The Design of a Cuff Electrode for Long Term Neural Signal Recording with Selective Recording Capability to Rehabilitate People with Nerve Damage

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Abstract:- This research includes solutions or strategies for recording neural signals at a single point selectively, involving electrical and biomedical issues in relation to the nature and electrical characteristics of the nerve at the point of signal recording and neural channels. During this process, by studying the field of operation of nerve strands and their electrical behavior of them, the identity of each signal be recorded by so-called cuff electrodes as selectively as possible. The goal of current research is to provide a practical solution using scientifically targeted research based on world-class scientific methods for the rehabilitation of people with neurological disorders, people with disabilities and helping them to regain their abilities and return to life as soon as possible with the help of medical and electronics science. In this research, we tried to design a kind of cuff electrode that has a higher sensitivity, better signal detection capability, and better overall performance than previous ones. In this study, nerve environments were simulated in the Comsol Multiphysics -5.2a and the available simulation data for the design of the cuff electrode and the type of the arrangement of the contacts and their placement were applied, and then the signal analysis process and the recorded signal differences in case of Flow and electric flux were investigated and finally the results of the approach were obtained. Based on the data provided by the simulations and medical studies, it was concluded that the electrical behavior of the nerve is completely different from the simple conductors and the electrical current of the nerve is not linear, is completely variable and nonlinear, and follows the Fitzhugh-Nagumo model under specific parameters. Then, by understanding the above subject and forming the above relationship in the ComSol software under specific parameters that are specifically discussed in the thesis, the flow was constructed like the actual flow of the nerve in the simulation environment, and then, the type of arrangement and the material of the contacts for recording were analyzed The precise current flow was examined and the proposed model of the thesis was formed.

Keywords:- *Neural cuff electrode, Neural Problems, Neural impulse, Peripheral nerve, Nerve rehabilitation, Selective recording.*

I. INTRODUCTION

As a start and introduction, a general review of statistics shows that a significant segment of the world's population suffers from disabilities and neurological damage, and the lack of practical solutions to help rehabilitate this percentage of the population on the one hand and the possibility of help and Rehabilitation of these people to restore basic movement or sensory abilities using research and scientific designs led the researchers to design and fabricate electrodes to record signals sent from the brain to repair or improve nerve function in damaged areas.

Till now, these processes have met with great success and with the introduction of electrodes known as cuff electrodes, the reliability and time continuity of the electrodes have increased, butmost of the research is on batch recording of signals and relatively less research on selective recording is done.

Therefore, the present study will examine the design of the cuff electrode for the long-term selective recording of neural signals and will attempt to introduce a practical solution for thispurpose.

The neural cuff electrode microprobe is designed to record and log neural signals reliably. These electrodes are suitable for use in recording all electrical events at the desired location, and are available in a variety of lengths, diameters, and types appropriate to the anatomy of their target nerve.

This research will include providing a solution or solutions for recording nerve signals at a selective point and includes biomedical and electrical topics related to the nature and electrical behaviors of the nerve at the site of recording and neural channels.

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II. MECHANISM OF ACTION POTENTIAL AMPLIFICATION AND ELECTRICAL BEHAVIOR TO NEURONS

Neurons communicate with each other through synapses, where the axon terminal of a cell strikes a dendrites, cell body, or axon of another neuron. For example, neurons such as Purkinje cells in the cerebellum can have more than 1,000 dendritic branches that communicate with tensof thousands of other cells.

Other neurons, like the giant cell neurons of the supraptic nucleus, have only one or two dendrites, each receiving thousands of synapses. Synapses can excitated or inhibit or increase or decrease the activity of target neurons, some neurons also communicate through electrical synapses, which are electrically conductive direct connections between cells.

In 1937, John zachary young showed that the large squid axon could be used to study the electrical properties of neurons, squid cells are larger than human neurons but similar in nature and squid cells are easier to study then accurate measurements of membrane potential were performed by placing electrodes inside squid axons.

The cell membrane of the axon and the cell body contain voltage-sensitive ion channels that allow neurons to produce and amplify an electrical signal (action potential). These signals areproduced and amplified by charged ions of sodium, potassium, chloride and calcium.

There are several stimuli that can activate neurons and lead to electrical activity, including chemical, tensile and compressive transmitters, and changes in electrical potential in the cell membrane. excitation open specific ion channels within the cell membrane, causing ions to flow through the cell space and alter the membrane potential.

Potential is established in the cell membrane. There is an unequal distribution of charged ions on either side of the nerve cell membrane. Using an electrode placed inside the neuron and outside the other neuron, the voltmeter measures the difference in the distribution of ions inside and outside ions. And in this example, the voltmeter reads -70 ^{mV}. In other words, the inside of the neuron is slightly negative relative to the outside. This difference is expressed as the potential ofthe resting membrane.

The membranes of all cells have a potential difference. In neurons, stimuli can reverse this potential difference by opening sodium channels in the membrane. For example, neurotransmitters specifically interact with sodium channels (valves). So sodium ions flow into the cell, lowering the membrane voltage.

