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Implantable Myoelectric Sensors (IMES)

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ABSTRACT: We are developing a multi-channel/multifunction prosthetic hand/arm controller system capable of receiving and processing signals from up to sixteen Implanted MyoElectric Sensors (IMES). The appeal of implanted sensors for myoelectric control is that EMG signals can be measured at their source providing relatively cross-talk free signals that can be treated as independent control sites. Therefore the number of degrees-of-freedom that can be simultaneously controlled and coordinated in an externally-powered prosthesis will be greater than with surface EMG or mechanical control sites. To explore the issue of intra-muscular signal independence and the ability to control them, human subject experiments have been performed in which intra-muscular EMGs were obtained. Choice of muscles was based on a desire to be able to independently control a two degree-of-freedom (DOF) wrist, and 3 DOF prosthetic hand. This paper provide our result so far.

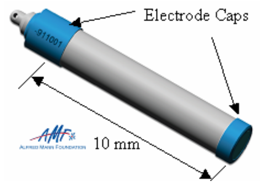

Fig 1: Alfred E. Mann Foundation RFB BION® Package. The inter-electrode spacing is appropriate for whole muscle activity recordings.

INTRODUCTION: The limitation of current prostheses is not the devices themselves but rather the lack of sufficient independent control sources. A system capable of reading intra muscular EMG signals would greatly increase the number control sources available for prosthesis control. We are developing a chronically implantable sensor system to create multiple control sites to detect commanded movements. We are developing myoelectric sensor capsules **(Fig. 1)** that can be chronically implanted into the residual muscles of an amputee’s arm. By localizing the points at which myoelectric activity is detected, these points can be treated as independent control sites with minimal cross-talk. Consequently, the number of degrees-of-freedom that can be simultaneously controlled and coordinated in an externally powered prosthesis will be greatly increased in comparison with surface EMG sites, while obviating the problems of tapping into cut motor control nerves. Our Implantable MyoElectric Sensor (IMES) system[1] will be capable of reading EMG signals from up to 16 inductively coupled, implanted bipolar differential electromyographic (EMG) sensors. These sensors receive their power, digital addressing, and command signals from an external transmitter/receiver coil worn by the patient. The external coil required for the inductive link is laminated into the prosthetic socket such that this coil will encircle the implanted electrodes (**Fig. 2**). Each implanted sensor acts as an intramuscular electrode to detect the electrical activity generated as a by-product of normal muscle contraction. The implants transmit these muscle signals, or myoelectric (EMG) signals, over a shared transcutaneous magnetic link. Each sensor’s electronics and associated circuitry will be housed in a *previously developed* RFB BION® hermetically sealed package provided by the Alfred Mann Foundation

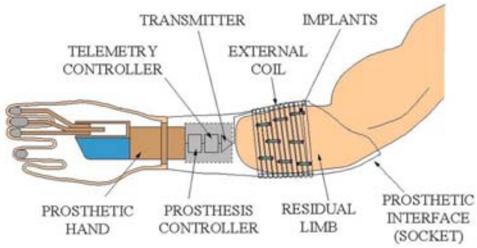

Fig 2: Schematic of planned Implanted Myoelectric Sensor (IMES) system. The external coil is laminated directly into the prosthetic interface. Signals from the implants in the arm, linked through the external coil, control the prosthesis via reverse telemetry. Implant power is supplied through the

(AEMF)[1]. *A major attraction of the BION® technology is that the hermetically sealed ceramic capsule and electrodes necessary for long-term survival in the body have previously been granted FDA IDE approval for use in Functional Electrical Stimulation applications.* Furthermore, no wires are required to be surgically threaded down the arm. No wires are required to penetrate the skin. implant and telemetry design. An external prosthesis controller will decipher user intent from telemetry sent over a transcutaneous magnetic link by the implanted electrodes. The same link will provide power for the implanted electrodes.

