# Control of upper-limb prostheses: a case for neuroelectric control

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*A discussion is presented on the control aspects of upper-limb prostheses, with emphasis on the areas of necessary improvements in current designs. Arguments are presented to indicate that it should be possible to obtain a substantial improvement in prostheses control by properly training the amputee, improving the dynamics response of the prostheses, and improving the quality of the forward-path control signal. Augmentation of feedback information, although useful, may not be essential. The limitations of the myoelectric (muscle) signal as a forward-path control signal, especially for multiple degrees-of-freedom prostheses, is discussed. Most of the limitations of the myoelectric signal can be overcome if the neuroelectric (nerve) signal is used as a forward-path control signal. Results of a series of experiments which demonstrate the feasibility of constructing an electrode capable of being implanted around severed nerves and of detecting neuroelectric signals for prolonged periods of time are presented. A possible scheme for employing neuroelectric control is also presented.*

## Concept of upper-limb prostheses control

There is substantial evidence [1] suggesting that effective neuromuscular control of the position of a limb depends on a closed-loop control system. Motor (command) signals from the central nervous system are used in the forward path; sensory signals, such as those from the proprioceptive receptors in muscles and joints, are used in the feedback path. Ideally, a prosthesis which replaces a limb should also work in a closed-loop fashion with the remaining neuromuscular system. Such an arrangement is presented in Figure 1. Note that both the motor and feedback paths require an appropriate interface between the prosthesis and the neuromuscular system of the amputee. The feedback is represented as those modes that are "available" and "unavailable" to the amputee in current prosthetic design. The two shaded interfaces as well as the dynamics response of the prosthesis remain as the main problematic areas of prosthesis design.

Current versions of prostheses rely on the available feedback information provided by visual monitoring of the prosthesis movement, the mechanical noise, and the variations of vibration and pressure exerted by the socket as the prosthesis moves. It is generally accepted that an improvement in the quality of feedback to the central nervous system is desirable. Attempts in this direction have as yet met with limited success [2,3,4,5,6].

However, deficient feedback information may not be the limiting factor for creating a symbiotic association between an amputee and his prosthesis. Recently, investigators have shown that intact proprioceptive sensory feedback is not essential for controlling limb movements. It has been shown that primates with totally severed dorsal roots remain capable of performing limb-positioning tasks that they were trained to perform prior to dorsal root severance [7]. The results of Taub *et al.* [8] were even more astounding. They reported that infant monkeys whose afferent nerves were severed at birth developed motor control that enabled them to perform most types of forelimb movements, even when deprived of a visual input. In a recent study, Mooney and Reswick [6] stimulated nerves of amputees with a signal of variable frequency corresponding to the position of a hook that was myoelectrically controlled. Preliminary results indicate that the additional neuroelectric feedback is not an overwhelming addition to the functional use of a prosthesis. In light of these findings it is conceivable that a substantial improvement in the performance of a prosthesis can be achieved with a combination of proper training techniques, better dynamics characteristics, and an improvement in the forward path man-machine interface.

Most of the prostheses currently worn by amputees are controlled mechanically. Power for the actuation of the prosthesis is obtained from relative movements of body musculature, which is transmitted to the prosthesis by a Bowden cable. The return motion is executed by means of gravity or, in some cases, a spring. This arrangement restricts the number of possible movements of a prosthesis. Furthermore, this type of control requires great concentration and effort by the amputee. In order to obtain a substantial movement in the prosthesis, large muscle forces and displacements are required. The effort of meeting these demands can cause muscle fatigue and loss of interest for the amputee.

Current prostheses design is focused primarily on the development of externally-powered, electrically-driven prostheses controlled by the myoelectric signal (representing the command signal). This signal is detected from the skin above the residual muscles of the limb and trunk that remain under the control of the central nervous system. Electrically-driven prostheses with myoelectric control have dominated prosthetic development because they have several advantages over other types of prostheses: the myoelectric signal is conveniently detected on the surface of the skin; the electric battery is possibly the most convenient form of power supply that can be incorporated into a prosthesis; they can be adapted to proportional control with relative ease; the required electronic circuits can be continuously improved and condensed with the current rapid development of electronics components and techniques; and they appear to have the prospect of better long-term reliability. Nevertheless, there are noteworthy exceptions, such as the pneumatically-driven, mechanically-controlled upper limb prostheses described by Simpson and Smith [9].