When the potential difference reaches the threshold voltage, the reduced voltage causes hundreds of sodium valves in the membrane region to open briefly. Sodium ions flow into the cell and completely depolarize the membrane. In the adjacent membrane, the voltage-sensitive ion channels open further, causing a wave of depolarization along the action-potential cell. When theaction potential approaches its peak, the sodium valves close and the potassium valves open. And allow ions to leave the cell to maintain the normal potential of the membrane.

A resting membrane potential means that there is an uneven distribution of ions on both sides of the nerve cell membrane. This potential is usually measured at about 70 mV (with a negative charge inside the membrane relative to the outside). Therefore, the resting membrane potential is shown as -70^{mV}.

And a negative sign indicates that the inside of the membrane is negative relative to the outside.That potential is called resting because it occurs when the membrane is not trickled or the impulses are not directed (in other words, it is at rest).

III. CUFF ELECTRODE

The neural cuff electrode microprobe is designed for reliable recording or stimulation for peripheral nerves and is available in a variety of lengths and diameters to best fit the target anatomy. The contact distance between the electrode and the cuff design is fully adjustable. , And layout options include cluster, multipolar, or concentric.

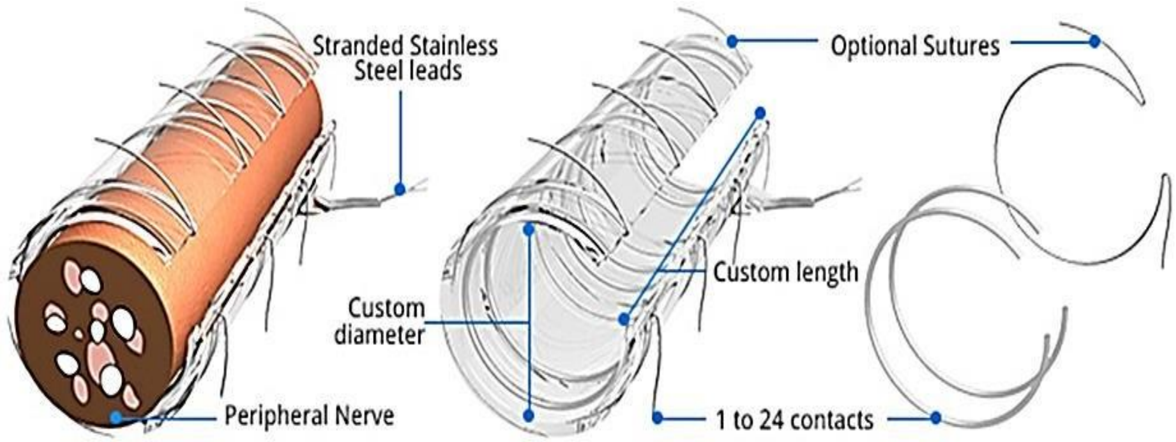

Fig. 1: a cuff electrode view

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Today, cuff electrodes are used to record or stimulate long-term neural signals, and an animal or human can move easily during this recording or stimulation. As shown in Figure 1, these types of electrodes are cylindrical and made of polymer, with internal metal-compatible contacts on the inside, and this electrode can be wrapped around a nerve and tightened. . In some cases, more metal contacts may be embedded inside the electrode, which can be used to amplify stimulation sites or record signals from multiple nerve points (to determine conduction velocity or to pursue nerve regeneration or selective stimulation).

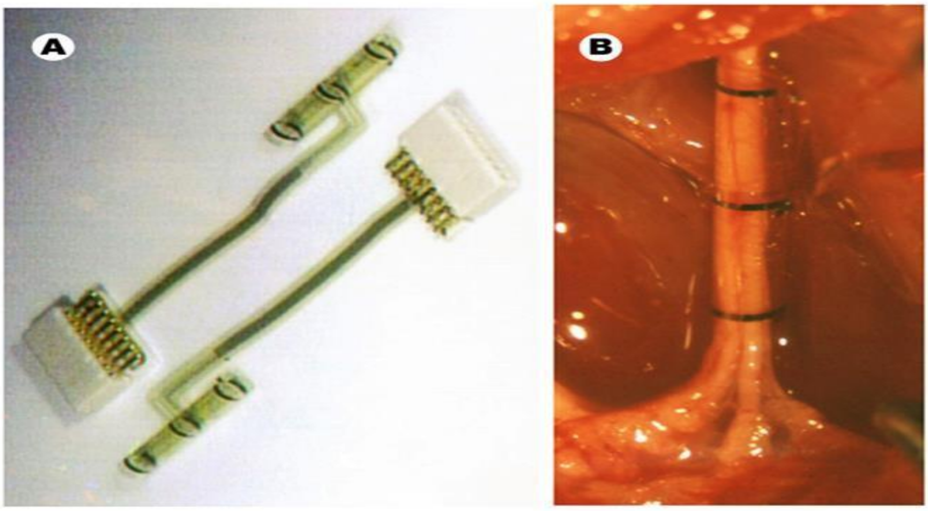

Fig. 2: Previously designed Cuff electrode chronicaly implanted

To eliminate external annoying signals such as signals from adjacent muscles or tissues (EMG), it is best to use three-pole electrodes in which the contacts are spaced evenly apart and the side contacts are short-circuited.

