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Experiments on the development and use of a new generation of intra-neural electrodes to control robotic devices

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Abstract - **The development of interfaces linking the human nervous system with artificial devices is an important area of research and several groups are now addressing it. Interfaces represent the key enabling technology for the development of devices usable for the restoration of motor and sensory function in subjects affected by neurological disorders, injuries or amputations. For example, current hand prostheses use electromyographic (EMG) signals to extract volitional commands but this limits the possibility of controlling several degrees of freedom and of delivering sensory feedback. To achieve these goals, implantable neural interfaces are required. Among the candidate interfaces with the peripheral nervous system intra-neural electrodes seem to be an interesting solution due to their bandwidth and ability to access volition and deliver sensory feecback. However, several drawbacks have to be addressed in order to increase their usability. In this paper, experiments to address many of these issues are presented as part of the development of a new generation of intra-neural electrodes. The results showed seem to confirm that these new interfaces seem to have interesting properties and that they can represent a significant improvement of the state of the art. Extensive experiments will be carried out in the future to validate these results.**

Index Terms – Neuro-robotics, neural interfaces, PNS, intra-neural interfaces.

I. Introduction

In the last thirty years, the combination of neuroprosthetic devices (e.g., for spinal cord injured persons) with robotic mechanical devices in hybrid bionic systems (e.g., hand cybernetic prostheses for amputees) has been an area of active research by several groups [1-2]. In all these applications, an artificial neural interface should be introduced to the subject's peripheral nervous system (PNS) that has sufficient selectivity to address the residual afferent and efferent pathways to restore function and sensation in an effective and useful way. For example, in the case of a cybernetic prosthetic, the interface should be able to stimulate different populations of afferent nerves to deliver a variety of sensory feedback information originating from sensors in the prosthetic. Similarly, kinematic and kinetic information for the closed-loop control of a neuroprostheses could be detected from signals originating from natural sensors intercepted by the neural interface given sufficient recording selectivity

Starting from these needs, several neural interfaces have been developed with different characteristics (Figure 1). For example, cuff electrodes have been shown to offer a very reliable and robust platform with a relatively low degree of invasiveness but suffer from limited selectivity even in the case of multi-contact cuff electrodes which have shown some interesting results multi-site cuff electrodes [3]. On the other end of the invasiveness scale, sieve electrodes could represent a very interesting solution with potentially very high selectivity though several unresolved problems with their chronic stability and the requirement to sectioned nerves limit their usability [4].

For these reasons, intraneural electrodes, electrodes that penetrate the body of the nerve either inserted longitudinally (LIFE electrodes [5-6]) or transversally (USEA electrodes, [7]) into the PNS has been investigated as a compromise between selectivity and invasiveness.

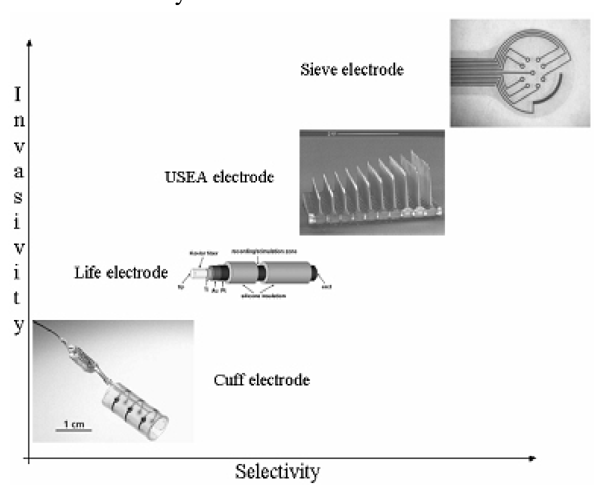

Figure 1: The characteristics of the different neural interfaces in terms of invasivity and selectivity.

Very promising results have been achieved during preliminary experiments for the bi-directional control of hand prostheses [7] with these devices. However, even though intra-neural PNS interfaces seem to be very promising, they currently suffer from several drawbacks limiting their usability : (1) the biomechanical/mechanical properties of the PNS has not been studied in detail in order to gather information useful for the design of effective; (2) the implantation procedure is carried out “blindly” without any possibility of selecting the final position; (3) advanced algorithms are required in order to extract useful neural information. In this manuscript, several experiments are presented in an attempt to address the above issues. The long-term aim of these activities is to develop a new generation of intra-neural PNS interfaces.

