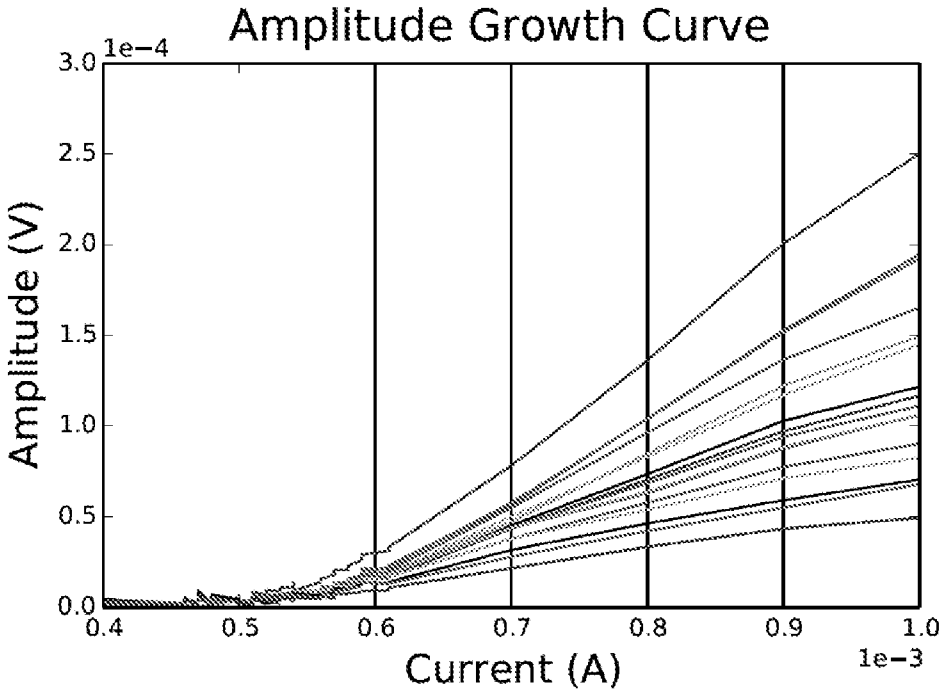

Figure 15b

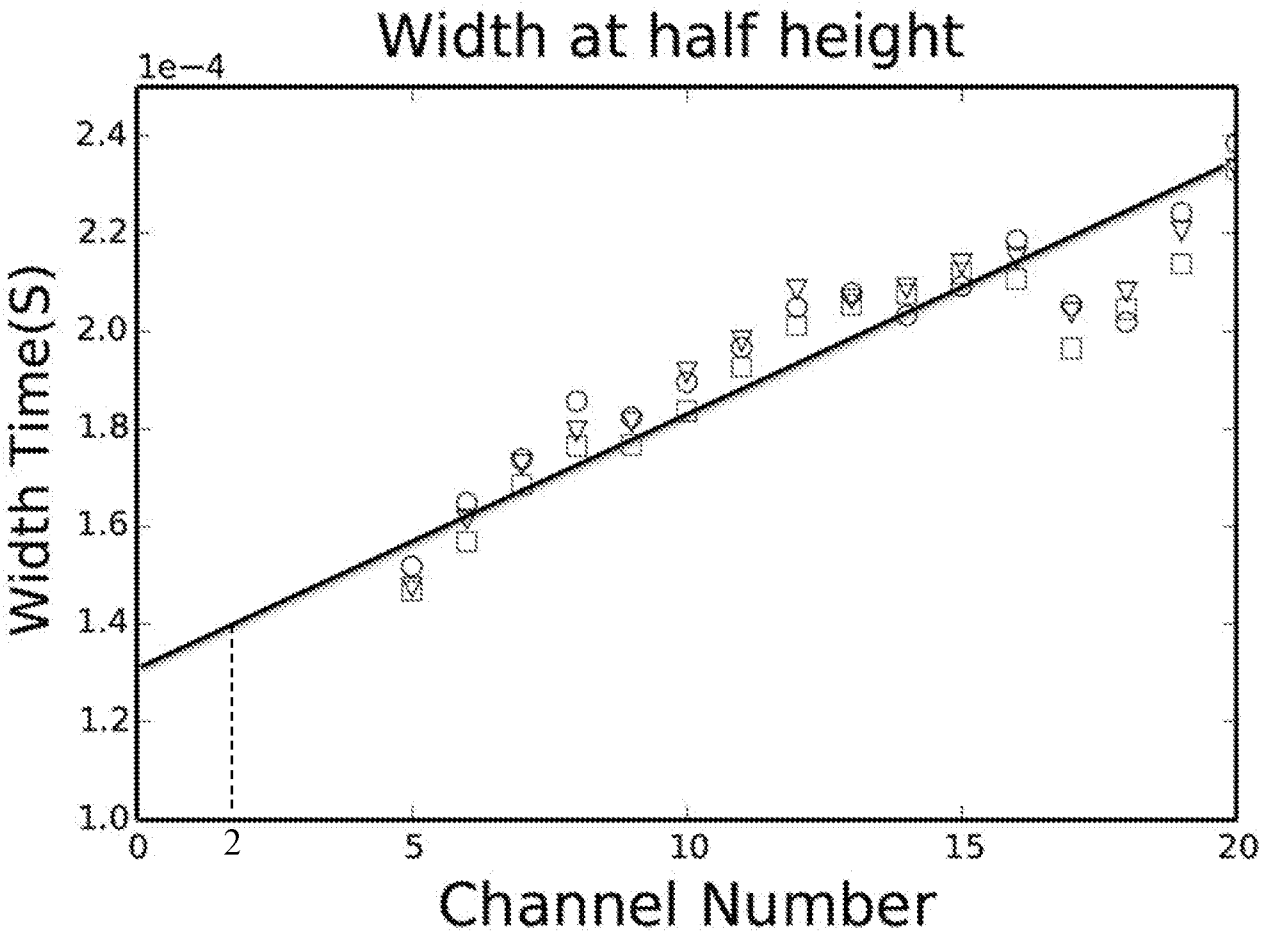

Figure 16a

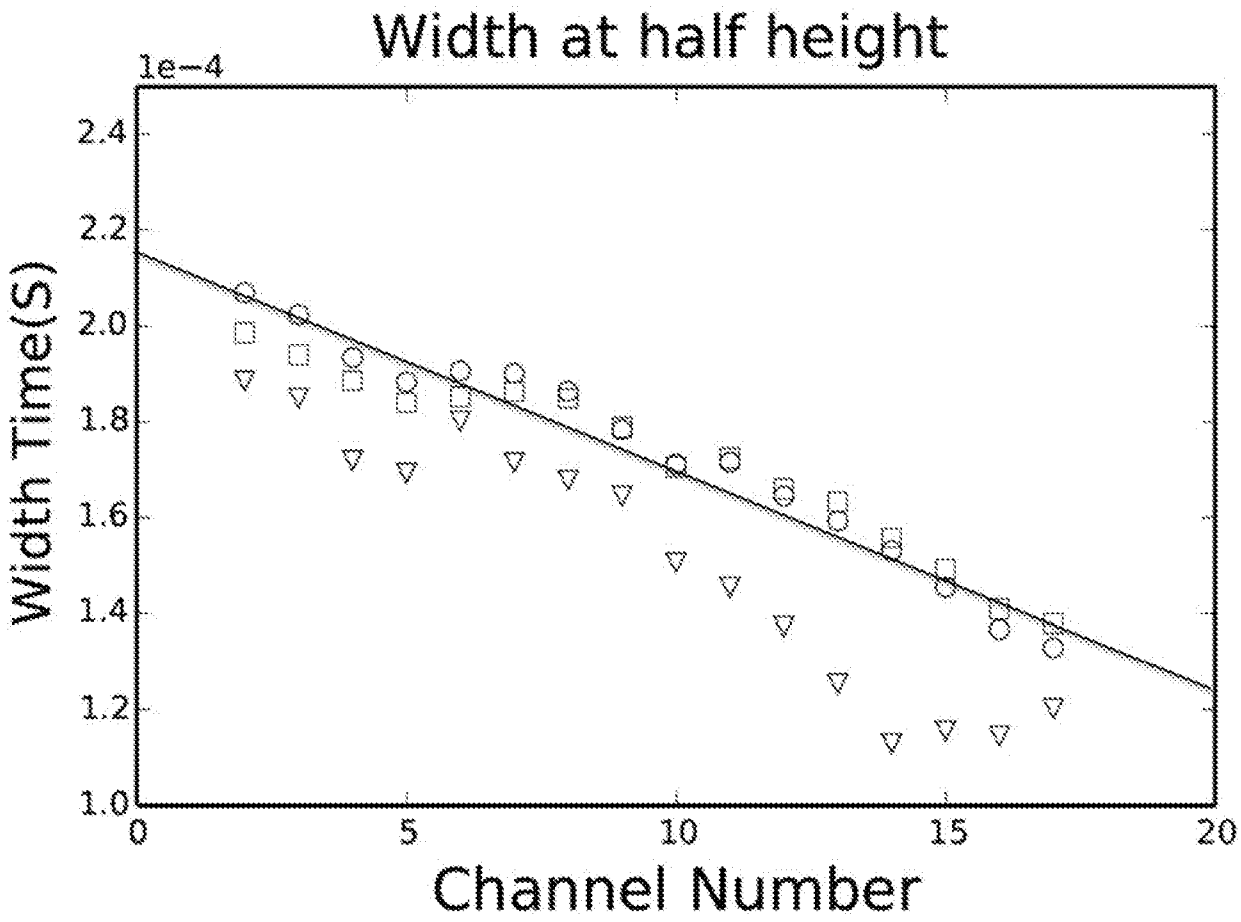

Figure 16b