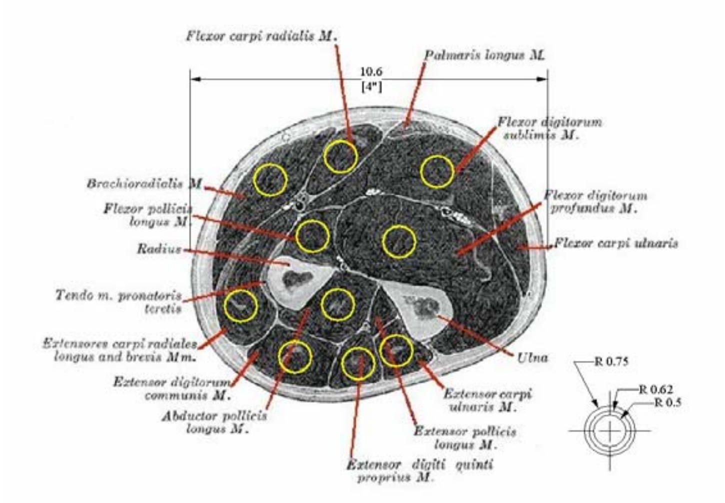

Fig. 3: Projected pick-up area for our Implanted Myoelectric Sensors (IMES) superimposed on an appropriately scaled section of through the proximal forearm. Image shows that pick-up area should not be an issue in the final device with each device having a pick-up area confined to an individual muscle.

We have simulated the potential pickup area for our implants, using electromagnetic finite element modeling techniques[2]. The simulation results suggest that the presence of a thin layer of encapsulation tissue around the IMES should not impede the detection of EMG signals from the surrounding muscle fibers and may in fact cause the amplitude of the EMG signal to increase modestly. The orientation of the bipolar electrode with respect to the fiber direction is an important factor in determining the selectivity of the implanted electrode. Alignment of the electrode along the fiber direction will be particularly important in smaller muscles. We found that for an implant placed along the fibers of the muscle in which it is inserted the pickup area for the sensor will be a cylinder about 5mm in radius about the implant (**See Fig. 3**).

CONTROLLER ALGORITHM DEVELOPMENT

We have conducted extensive human subject experiments in an effort to elucidate the best method of control to use to integrate the contributions from the 16 different implanted sensors. The multi-channel/multifunction prosthetic hand/arm controller must be capable of receiving and processing signals from the implant telemetry system. This same controller must then decipher user intent from the telemetry sent by the IMES to decide which actuators in the prosthesis to drive. From a control perspective there are a number of levels of sophistication that a prosthesis controller can implement.

The simplest control paradigm is to use one muscle to control one function in the prosthesis [*one muscle – one function]*. The implicit assumption underlying this approach is that a contracted muscle EMG signal power is much greater than the relaxed muscles EMG signal powers i.e. high SNR. This system will work with an identifiable signal controlling a single function. For example, using a biceps EMG for elbow flexion and a triceps EMG for elbow extension works quite well. When multiple discrete signals can be identified that directly relate to a given arm function, simultaneous control of multiple degrees of freedom can be obtained resulting in very natural, easy control for the amputee. From a controller design standpoint this is the simplest form of control to implement, and our initial experiments were targeted towards demonstrating this type of control. We anticipate this type of

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control will be possible for the shoulder and the elbow. This type of control is problematic for the wrist and hand because multiple muscles control the same functions—the distal arm is an indeterminate system. Even with all of the control signals available for the arm (such as recording EMG from every muscle in an able bodied person) there is not an exact solution defining movement and different people use their muscles in different patterns to achieve the same function. To use this type of control, without an internal model of the hand, the user needs to relearn how they use their muscles and to think/remember which muscle is tied to which function. This can cause excessive mental loading on the part of the user and yield less-intuitive control.

A more sophisticated approach is to recognize patterns/features of EMG activity associated with different training movements and/or functions and have the controller drive the appropriate prosthesis actuators *[i.e. one pattern of EMG activity - one function]*. The control is intuitive and easy to remember since users execute the movement the prosthetic limb is to perform with their “phantom limb.” Recognition is achieved through feature extraction algorithms, artificial neural nets, or some other similar high level classification method. We have been recording the patterns of EMG activity associated with different training movements and/or functions (**Fig. 4**) and then training the controller to recognize these patterns. We have explored recognition through automated classification techniques using neural networks, linear discriminant anslysis, fuzzy clustering[3], and multivariate linear regression. These systems “learn the patterns produced by each user and are optimized to map the EMG activity from multiple channels to the desired limb motion. This approach requires a pattern to be stored for **every** desired movement. *One EMG pattern - one function* does not provide true parallel/simultaneous control of multiple degrees-of-freedom but could enable seamless-sequential control to be executed. The problem with these approaches is that they are still task-based control systems. The user is still compelled to think in terms of what grip they need to do rather than subconsciously reaching out and picking up an object. We have yet to draw any definitive conclusion as to which classification technique we prefer, but the fuzzy clustering method does lend itself to a default “safe” [fail safe] approach when considering driving motors that are inherently non-backdrivable (as are commonly used in prosthetics to preserve power and to hold commanded position and force in the absence of power).