II. Characterization of the PNS biomechanical properties

The first step in our work has been the biomechanical characterization of the properties of the PNS during the insertion of needles in order to infer useful information for the design of new interfaces.

An Instron R4464 testing machine (Instron Corporation, Canton, Massachusetts, USA) was used to carry out to identify the mechanical characterization during piercing the peripheral nerve. Five porcine sciatic nerve samples, flash frozen immediately after extraction and thawed just before characterization, were used as test material in this study. A steel needle was linked to the Instron crossbar (see Figure 3 for a schematic of the experimental setup) and driven into the nerve samples. The R4464 Instron was connected via GPIB (IEEE-488) to a PC that controlled the crossbar position and measured the output of the load cell (max force 10 N, sensitivity 0-1% F.S.). The data were sampled at 20 Hz in a routine written in LabVIEW (National Instruments) For each specimen several piercing tasks at different velocities, including low velocities, were carried out and characterized.

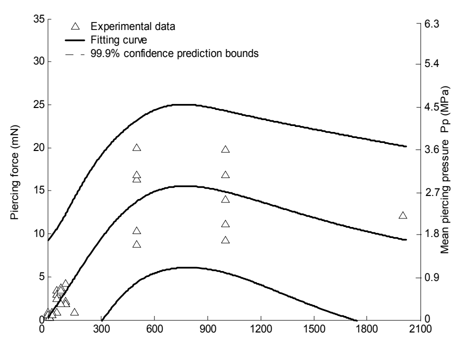

Figure 2: Piercing force in function of the needle tip position (x axis) at v=1000 mm/min. Typical shape of the piercing curve.

From the experimental data it was possible to determine indirectly the tissue sinking under the needle tip. For each experimental velocity (ranging from 1 to 2000 mm/min) a threshold, under which the tissue cannot be pierced, was determined. The force magnitude required for piercing was found to be in the range 0.3 – 25 mN for the different velocities (see Figure 2).

Moreover, interesting results have been achieved showing that differences between piercing carried out at very low velocity (multi-piercing) and piercing at low velocity (mono-piercing) exist and can be correlated with the physical mechanical characteristics of the peripheral nerve.

Experimental data were integrated with a theoretical analysis of the neural interface piercing structure. The problem of buckling, the main failure mode for non-piercing, was analyzed using a non-linear theoretical model that enabled the comparison of different needle geometries and piercing velocities (see Figure 3) to be made.

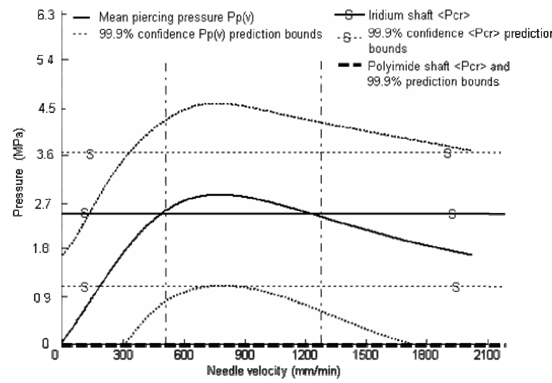

Figure 3: Piercing ability of different kinds of needle. The structures having the first buckling pressure line over the piercing pressure line (with the bounds) can pierce the tissue in experimental velocity range. The others cannot.

III. Actuation of intraneural electrodes

Although LIFEs have been shown to be a potentially interesting neural recording platform, a point of weakness could be their recording stability. Although in the best case, single unit activity has been tracked through the entire 12 months chronic study, it represents the exception rather than the rule. In many cases, the signal amplitude is lost within the first day of implant, before the electrode can be adequately anchored in place by connective tissue.. A working hypothesis to this problem has been related to the relative stiffness of wire type LIFEs, which could easily be displaced in the early stages of implantation following abrupt limb movement. To improve this situation, a flexible thin-film based electrode was developed tf-LIFEs [8-9] (Figure 4) which consists of multiple thin-film sites distributed on a microfabricated flexible polyimide film substrate. The flexibility of the innovative electrode combined together with a larger number of active sites would reduce the effect of small drift of the structure in the initial stages of implantation, and enhancing their stability.