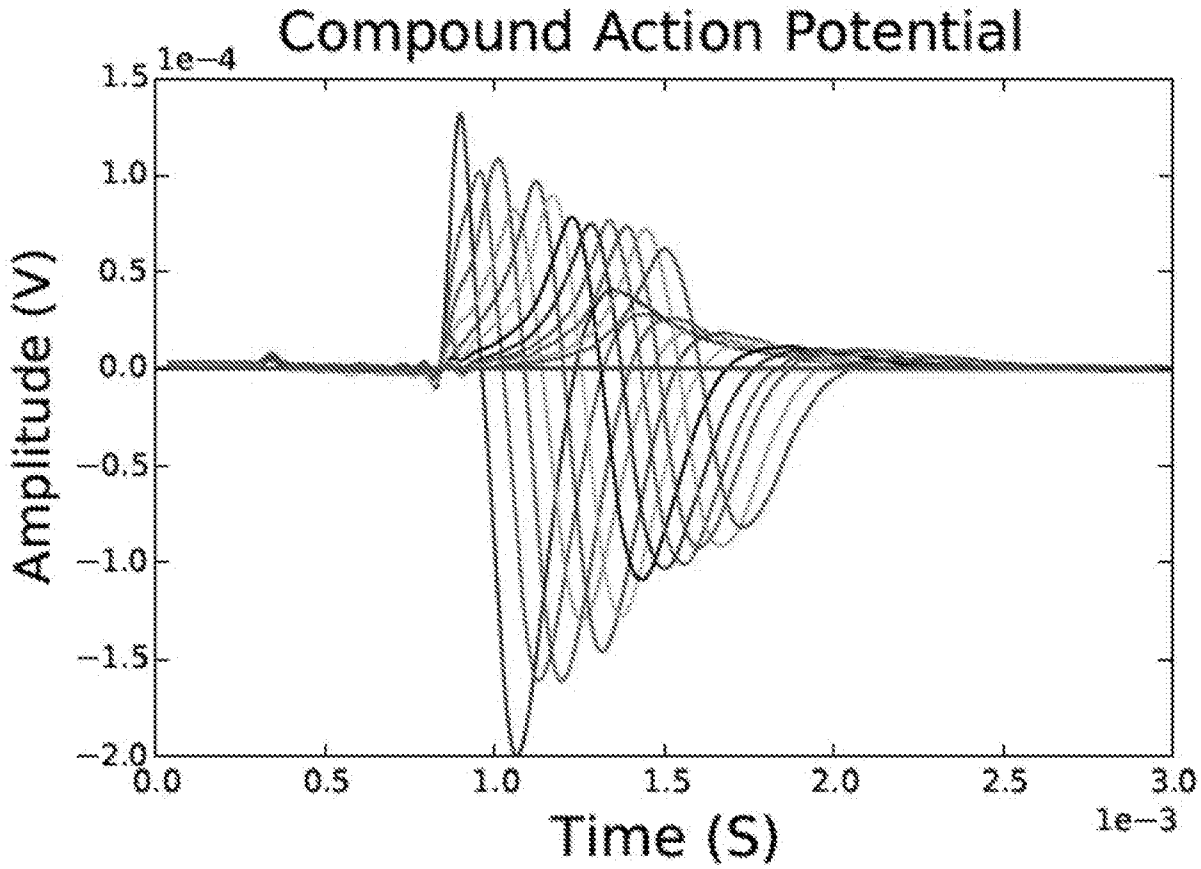

Figure 17

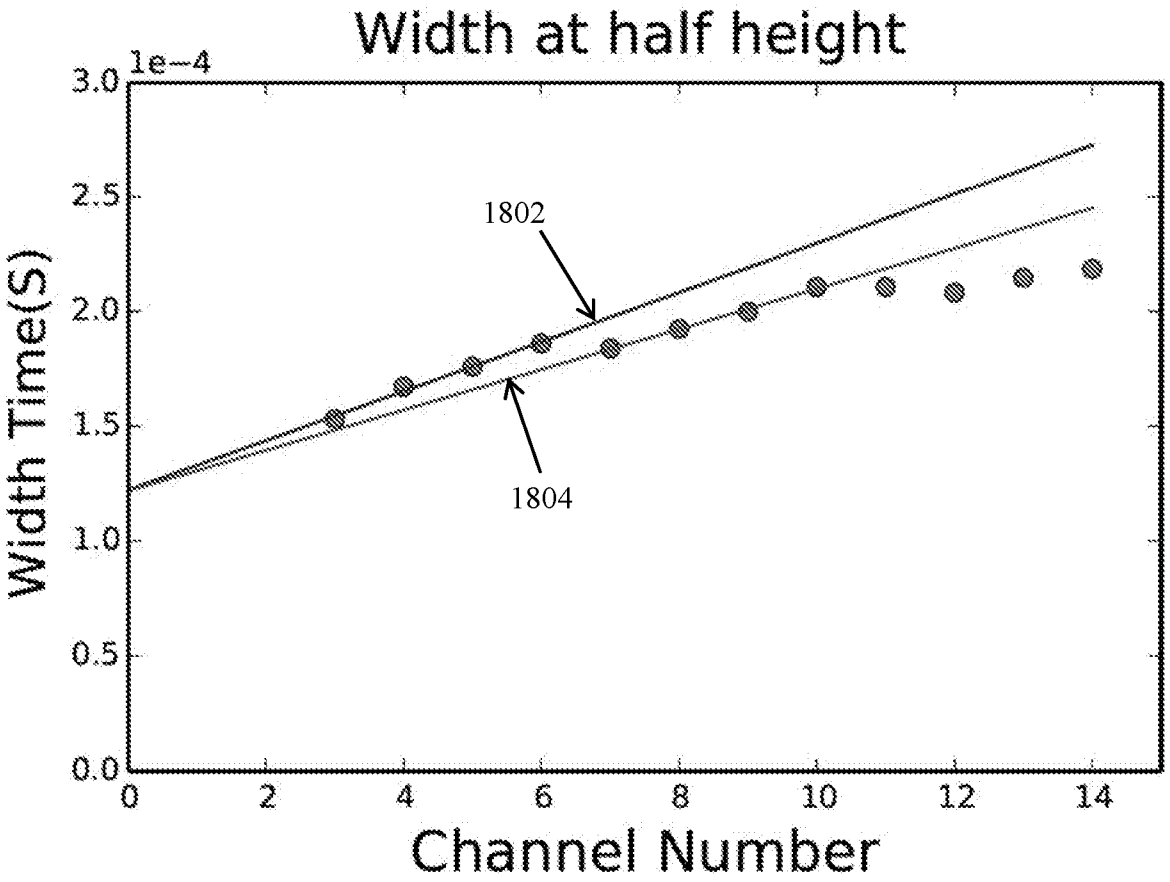

Figure 18

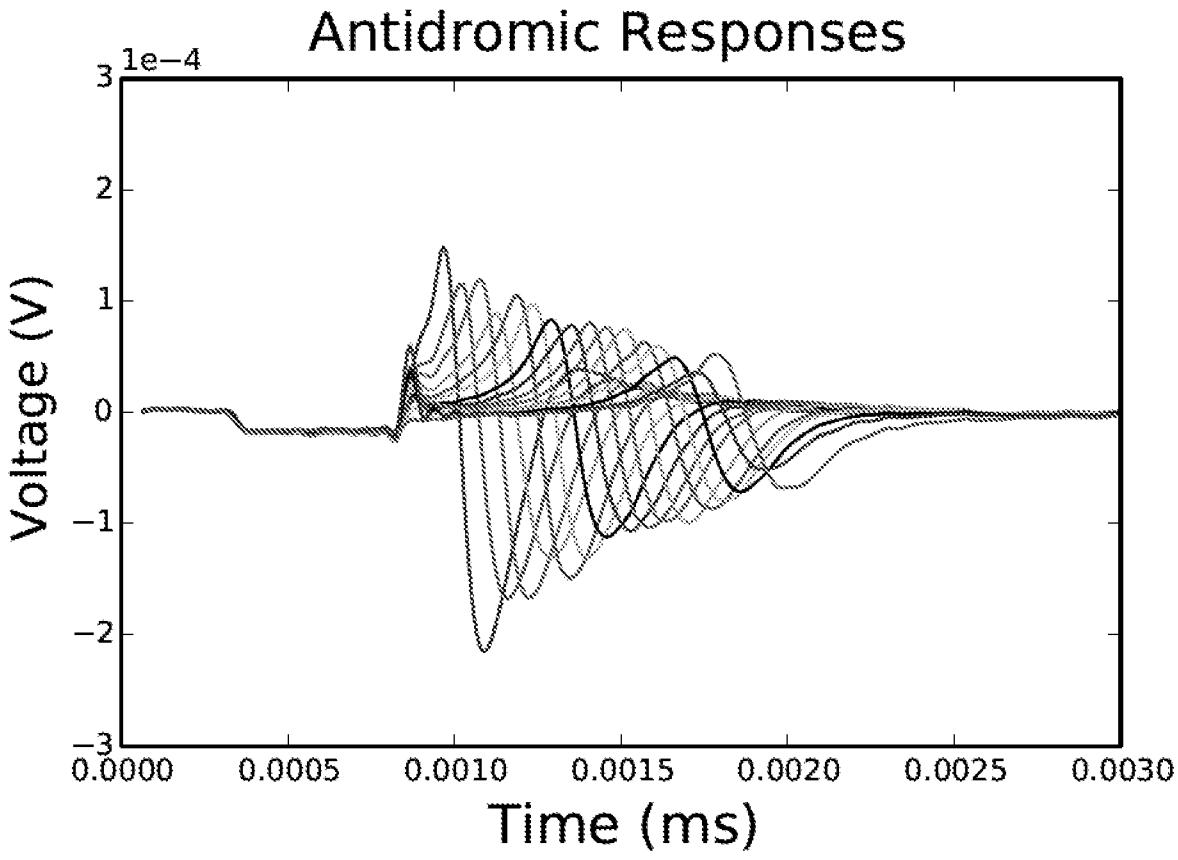

Figure 19a