However, the combination of the varied modes of feedback, in conjunction with the myoelectric signal forward path, is not always an entirely satisfactory arrangement for enabling an amputee to control a prosthesis in a "natural" fashion. This is particularly true when the available myoelectric signals do not originate from muscles that were directly related to the movement to be controlled, or when the displacement of a prosthesis is not proportionally related to the effort of the muscle contraction. In such cases, the amputee must concentrate to supervise the movement of the prosthesis. This is one of the limiting factors in the patient acceptance of the various single degree-of-freedom myoelectric prostheses that have been developed [10,11,12,13,14]. Yet another limitation is that the dynamics characteristics of current prosthesis designs are essentially different from the limb they replace [15].

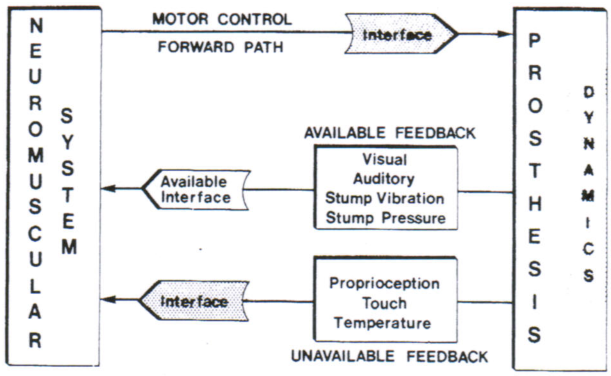


*Figure 1. A schematic representation of a prosthesis interacting with the body's neuromuscular system in a closed-loop fashion. The shaded sections indicate problematic areas which must be perfected to achieve a more symbiotic interaction between an amputee and the prosthesis.*

Throughout this article, the term "degree-of-freedom" will refer to a pair of antagonistic motions of a limb or prosthesis, for example, flexion and extension of the elbow. The term "function" will refer to an individual motion, for example, flexion of the elbow.

## Forward path of multiple degrees-of-freedom prostheses

A forward path man-machine interface becomes a major concern in the control of multiple degrees-of-freedom prostheses. The approaches using the myoelectric signal currently being investigated can be classified in two groups: those that use control sites associated with muscles not necessarily related to the desired movement [16,17] and those that use control sites associated with muscles directly related to the desired movement [18,19,20,21,22]. The second class is more desirable from the point of view of yielding a more "natural" control scheme.

Attempts are being made to control multiple degrees-of-freedom in the upper limb by using direct pattern recognition approaches. Examples are the Swedish hand developed by Herberts *et al.* [18] and the arm developed by Wirta and Taylor [19]. In these cases the myoelectric signal is simultaneously recorded from several remaining muscles. Distinct and repetitive patterns of the myoelectric signals are then used to control different degrees-of-freedom. This approach has proven to be partially successful in situations where the muscles that controlled the desired movement are present. For other situations, the approach has not yet been successful. In an effort to rationalize this approach, Jacobsen [20] has developed an elegant control theory employing statistical techniques coupled with a mathematical formulation of musculo-skeletal dynamics to estimate desired limb motions. However, the practical implementation of his approach in controlling a multiple degrees-of-freedom arm [21] requires substantial physiological and anatomical details not yet readily available.

In its present stage, the concept of myoelectric control for multiple degrees-of-freedom requires the use of several differential recording pairs of electrodes for detecting distinctly different signal patterns from the surface of the skin. The use of several recording pairs of electrodes presents two technical problems. First, the impedance of each differential electrode pair must be maintained reasonably stable with respect to each other. This becomes a problem of major concern during sweating because the electrolyte accumulation on the skin provides a conductive path between the two electrodes of a differential pair and thus partially short-circuits the differential pair. The other, as yet unresolved, problem concerns the harness that supports the recording electrodes. All the electrode pairs must be consistently positioned in the same location above the muscles each time the prosthesis is replaced on the amputee. Both of these problems will cause variations in the signal patterns corresponding to distinct degrees-of-freedom.

Freedy *et al.* [22], who detect signal patterns by recording from numerous locations in the shoulder and trunk, employ a microprocessor to update any temporal change in the signal patterns. In so doing they are confronted with a host of technical problems that for the present remain unresolved. Graupe [23] has attempted to circumvent the above problems by employing a microprocessor to perform real-time autoregressive analysis on the myoelectric signal detected by one judiciously located electrode pair. He states he can distinguish autoregressive parameters associated with four distinctly different limb functions with a success rate of 85 to 95 %. However, in his case, the problem of precise electrode relocation becomes paramount. Furthermore, amputees would find a failure rate of 5 to 15 % very annoying.