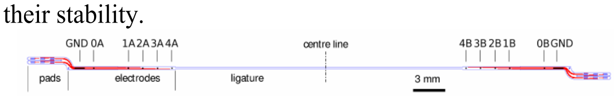

Figure 4 Scheme of a tf-LIFE electrode. Total length:60 mm, pad area length: 5mm. GND is a ground electrode and (L1-L4, R1-R4) represent the recording sites.

Another drawback of conventional LIFEs is that chronic intrafascicular electrodes cannot be adjusted in the event of movement after implantation.

This also means that electrodes cannot be repositioned near interesting cells and there is no flexibility in targeting specific cell types or receptive field positions. So it may be desirable to control the electrode's positions after they are implanted to improve the longevity of holding particular cells in chronic recordings.

In order to achieve this goal the tf-LIFEs were integrated together with micro-actuators create movable contacts on the structure. Shape memory alloys (SMA) were applied as smart actuators to move the contact points of the tf-LIFEs modifying their shape in a selective way. A "serpentine" like shape was memorized by SMA thin films which were attached to the polyimide thin films to simulate the tf-LIFE structure. In parallel, SMA were also glued directly to the electrode (see Figure 5 for the concept and Figure 6 for the first prototype of the SMA-actuated tf-LIFE).

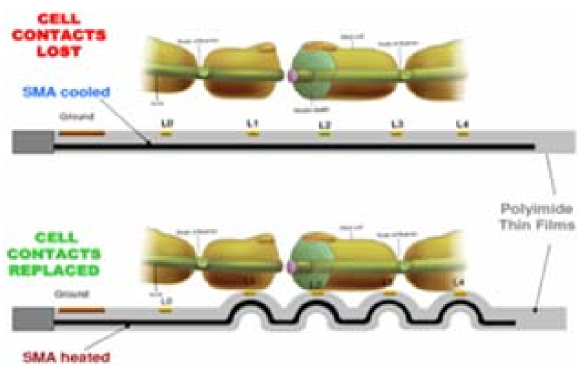

Figure 5 - A basic scheme which represents a tf-LIFE and how its lost contacts could be replaced by SMA actuation.

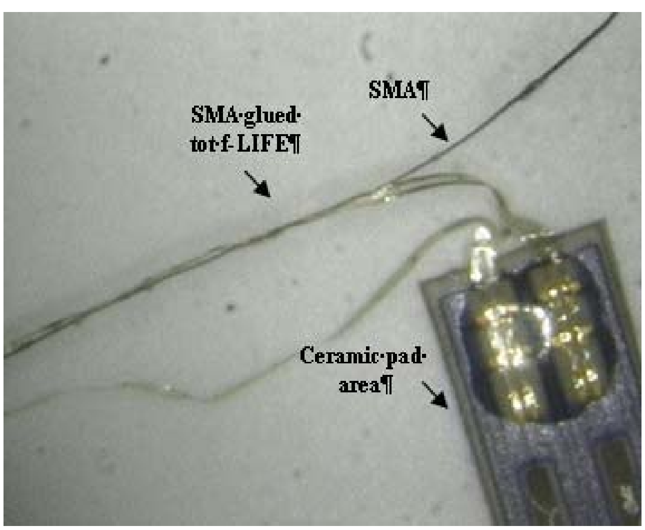

Figure 6. SMA thin film glued to one side of the electrode. Image acquired by Hirox microscopy (50x).

The results of the characterization showed that flexible intrafascicular electrodes actuation by SMA could be a new promising technique to control the position of the active sites of the tf-LIFE inside the nerve, thus improving their performance for long-term applications.

IV. Extraction of Neural Information Using Advanced Processing Algorithms

Much work has been done in the processing of ENG signals recorded using cuff electrodes (e.g., [10-11]). However, relatively little work has been performed to optimize the extraction of neural information from signals originating from intra-neural electrodes, in particular with tf-LIFEs. For example, in ENG signals recorded using intraneural electrodes, it is possible extract single unit information appearing as spikes in the neural recording, while cuff electrodes generally can only resolve the mass activity of the nerve (see Figure 7). This characteristic can allow to develop algorithms potentially that are able to detect different spikes (related different neural stimuli) increasing the selectivity of the electrode.