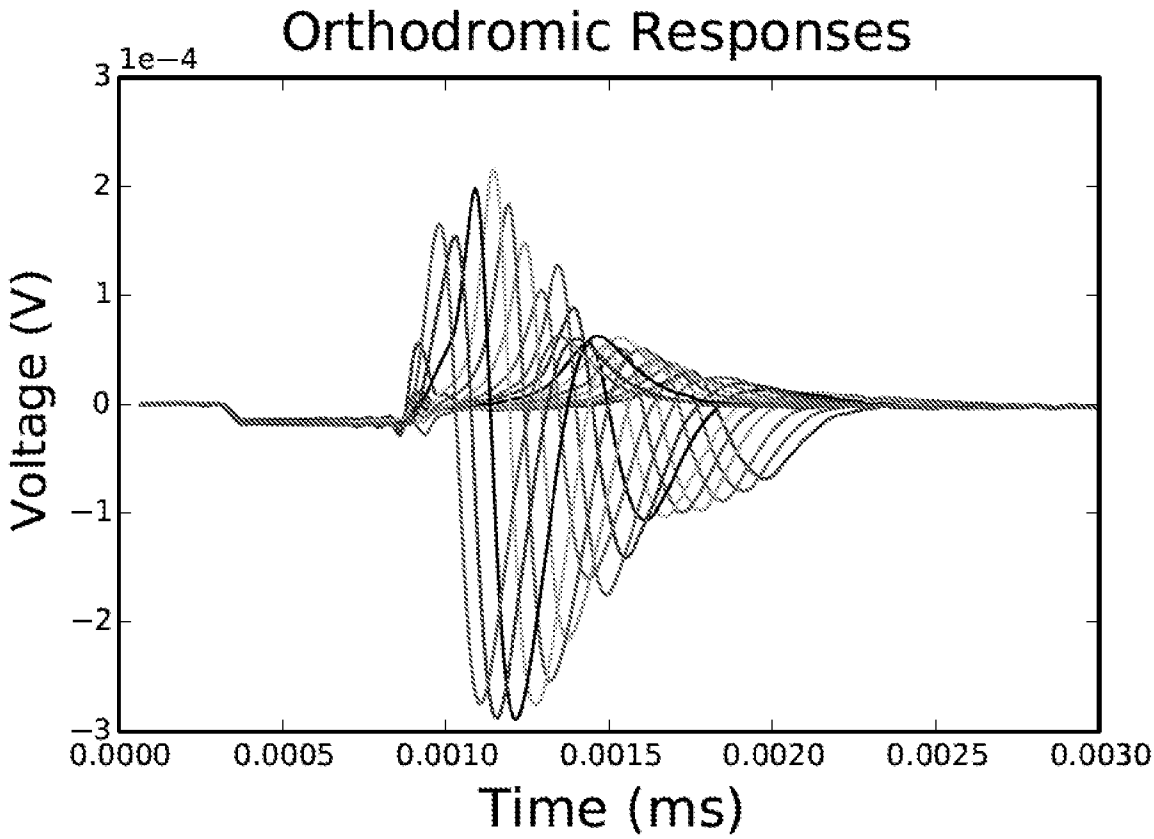
Orthodromic Responses
1e−4
3
2
1
0
−1
−2
−3
Voltage (V)
0.0000
0.0005
0.0010
0.0015
0.0020
0.0025
0.0030
Time (ms)