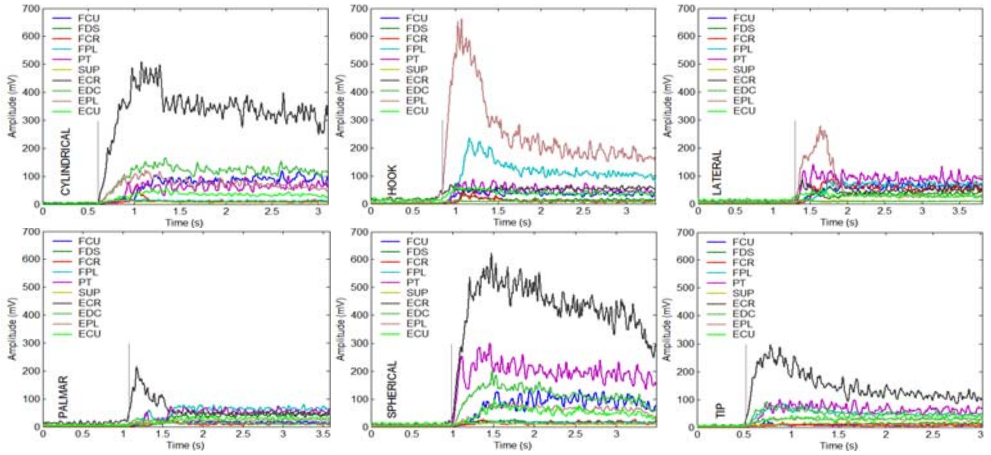

Fig. 4: We have been performing fine wire intramuscular EMG recordings from up to 10 muscles simultaneously in the forearm of human subjects while these subjects perform various grasp patterns and joint functions. The traces above show a typical set of recordings for one subject. Notice how the pattern of EMG activation is distinct for each grasp type.

SYSTEM ARCHITECTURE DEVELOPMENT:

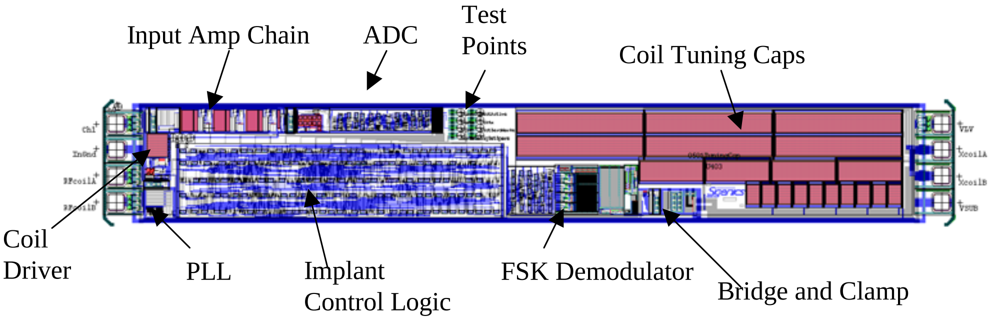

Fig. 5: The Current Implant IC.

A large portion of our efforts have been dedicated to the development of an overall system architecture that maximizes the number of implant devices, and the telemetry bandwidth of those implants, that can be serviced by a single external controller. The Texas Instruments MSP430 family of low-power microprocessors has been selected to perform the high-level Telemetry controller functionality, such as power and implant control as well as outward data decommutation. Most of the higher-level programming will be done in C, with lower-level data streaming routines programmed in assembler. A communications protocol and command set for the telemetry controller has been defined. The prosthesis controller decides which implants require monitoring, and sets up the telemetry controller using a command language to tell the telemetry controller how to configure a serial stream of telemetry output to prosthesis controller input data.

Development of the system is well under way. We are assembling new Class-E exciter modules to test the new implants[4]. PC boards are in hand waiting for the new Class-E Controller chip. We will continue the magnetics design to verify reception of outward telemetry. Next we need to Release the official Interface Control Document (ICD), which completely specifies the interface between the telemetry controller (MSP430) and the prosthesis controller. So that the prosthesis controller can take the data sent from the implant system and use it to control the prosthesis. To date, we have submitted three fabrication runs on the XFab CX08 process. Devices from the first and second runs are being tested. We have chips back from the third run and these are currently in test. These chips are in their near final form factor to fit in the AEMF capsules (**Fig. 5**), however, initial tests have revealed some non-insurmountable defects that are being corrected in a number of up coming wafer runs. The near future goal for the hardware development is an end-to-end demonstration of the system.

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