As far as possible, the electrical impedance of all contacts should be low and similar, which will reduce noise and improve common-state noise reduction in differential amplifiers. Metals used as contacts must be non-toxic and compatible with the human body. (Bio-Compatible) Aluminum or copper or silver cannot be used as contacts because they are toxic. The best metalsthat can be used are gold, platinum or stainless steel.

After placing the electrode around the nerve, after a while, the muscles on this electrode will cover and may put pressure on the nerve, so the diameter of the electrodes should be about 20%larger than the diameter of the nerve to prevent pressure from the electrode on the nerve And prevent nerve damage.

Care must also be taken in making and placing the contacts in place so that no smallattachments form on them, as they can cause nerve damage.

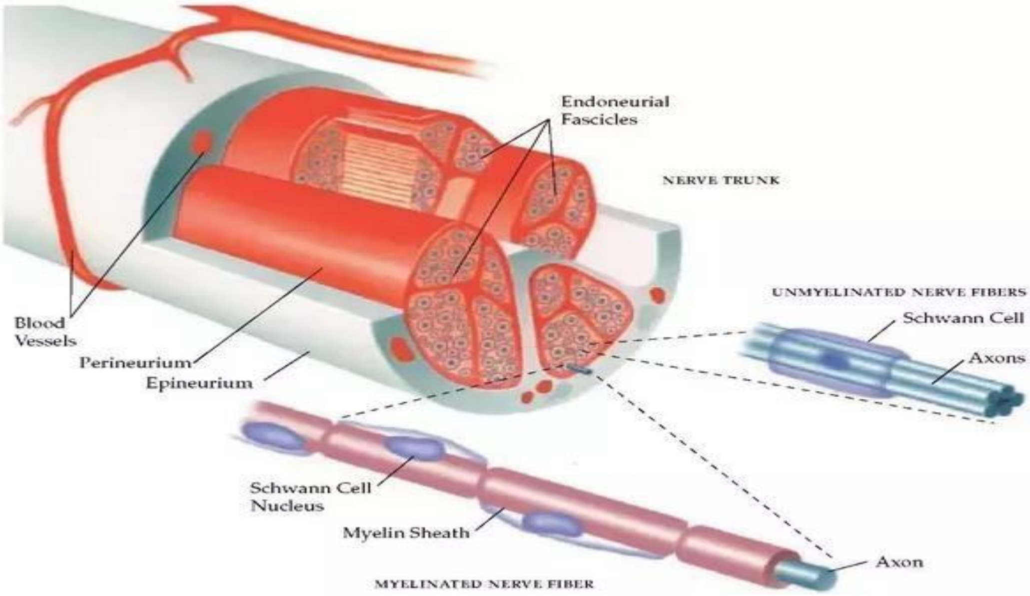

Fig. 3: A nerve anatomy

IV. FITZHUGH- NAGUMO MODEL

In 1962, Richard Fitzhugh introduced a simplified model of common neural models that incorporated most of the features of nerve impulses (except in cases such as a large number of high-frequency pulses igniting all at once). Fitzhugh showed that the variables *m*, *h* and *n* followa first-order differential equation and the answer to these differential equations is:

$$x(t) = x_\infty - (x_\infty - x_0)e^{-\frac{t}{\tau_n}}$$

Eq.1 Fitzhugh- Nagumo equation

x_∞ is the final value or the steady state value of the desired quantity. This equation shows thatall three parameters move exponentially towards their final value.

Now if the time constant is low, it can be shown that $x(t) \approx x_\infty$ which includes variables that change rapidly, such as the variable m for sodium, which controls the opening of sodium channels. So it can be concluded that:

$$\boldsymbol{m(t) \approx m_\infty = m_s(t)} \quad \textbf{(4-1)}$$

And if the rate of change of the variable is low or the time constant is high, we can say that thisvariable has a slow dynamic, such as *h* for sodium and n for potassium.

Since the dynamic behavior of these two variables is almost the same and both are used for the repolarization phase, these two variables can be summarized in one variable and shown with **w (t)** to be related to low-velocity processes. In this case we have two equations to show the membrane potential states *v (t)* and the recovery state *w (t).*

Also in 1962, Nagumo and colleagues were able to develop an electrical circuit using a tunnel diode for a model proposed by Fitzhugh and validate his results. This is why this model is called Fitzhugh-Nagumo and is displayed as follows:

$$\boldsymbol{\frac{dV}{dt} = V(\alpha - V)(V - 1) - W + I} \textbf{(4-2)}$$

$$\boldsymbol{\frac{dW}{dt} = \varepsilon(V - \gamma W)} \quad \textbf{(4-3)}$$

Where ε is excitation parameter , γ parameter is responsible for changing the resting state and dynamics of the model and I membrane stimulation current. In the present study, these equationshave been used to model the propagating nerve impulses.

V. STEP-BY-STEP METHOD OF MODELING AND SIMULATING THE NERVE ENVIRONMENT

The difference between the minimum and maximum value of an action potential of a nerve stringis the basis for recording potentials and is used to evaluate the degree of spatial selectivity of the electrode. In this research, the finite element method (FEM) is used for modeling. This model is simulated by COMSOL software version 5.2, which is suitable for solving finite element problemswith the ability to communicate with MATLAB software.