Figure 19b

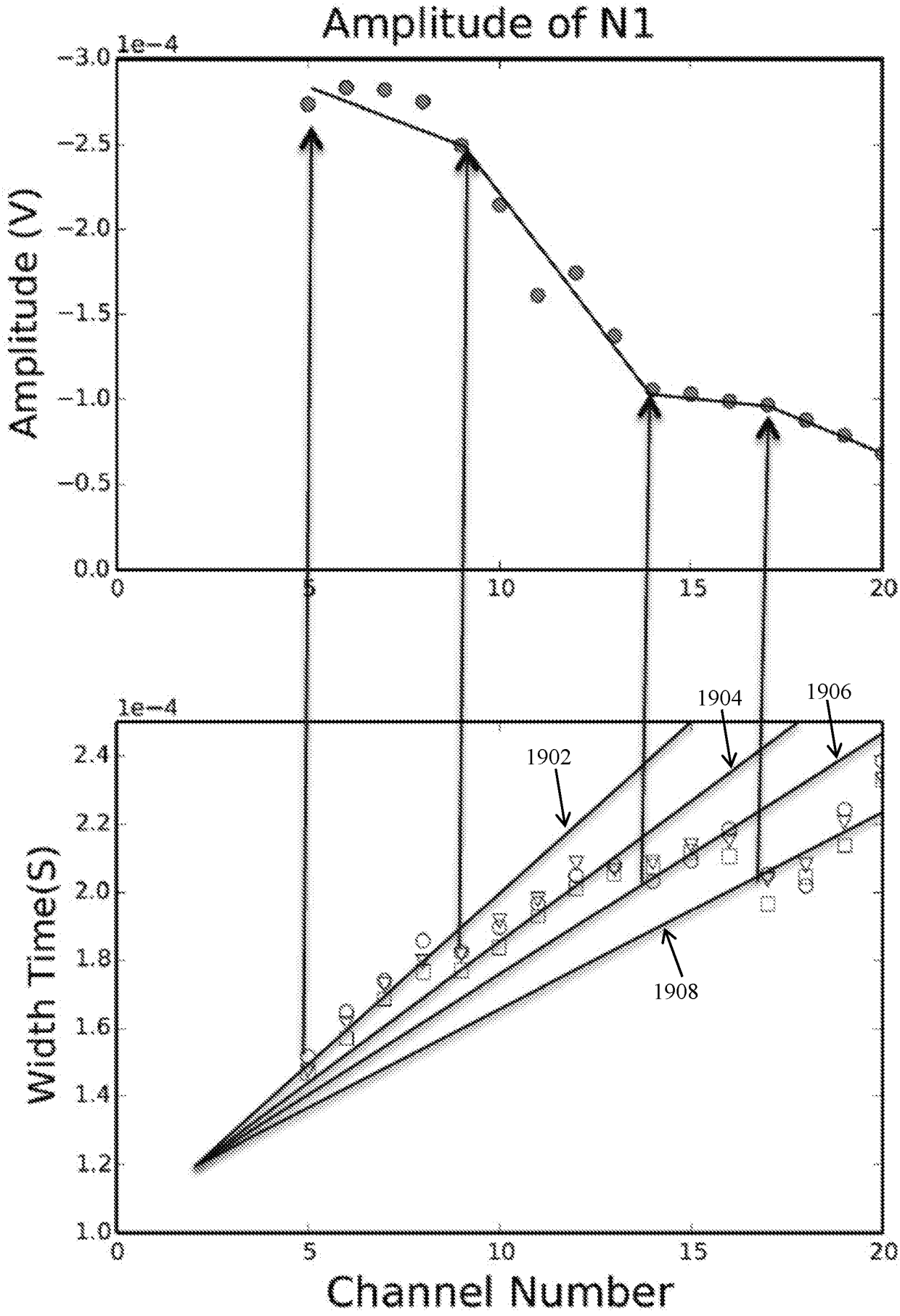
Amplitude of N1
1e−4
−3.0
−2.5
−2.0
−1.5
−1.0
−0.5
0.0
Amplitude (V)
0
5
10
15
20
1e−4
2.4
2.2
2.0
1.8
1.6
1.4
1.2
1.0
Width Time(S)
1902
1904
1906
1908
0
5
10
15
20
Channel Number

Figure 19c

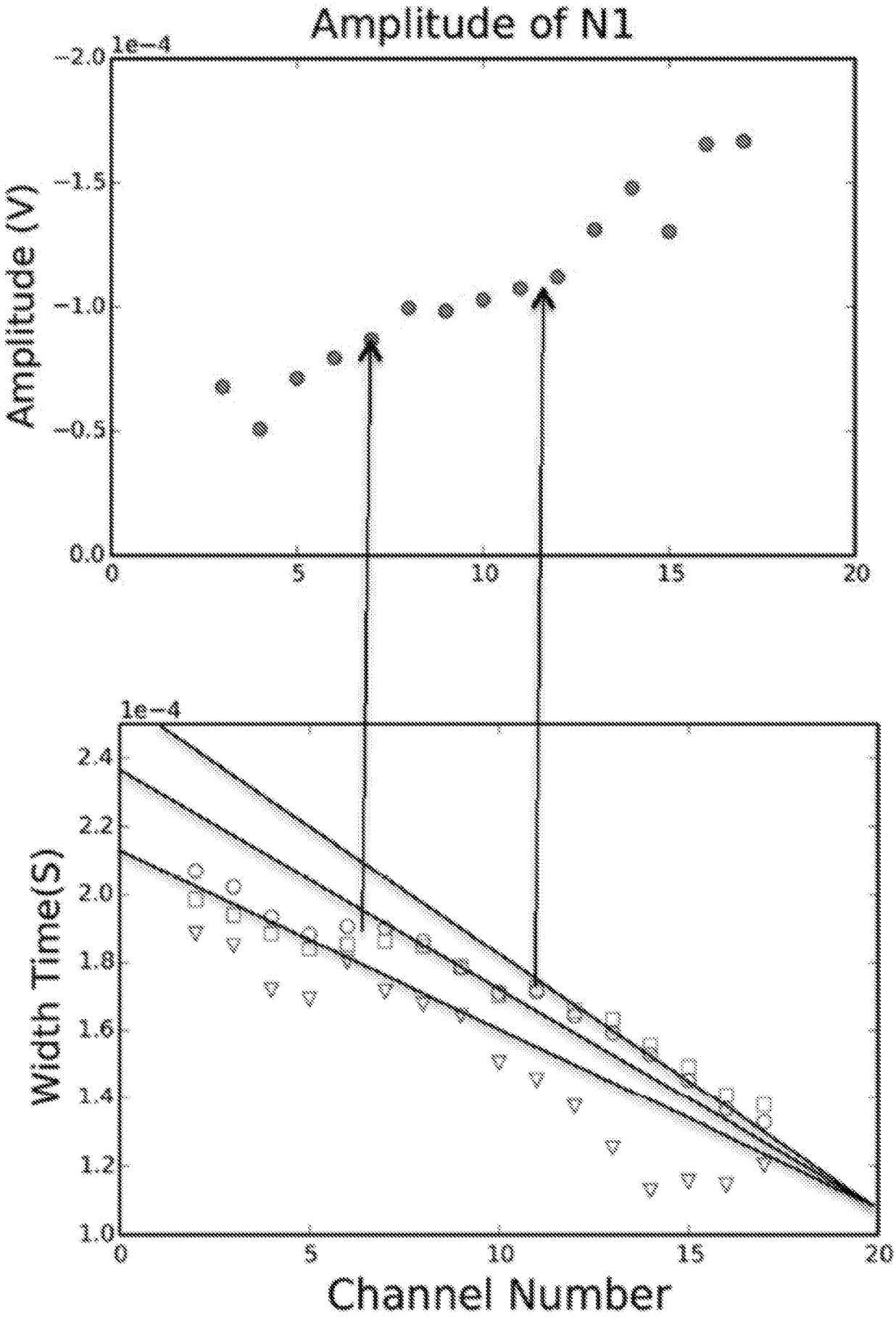
Amplitude of N1
1e−4
−2.0
−1.5
−1.0
−0.5
0.0
Amplitude (V)
0
5
10
15
20
1e−4
2.4
2.2
2.0
1.8
1.6
1.4
1.2
1.0
Width Time(S)
0
5
10
15
20
Channel Number