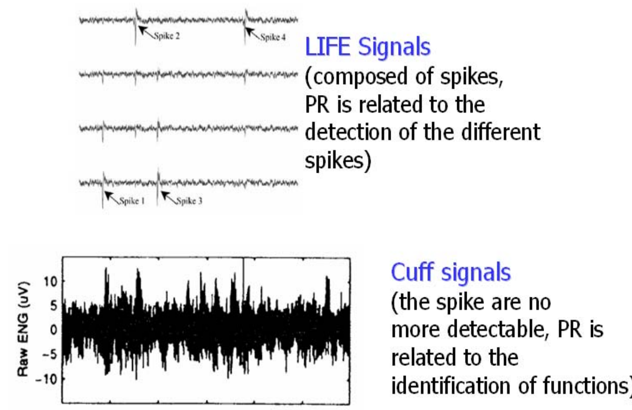

Figure 7: Typical ENG signals recorded using tf-LIFEs (top) and cuff electrodes (bottom)

In order to start addressing this issue, tf-LIFEs were implanted in the sciatic nerve of rabbits. Different kinds of stimuli were applied to the paw of the rabbit: (1) squeezing the foot with knee at 90o; (2) squeezing the foot with knee released; (3) ankle flexion; (4) toe extension; (5) toe extension combined with ankle flexion. Each was repeated 4 to 6 times.

These signals were processed to determine whether the different modes of information could be decoded. Signals were Kalman filtered, wavelet denoised, and spike sorted. The classes of spikes found were then used to infer the stimulus applied to the rabbit. Although the signals acquired from a single tLIFE gave poor stimulus recognition, the combination of the signals from multiple sites led to better results (see Figures 8 and 9). The spike sorting algorithm is also helped by the use of temporal correlation among the channels. The results achieved seem to show the possibility of extracting different neural information exploiting the potentials of multi-site neural interfaces.

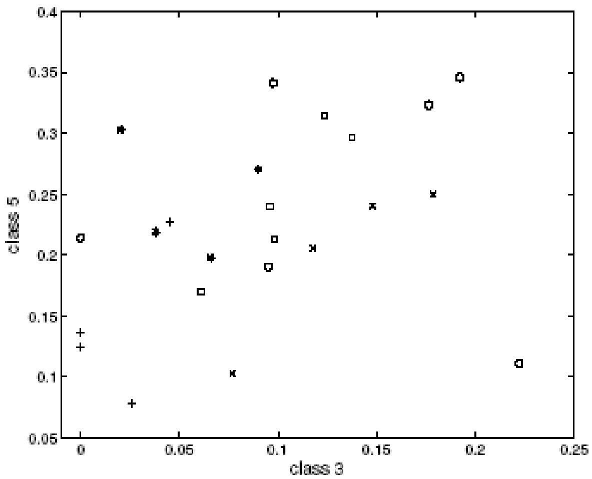

Figure 8. Scatter plot of fraction of spikes in class 3 vs fraction of spikes in class 5 for different stimuli: squeezing foot with knees at 90 degrees (o) and released (×), ankle flexion (+), toe extension (*), toe extension combined with ankle flexion (□). The different kinds of stimuli are hardly separable.

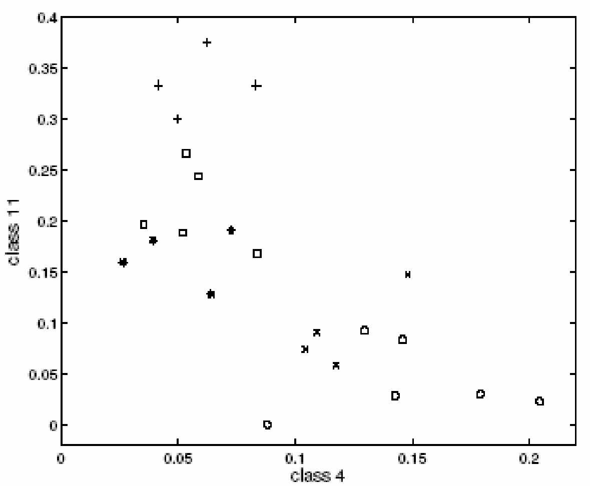

Figure 9. Scatter plot of fraction of spikes in class 4 vs fraction of spikes in class 11 for different stimuli: squeezing foot with knees at 90 derees (o) and released (×), ankle flexion (+), toe extension (*), toe extension combined with ankle flexion (□). The different kinds of stimuli are approximately separable. Two categories are well separated touch/pain (o,×) from proprioception sensations (+,*,□).