Using the standard model of an active nerve fiber and simulating a nerve current with a volume conductor, the action potential of a fiber (SFAP) can be obtained by measuring the potential of theactive sites of the electrode. In this study, an anisotrpic and inhomogeneous three-dimensional volume conductor of a rat sciatic nerve was modeled using COMSOL software with the ability to solve Maxwell equations.

In this software, to solve the equations in all models, the linear solver method with a relative error of $1*10^{-6}$ has been used. A cross-section of the rat sciatic nerve consisting of four fascicles is designed in three direct dimensions (unlike previous designs). In this study, the diameter of the sciatic nerve is considered to be one millimeter due to the limitation of the cuff electrode. The fascicles are defined as follows to show the asymmetry of the fascicles in the real environment asfollows:

Table 1: Model sizes and conductivity of materials used in the model

Material	Conductivity(s/m)	Parameter	Value
Saline (NaCl solution)	2	Nerve length	30mm
Epineurium	$8.26*10^{-2}$	Cuff length	20mm
Perineurium	$2.1*10^{-3}$	Diameter of nerve	1mm
Endoneurium	x,y: $8.26*10^{-2}$ z: 0.571	Surface area ofcathodes	$4.8*10^{-4}$ m^2
Cuff (Silicon)	$1*10^{-12}$	Surface area of ringanodes	$3.8*10^{-3}$ m^2
Contacts (Platinum)	$8.9*10^{-6}$		

To prevent pressure on the fascicles, a cylindrical cuff electrode wrapped around the fascicle wasused, as shown in Figure 5. The Association for the Advancement of Medical Equipment (AAMI) has recommended that the inner diameter of the electrode be chosen to be 20% larger than the diameter of the larger fascicle in order to prevent pressure on the fascicle due to electrode implantation and post-implant inflammation.

The rotating cuff electrode contacts are rectangular in shape with dimensions of 2mm × 0.25mm × 0.038mm, which are located at angles of 0, 90, 180 and 270 degrees around the electrode, with a three-pole structure.