Figure 19d

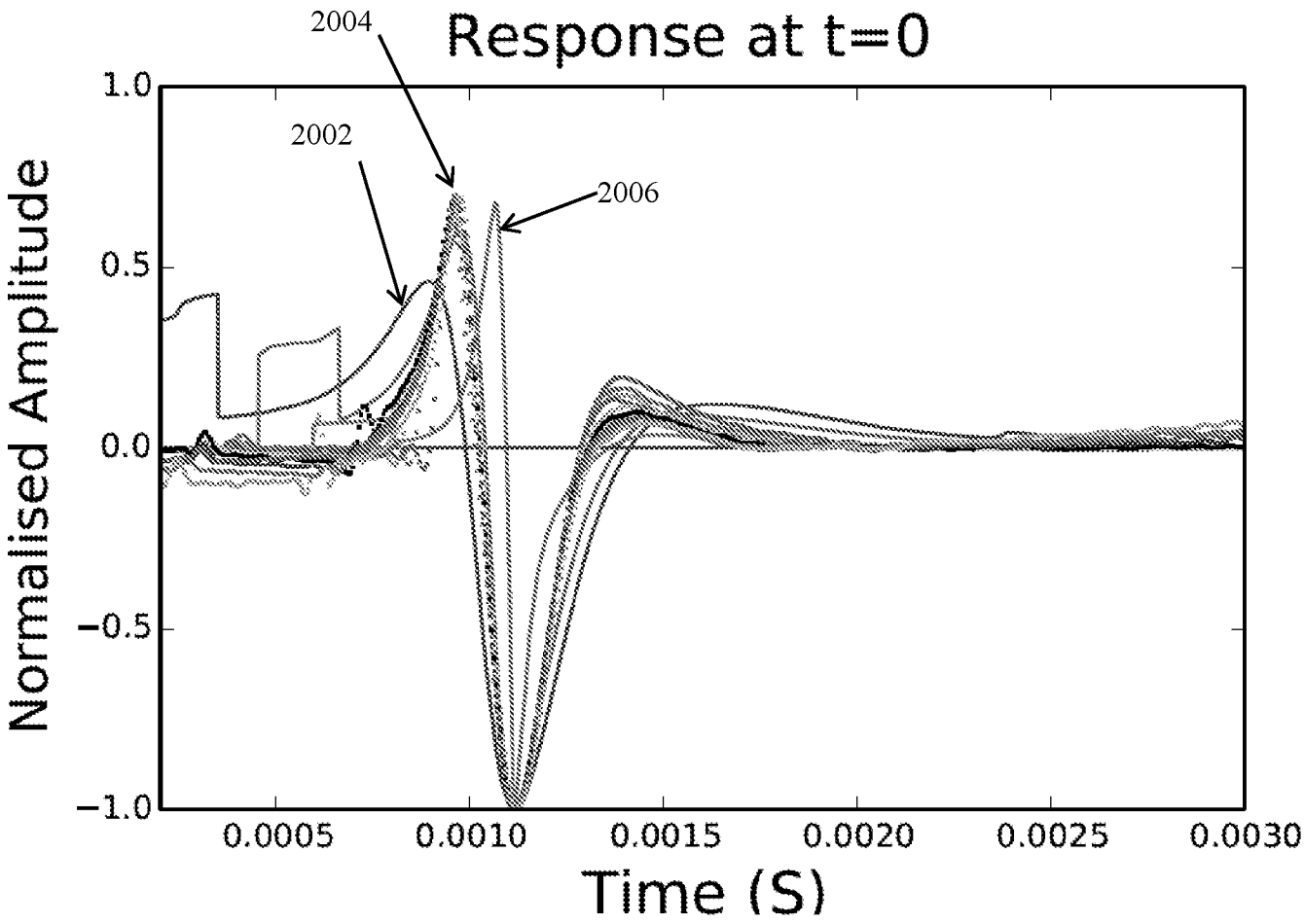

Figure 20

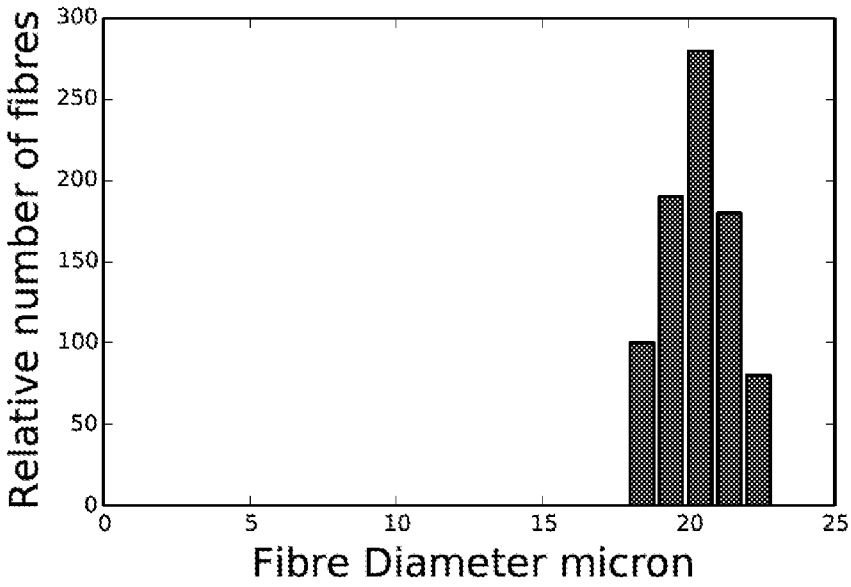

Figure 21

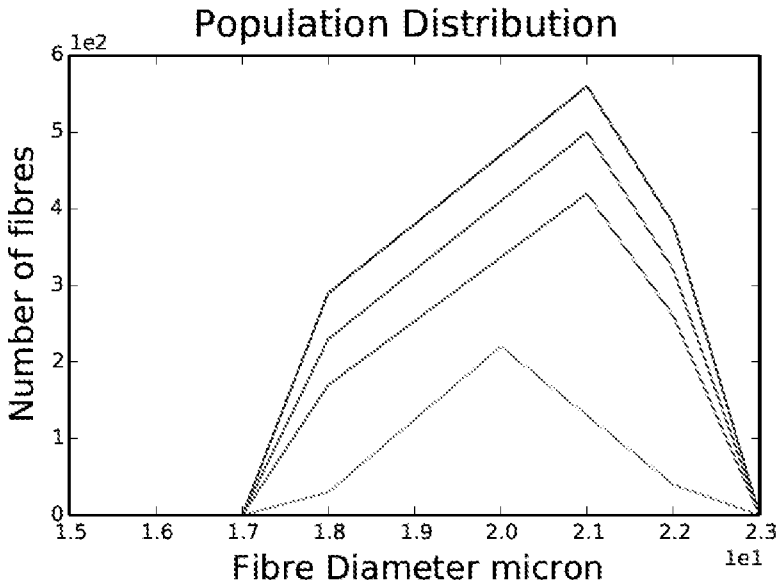
Population Distribution
1e2
6
5
4
3
2
1
0
Number of fibres
1.5
1.6
1.7
1.8
1.9
2.0
2.1
2.2
2.3
1e1
Fibre Diameter micron

Figure 23

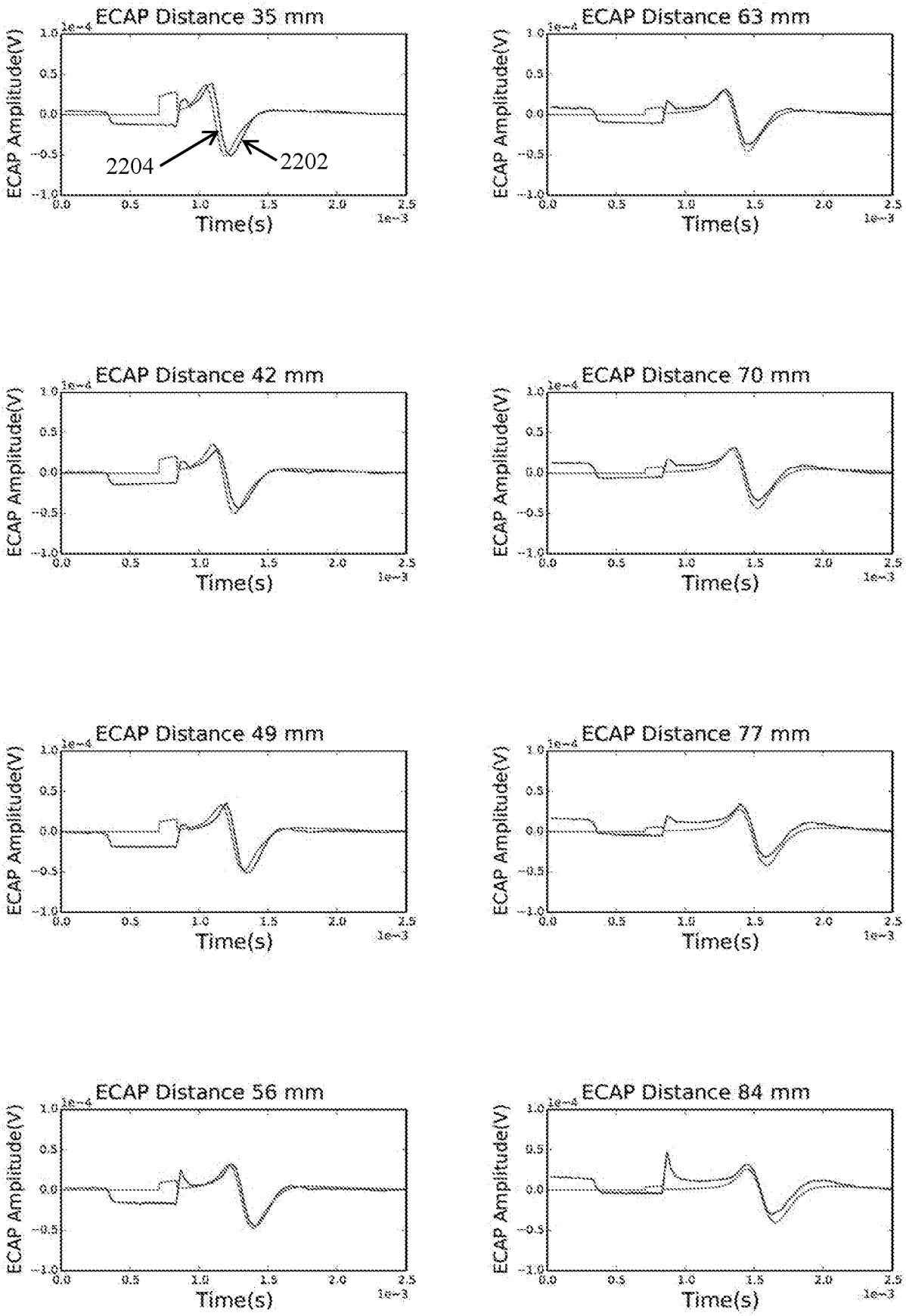
ECAP Distance 35 mm
ECAP Amplitude(V)
Time(s)
2204
2202
ECAP Distance 63 mm
ECAP Distance 42 mm
ECAP Distance 70 mm
ECAP Distance 49 mm
ECAP Distance 77 mm
ECAP Distance 56 mm
ECAP Distance 84 mm