IV. Discussion and Conclusions

The development of an effective neural interface is essential for obtaining an intuitive and natural control of hybrid bionic systems. Intra-neural PNS interfaces seem to represent a very interesting solution in the short term because of their good properties in terms of reduced invasivity and good selectivity. However, they have many drawbacks limiting their usability. In this manuscript, some experiments are presented aiming at addressing these issues.

The results of these experiments could allow in the next future the development of a new class of intra-neural interfaces able to reduce the invasiveness (for example thanks to the use of advanced materials and of the information gathered using biomechanical models), to extract several information (using advanced techniques), to deliver a sensory feedback (increasing the number of contacts in the interface), and to place the electrodes optimizing the desired signal-to-noise ratio.

Extensive experiments will be carried out in the future to verify the potential of this approach.

Acknowledgments

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References

[1] Popovic D., Sinkjaer T., “Control of the movement for the physically disabled”, Springer-Verlag; 1st edition, May 15, 2000.

[2] Navarro X., Krueger T.B., Lago N., Micera S., Dario P., and Stieglitz T., “A critical review of interfaces with the peripheral nervous system for the control of neuroprostheses and hybrid bionic systems”, J Pher Nerv Sys, (in press), 2005.

[3] Tarler M.D.; Mortimer J.T., “Selective and independent activation of four motor fascicles using a four contact nerve-cuff electrode”, IEEE Trans Neural Sys Rehab Eng, vol. 12 , no.: 2 , June 2004.

[4] Lago N., Ceballos D., Rodriguez F., Stieglitz T., Navarro X., “Long term assessment of axonal regeneration through polyimide regenerative electrodes to interface the peripheral nerve”, Biomaterials. 2005 May;26(14):2021-31.

[5] Lawrence S.M., Dhillon G.S., Jensen W., Yoshida K., Horch K.W., “Acute Peripheral Nerve Recording Characteristics of Polymer-Based Longitudinal Intrafascicular Electrodes”, IEEE Trans Neural Sys Rehab Eng, vol. 12, no. 3, pp.345-8 , 2004.

[6] McDonnall D, Clark GA, Normann RA, “Interleaved, multisite electrical stimulation of cat sciatic nerve produces fatigue-resistant, ripple-free motor responses”, IEEE Trans Neural Syst Rehabil Eng. 2004 Jun;12(2):208-15.

[7] G. S. Dhillon, S. M. Lawrence, D. T. Hutchinson, and K. W. Horch, “Residual function in peripheral nerve stumps of amputees: implications for neural control of artificial limbs,” J Hand Surg-AM, vol. 29, pp. 605–18, Jul 2004.

[8] K. P. Hoffmann, K. P. Koch, T. Dörge, S. Micera, “New technologies in manufacturing of different implantable microelectrodes as an interface to the peripheral nervous system”. IEEE BIOROB 2006.

[9] K. Yoshida, D. Pellinen, D. Pivin, P. Rousche, and D. Kipke, “Development of the Thin-Film Longitudinal Intrafascicular Electrode”.Proceeding of the fifth Annual Conf.of the IFESS pp.279-284,2000.

[10]Micera S, Jensen W, Sepulveda S, Riso RR, Sinkjaer T (2001). Neuro-fuzzy extraction of angular information from muscle afferents for ankle control during standing in paraplegic subjects: an animal model. IEEE Trans Biomed Eng 48: 787–794.

[11]Cavallaro E, Micera S, Dario P, Jensen W, Sinkjaer T (2003). On the intersubject generalization ability in extracting kinematic information from afferent nervous signals. IEEE Trans Biomed Eng 50:1063–1073.