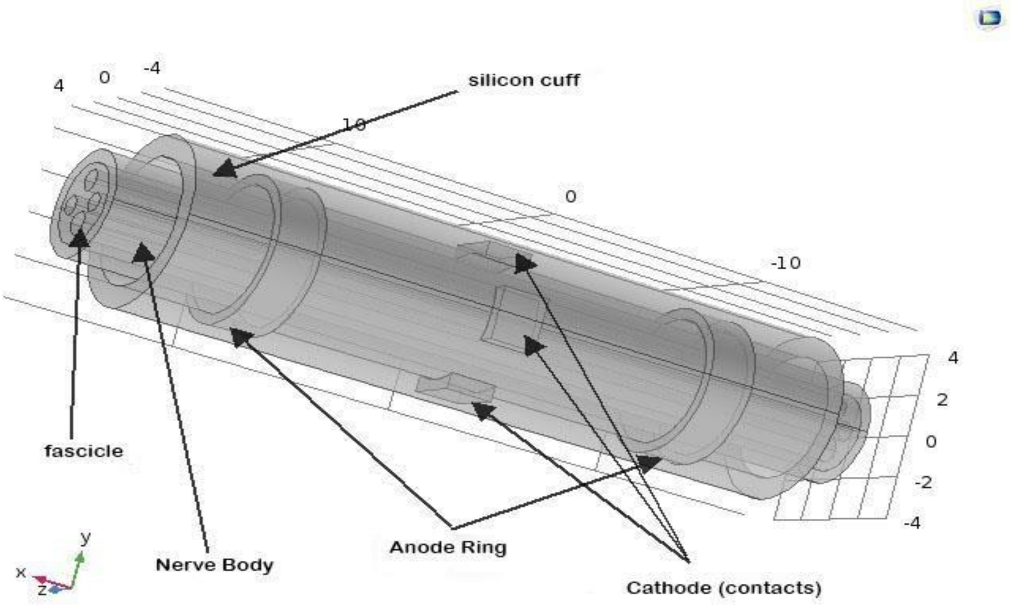

Fig. 4: Side view of the designed nerve and cuff model

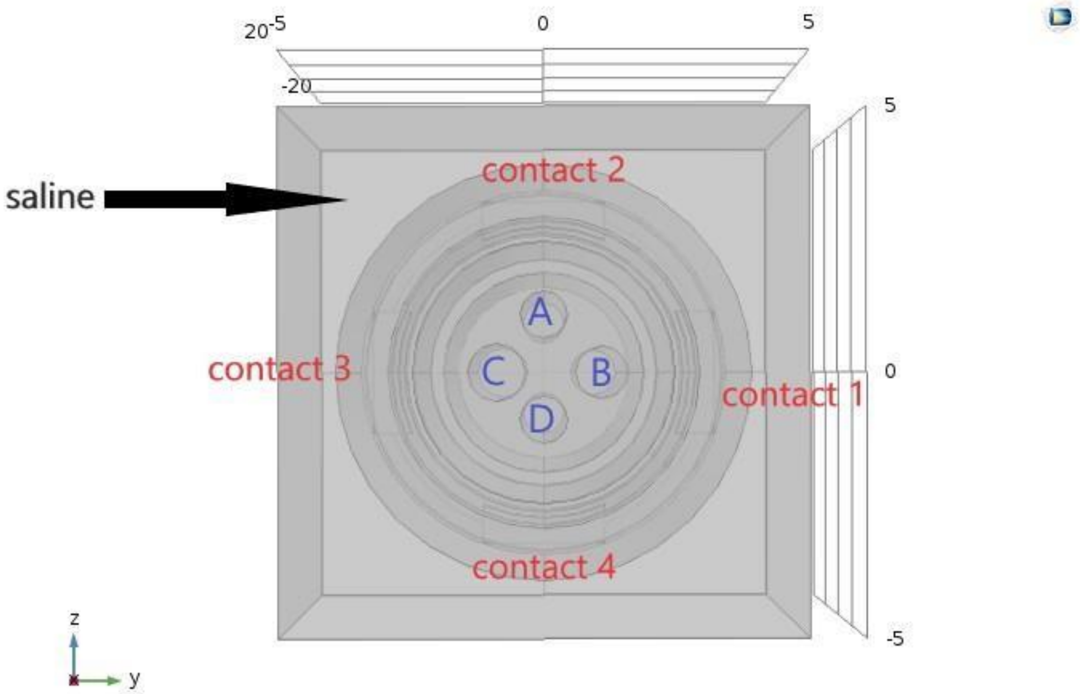

Fig. 5: Facing view of the nerve and cuff design inside the saline environment

Table 2

Parameter	Value	Explanation
alpha	0.159	Excitation threshold (V)
epsilon	0.008	Excitability
beta	1	system parameter
gamma	2.54	system parameter
delta	0	system parameter
V0	1	Elevated potential
nu0	0.3	Elevated inhibitor
D	$1*10^{-0.9}$	Diffusion coefficient

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VI. NERVE DESIGN SIMULATION WITH THE INTRODUCED MODEL

After designing the nerve and cuff model and also creating the Fitzhugh-Nagumo relationship as shown in Figure 6 in comsol 5.2a software, which is a finite element solution software, a three- dimensional, anisotropic and heterogeneous volume emission medium Rat sciatic nerve was modeled by COMSOL software using Time Depended tool to solve Maxwell equations.

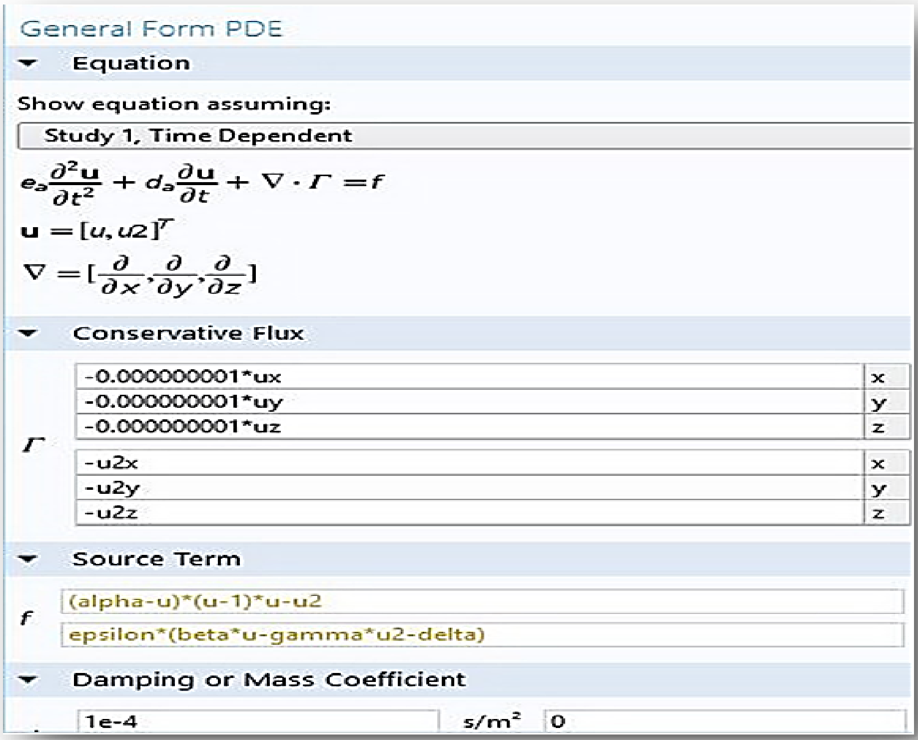

Fig. 6: Simulation of equations in the software environment

In this simulation, a time based solver with a relative error of $1*10^{-6}$ is used for the dependentvariables u and $u2$. In this modeling, a cross section of a rat sciatic nerve consisting of four fascicles of different sizes was simulated.

For the cuff electrode, a nerve with a diameter of one millimeter (rat nerve) was considered. The diameters of the sciatic nerve fascicles are as follows:

- 0.3 mm for Fascicle A.
- 0.4 mm for fascicle B.
- 0.45 mm for fascicle C.
- 0.28 mm for fascicle D.

None of the fascicles were pressurized by the cuff electrode. The thickness of the perianurium layer is equal to 3% of the diameter of the fascicle.