Figure 22

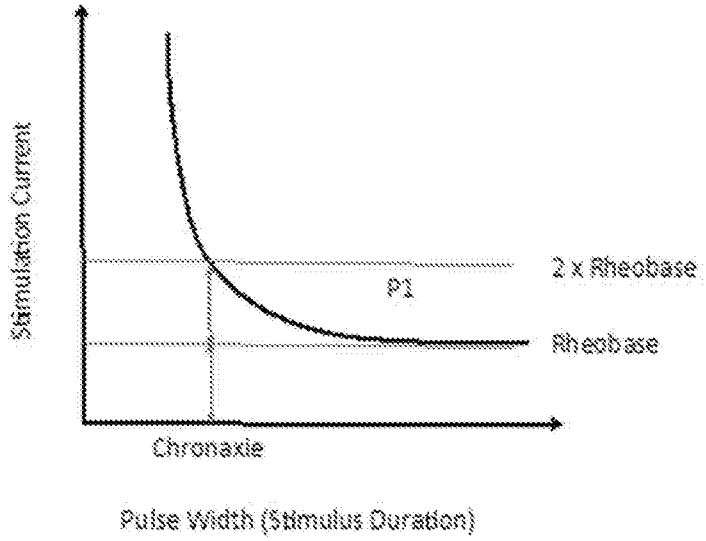

Figure 24

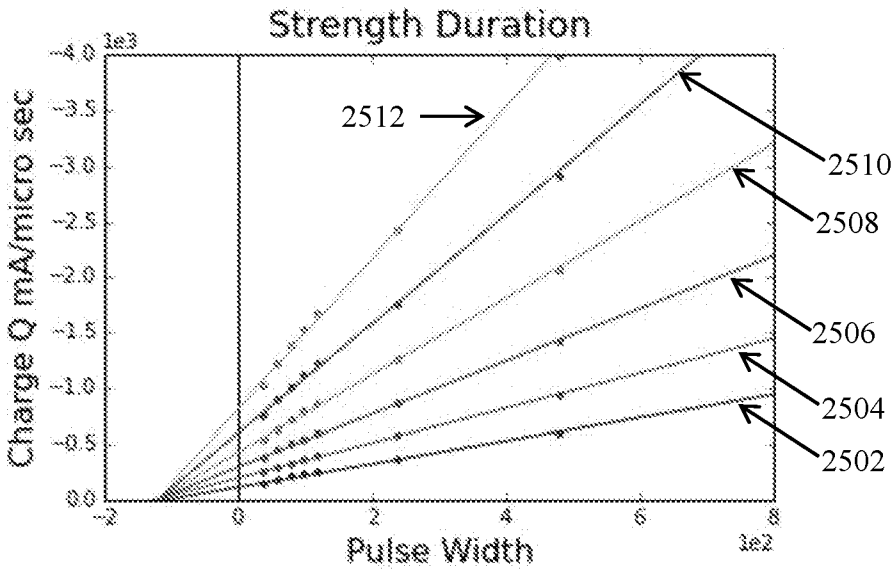

Figure 25

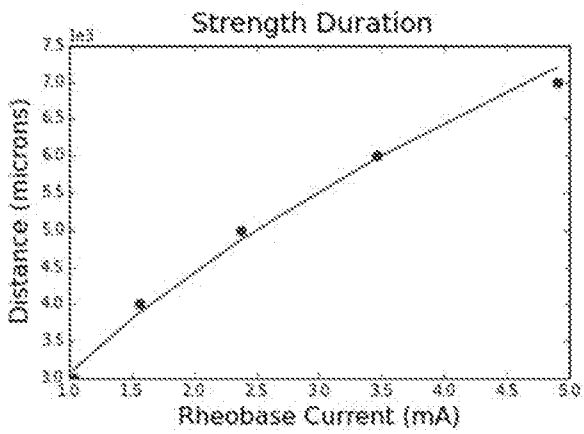

Figure 26

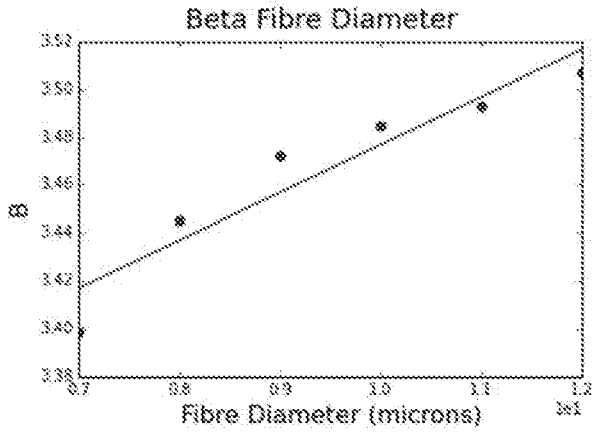

Figure 27

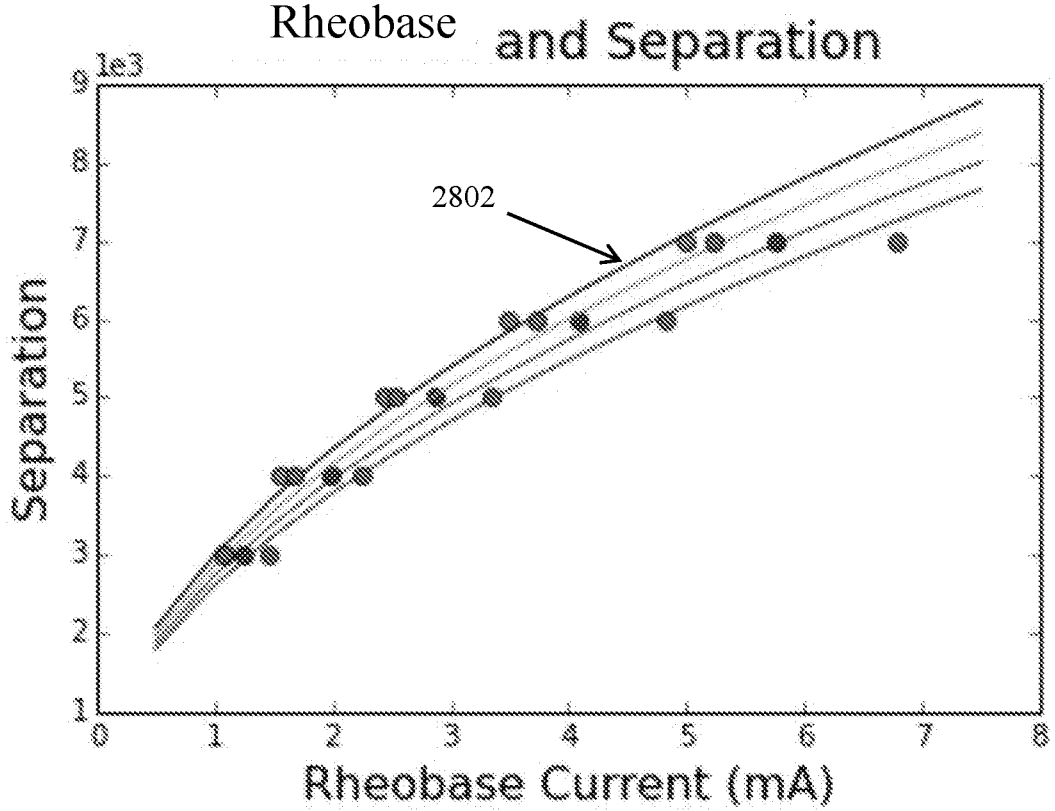

Figure 28

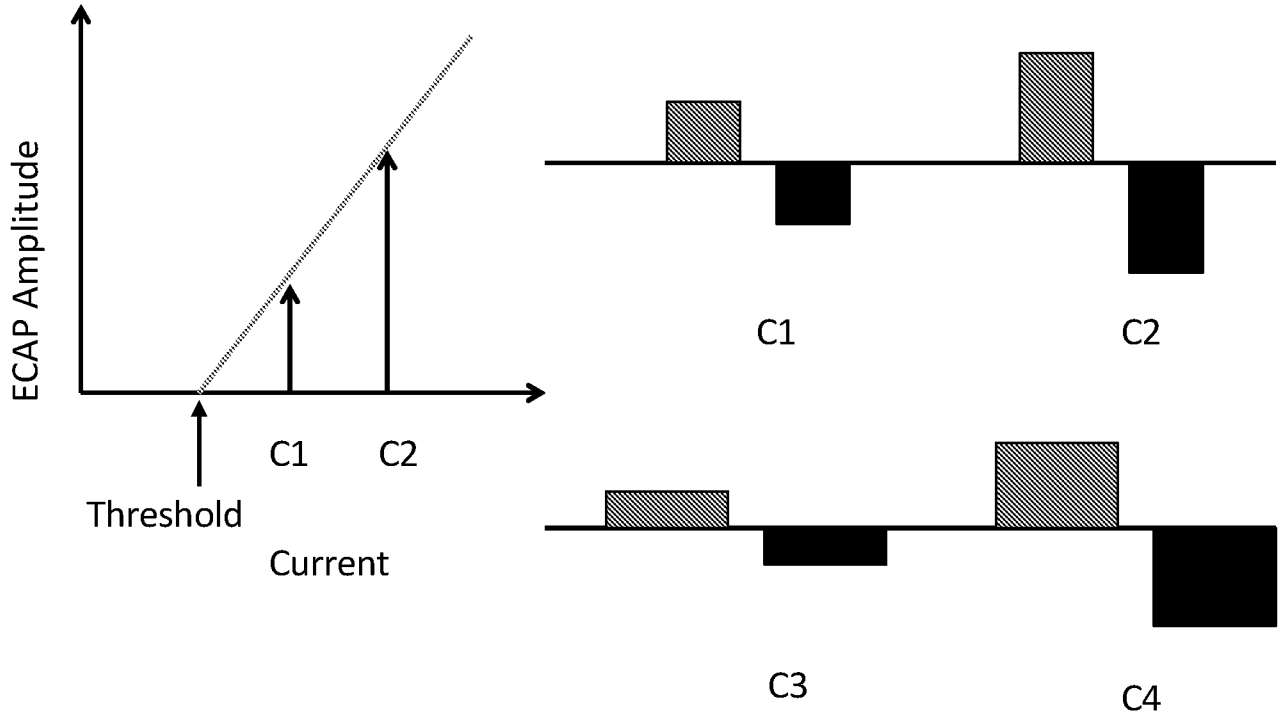

Figure 29

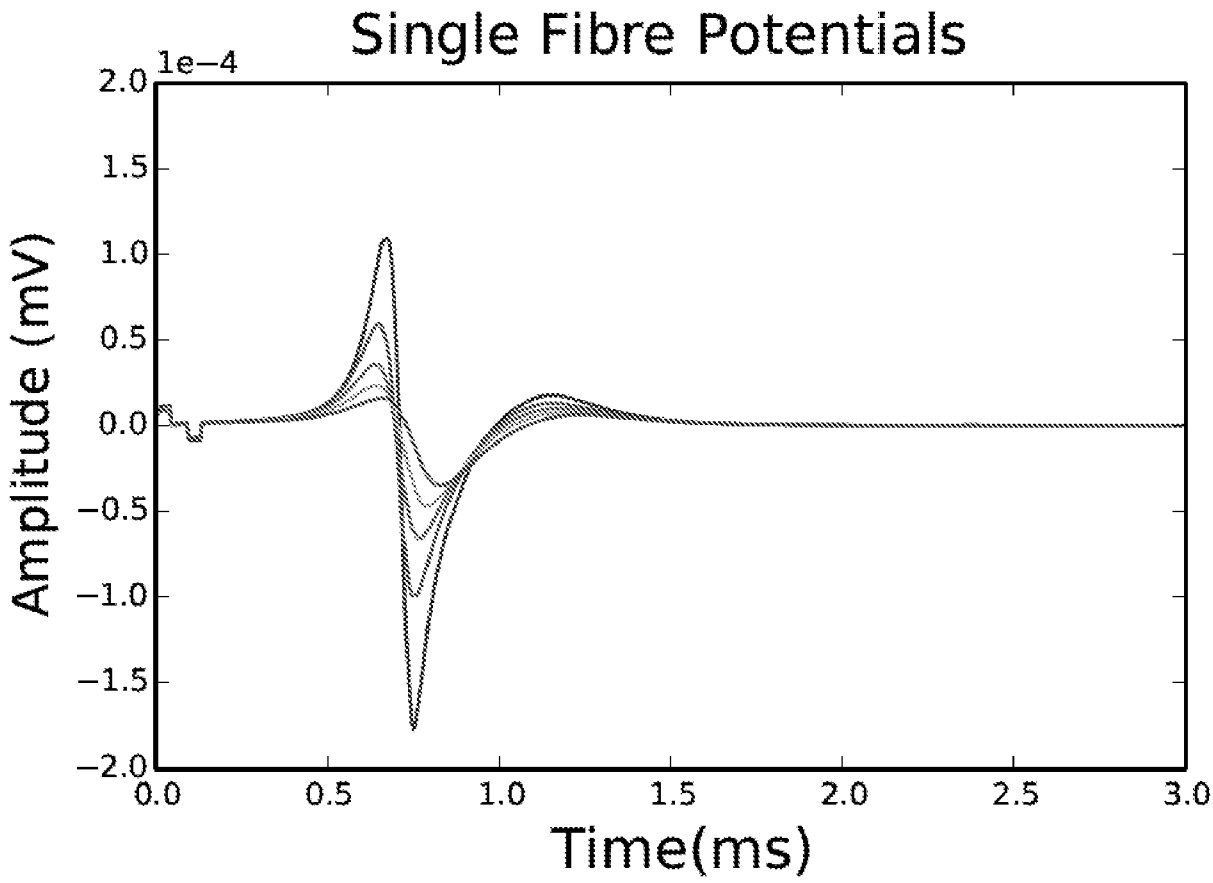

Figure 30a

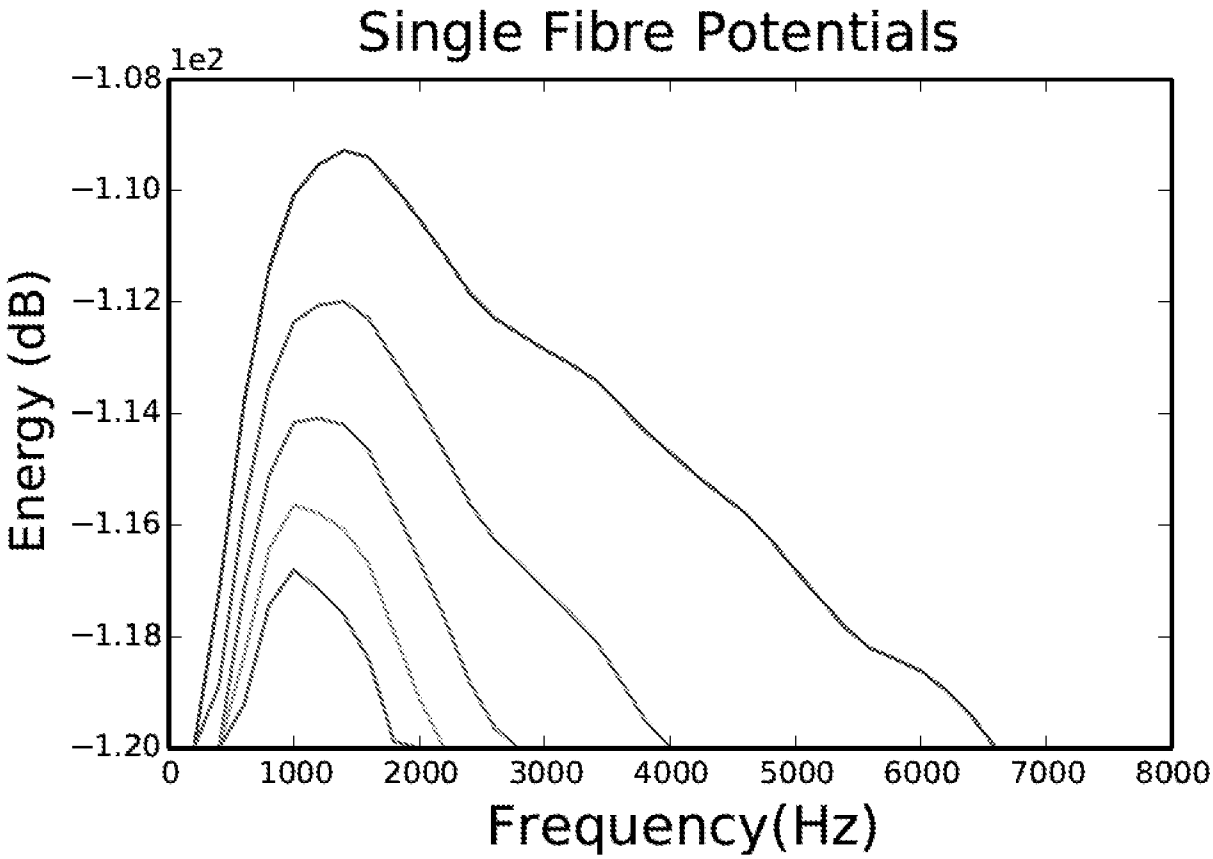

Figure 30b