The 2 × 2 mm diffusion medium surrounds the cuff electrode and the nerve. The resting potential of the nerve fiber membrane is almost constant and is considered to be -70 mV. But the production potential of the action potential is -35 ^{mV}. In fact, the activated area is the part of thenerve that has a potential beyond -35^{mV} volts.

VII. SIMULATION RESULTS

Figure 7 shows the potential of the membrane produced by the Fitzhugh-Nagumo equations at three different points of the simulated axon, 5mm, 15mm and 25mm.

As shown in this figure, it takes approximately 600 microseconds for the action potential to travel 3 cm across the nerve. This indicates that the nerve conduction velocity (NCV) is about 0.5 m /s.

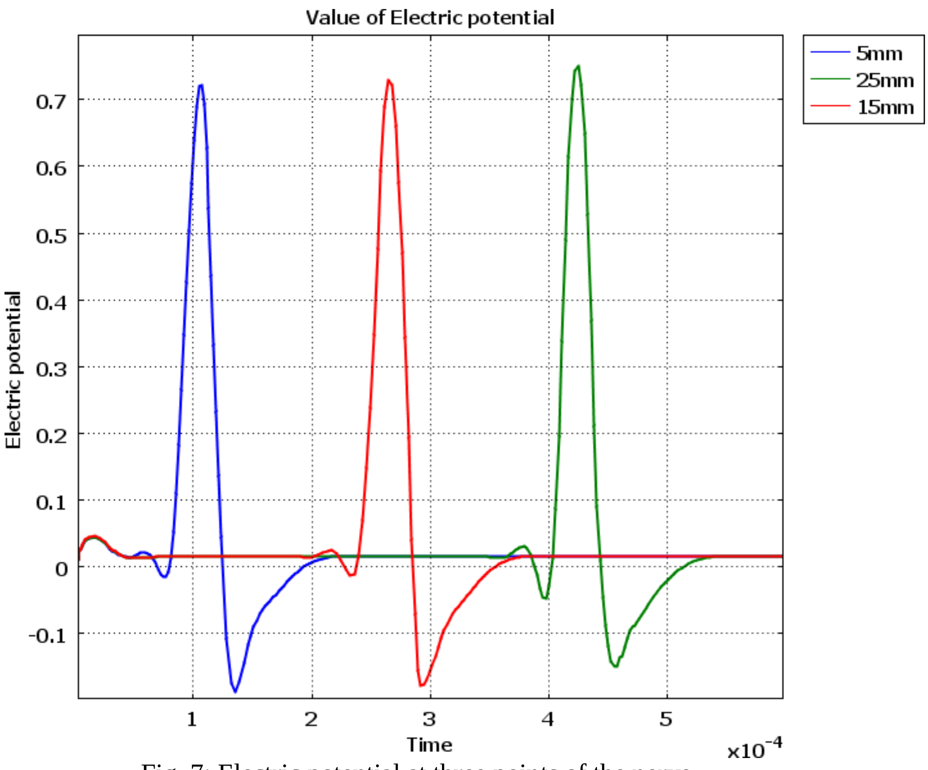

Fig. 7: Electric potential at three points of the nerve

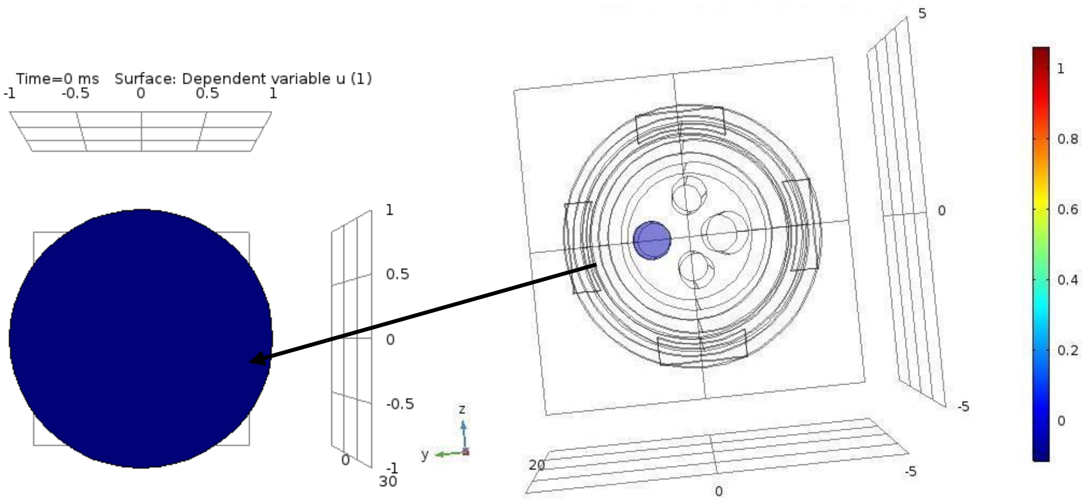

Fig. 8: A view of fascicle B at the first moment (0 milliseconds)

Figures 9 and 10 show that the nerve impulse reaches the center of the cuff electrode at 0.28 milliseconds. At this point, the induced electric potential on the corresponding cathode will haveits maximum value.

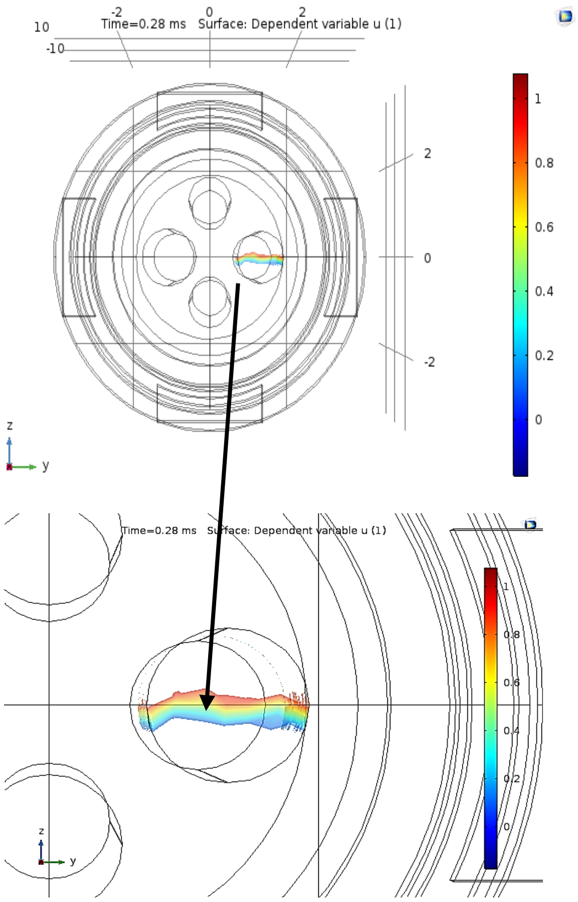

Fig. 9: Slice view on the x-axis when the nerve impulse reaches the center of the fascicle B.

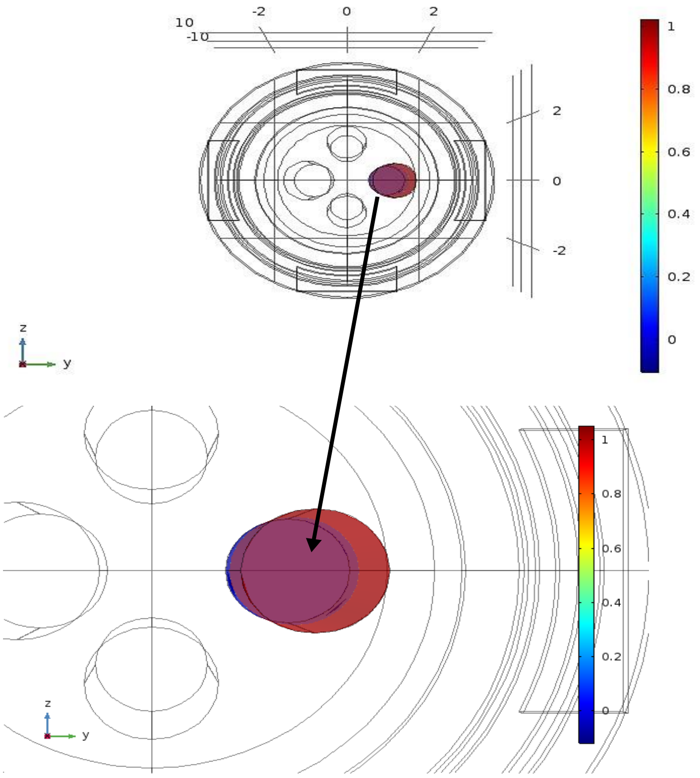

Fig. 10: Surface view on the x-axis when a nerve impulse reaches the center of the fascicle

Table 4 lists the induced electrical potential values on each of the contacts. As mentioned above, the maximum action potential occurs in the middle of the cuff electrode at 0.28 ms. Now, looking at the previous images, we can record the following table for the contact corresponding to fascicle B:

Table 3: recorded Values of action potential for contact 1

time	Contact 1
0.26 msec	15 mv
0.27 msec	45 mv
0.28 msec	80 mv
0.29 msec	50 mv

Now, by repeating the experiment for other fascicles, the following results are recorded:

Table 4: recorded Values of action potential for other contacts

Time	Contact2	Contact3	Contact4
0.26 msec	18mv	7 mv	12 mv
0.27msec	35mv	10 mv	22mv
0.28msec	60mv	14mv	33mv
0.29 msec	20mv	11mv	12 mv

It can be seen that the same results occur for other fascicles and the time of 0.28 milliseconds will be the maximum contact absorption potential corresponding to each contact.

VIII. CONCLUSIONS

In this study, nerve conditions in the real environment and the amount of electrical conductivity and its electrochemical behavior were investigated and then the type of nerve damage was pointed out, then an electrode was introduced as a cuff electrode as research platform, and finally a cuff electrode with 4 Cathodic contacts and 2 anodic contacts were designed and simulated By measuring the transient potential of each fascicle and its effect on each of the contacts, we were more accurate in locating the fascicle carrying the neural signal. It can be stated with a high percentage that in case of accurate fabrication and observance of correct fabrication standards, theexact location of the neural signal can be commented and selected.