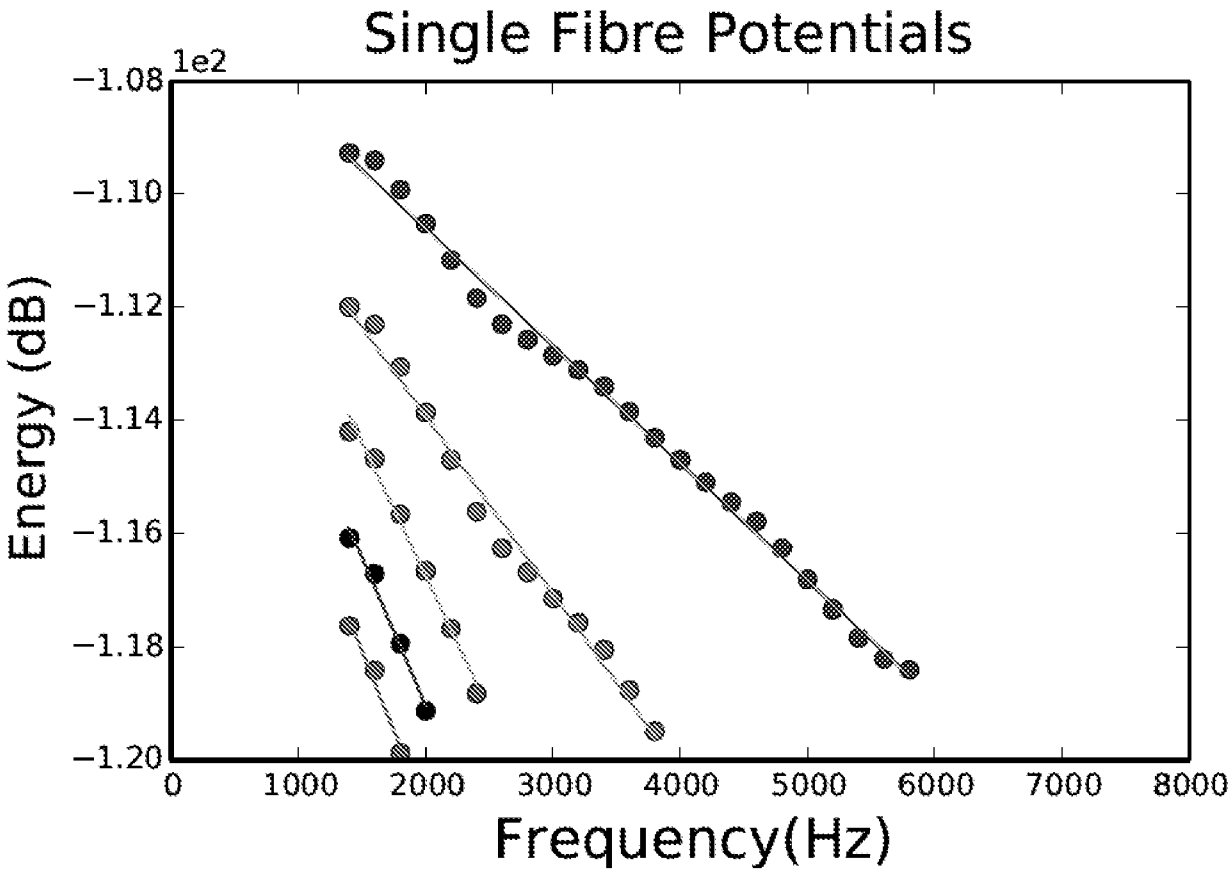

Figure 30c

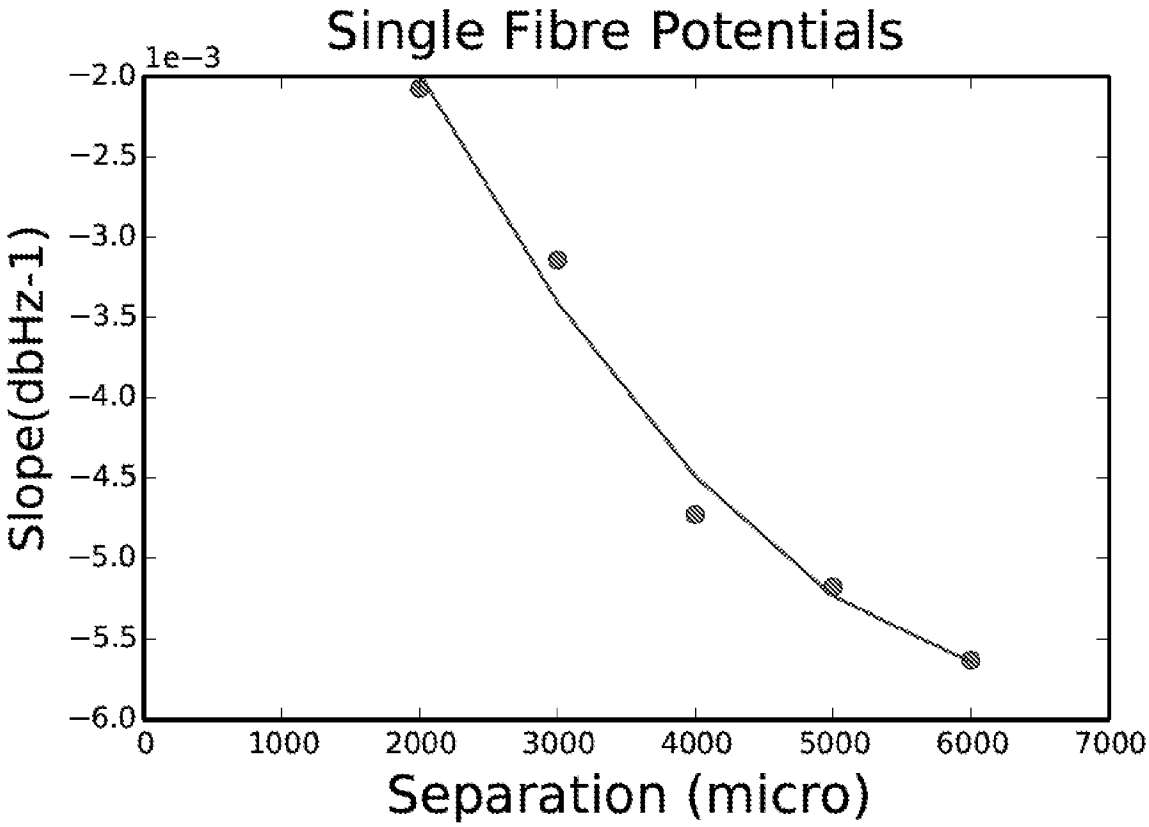
Single Fibre Potentials
1e−3
−2.0
−2.5
−3.0
−3.5
−4.0
−4.5
−5.0
−5.5
−6.0
Slope(dbHz-1)
0
1000
2000
3000
4000
5000
6000
7000
Separation (micro)

Figure 30d