This relatively accurate location detecting has provided the selectivity for this approach because the location of the transient action potential that will affect one of the contacts with a higher percentage and higher voltage, and based on the results of Tables 7-1 and 7-2 at a certain point in time, one of contacts will have a higher voltage and that means Selective Recording.

The importance of this issue becomes more tangible and vital when the physician needs to analyze the signal passing through a fascicle or nerve fiber long-term after implantation to comment based on sound data to apply a particular treatment. In this case, by recording the signals passing through a particular fascicle in a specific time period and creating a time-based database for the same fascicle through the recorder contact, we can can comment on a person's neurological problems, lesions, and neural characteristics with greater confidence.

REFERENCES

[1.] Brian Wodlinger “EXTRACTING COMMAND SIGNALS FROM PERIPHERAL NERVERECORDINGS”,jan 2011

[2.] Hamed Taghipour-Farshi, Javad Frounchia, Nasser Ahmadiaslb, Parviz Shahabib and

[3.] Yaghoub Salekzamanic “Assessment on selectivity of multi-contact cuff electrode for recording peripheral nerve signals using Fitzhugh-Nagumo model of nerve excitation”, Journal of Back and Musculoskeletal Rehabilitation 29 (2016) 749–756

[4.] A.Seggio, A. Narayanaswamy, B. Roysam, and D. Thompson, "Self-aligned Schwann cell monolayers demonstrate an inherent ability to direct neurite outgrowth," Journal of neural engineering, vol. 7, p. 046001, 2010.

[5.] Mesut Sahin, Dominique M. Durand, “SELECTIVE RECORDING WITH A MULTI-CONTACT NERVE CUFF ELECTRODE, IEEE Applied Neural Control Laboratory, Biomedical Engineering Department, Case Western Reserve University, Cleveland, OH 44 106.

[6.] Olivier Rossel ‹Fabien Soulier Jonathan Coulombe ‹Serge Bernard ‹Guy Cath´ebras,” Fascicle-selective multi-contact cuff electrode”, 2011 IEEE annual conference

[7.] W. English, G. Schwartz, W. Meador, M. J. Sabatier, and A. Mulligan, "Electrical stimulation promotes peripheral axon regeneration by enhanced neuronal neurotrophin signaling," Developmental neurobiology, vol. 67, pp. 158-172, 2007.

[8.] Al-Majed, C. M. Neumann, T. M. Brushart, and T. Gordon, "Brief electrical stimulation promotes the speed and accuracy of motor axonal regeneration," The Journal of neuroscience, vol. 20, pp. 2602-2608, 2000.

[9.] Guyton and Hall Textbook of Medical Physiology 13th edition 2013

[10.] D. R. Merrill, M. Bikson, and J. G. Jefferys, "Electrical stimulation of excitable tissue: design of efficacious and safe protocols," Journal of neuroscience methods, vol. 141, pp. 171-198, 2005

[11.] D. R. Merrill, M. Bikson, and J. G. Jefferys, "Electrical stimulation of excitable tissue: design of efficacious and safe protocols," Journal of neuroscience methods, vol. 141, pp. 171-198, 2005.

[12.] L. Hodgkin and A. F. Huxley, "A quantitative description of membrane current and its application to conduction and excitation in nerve," The Journal of physiology, vol. 117, p. 500, 1952.

[13.] E. Neher, B. Sakmann, and J. H. Steinbach, "The extracellular patch clamp: a method for resolving currents through individual open channels in biological membranes," Pflügers Archiv, vol. 375, pp. 219-228, 1978.

[14.] Thomas Sinkjær, Morten Haugland, Johannes Struijk and Ronald Riso "Long-term Cuff Electrode Recordings from Peripheral Nerves in Animals and Humans"

[15.] H. Taghipour farshi "Design of a Front-End for a Multichannel Neural Signal Microstimulation System"Phd Thesis Tabriz University.

[16.] J. Zariffa, M. K. Nagai, M. Schuettler, T. Stieglitz, Z. J. Daskalakis, and M. R. Popovic, "Use of an Experimentally Derived Leadfield in the Peripheral Nerve Pathway Discrimination Problem," Neural Systems and Rehabilitation Engineering, IEEE Transactions on, vol. 19, pp. 147-156, 2011.

[17.] J. O. Larsen, M. Thomsen, M. Haugland, and T. Sinkjær, "Degeneration and regeneration in rabbit peripheral nerve with long-term nerve cuff electrode implant: a stereological study of myelinated and unmyelinated axons," Acta neuropathologica, vol. 96, pp. 365-378, 1998.

[18.] K. Thurgood, D. J. Warren, N. M. Ledbetter, G. A. Clark, and R. R. Harrison, "A wireless integrated circuit for 100-channel charge-balanced neural stimulation," Biomedical Circuits and Systems, IEEE Transactions on, vol. 3, pp. 405-414, 2009.

[19.] https://www.microprobes.com/products/peripheral-electrodes/nerve-cuff

[20.] L. Zeng and M. D. Dominique, "Extracellular voltage profile for reversing the recruitment order of peripheral nerve stimulation: a simulation study," Journal of Neural Engineering, vol. 1, p. 202, 2004.