The two technical problems of maintaining stable impedance between each electrode pair and consistent positioning of each pair could essentially be eliminated by implanting the electrodes for recording the myoelectric signal into the muscles of interest. However, surgical intervention lacks the convenience of myoelectric control with surface electrodes.

Irrespective of the technical approaches, the muscles from which the myoelectric signals are to be detected and the particular signal patterns used will depend on the type of amputation. If the degree-of-freedom to be duplicated is not significantly affected by the surrogate control muscles, the myoelectric signals will not be related to the desired movement. For example, the signals from the muscles in the shoulder girdle cannot be easily related to the movements of the hand with respect to the forearm. The muscles controlling the movements of the hand and the muscles in the shoulder girdle are innervated by separate nerves: thus, the hand can be moved with respect to the forearm without generating any appreciable forces and movements about the shoulder.

## Neuroelectric control

In cases which require several degrees-of-freedom and the muscles that are directly related to the desired movement are no longer present, an alternative approach, that of neuroelectric control, should be considered. Although the muscles normally controlling the desired functions may no longer be present, the peripheral nerves that contain the motorneurons for these muscles are accessible in the remaining part of the limb.

A specific case that would be well suited for neuroelectric control would be an above-elbow prosthesis. Three essential degrees-of-freedom of the upper limb to be duplicated are flexion-extension of the forearm, pronation-supination of the forearm, and opening-closing of the hand. Flexion of the forearm is governed by the musculocutaneous nerve and extension by the radial nerve. Pronation of the forearm is governed by the median nerve and supination by the radial and musculocutaneous nerves. Opening of the hand is governed by the radial nerve and closing of the hand by the median and ulnar nerves. These three nerves, the musculocutaneous, median and radial, carry sufficient and direct information relating to the three essential degrees-of-freedom.

It would be desirable to detect the neuroelectric signals associated with the distinct limb functions and employ them to control a multiple degrees-of-freedom prosthesis. The motion of a prosthesis activated in such a manner would be "volitional" and "natural"; the individual would merely desire a particular sequence of movements in order to spontaneously initiate a chain of events beginning in the higher centers of the central nervous system and ending with the appropriate movement of the prosthesis.

Neuroelectric control has a second advantage from the technical point of view. The neuroelectric signal has a bandwidth ranging from dc to approximately 7.5 kHz [24]. The

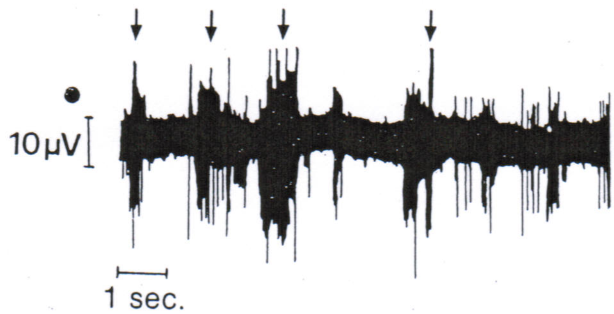


*Figure 2 The neuroelectric signal detected from the severed sciatic nerve while the rabbit was hopping. The main bursts indicated by the arrows were associated with distinct forward propulsion movements. The remainder of the activity was associated with non-specific movements of the limb containing the electrode unit. The signal was detected on the 29th day post implantation.*

frequency spectrum peaks at approximately 2 kHz. Therefore, the detected signal can be high-passed above 180 Hz, thereby removing the low-frequency electromagnetic interference associated with electric motors without losing a substantial amount of the signal. This is not the case with myoelectric control because the myoelectric signals recorded with surface electrodes have a power density spectrum that peaks at approximately 50 Hz.

The concept of neuroelectric control, however, introduces its own set of special problems. The first and most obvious is the development of an implantable recording electrode that may be attached to the nerve(s) without inducing severe degeneration of the nerve(s), while remaining capable of detecting the neuroelectric signal(s). The second problem is that the placement of the recording electrode requires surgical intervention. Thirdly, a preferable method of extracting the neuroelectric signals outside the body would be to transmit them. This requires the development of small, lightweight durable electronics devices (such as amplifiers, transmitters and receivers) that can be located near the nerve(s). Although the technology for miniaturized electronics is currently available, the difficulty of effectively encapsulating the implantable electronics packages beyond approximately three years persists. Alternatively, the signals could be carried outside the limb via a cable that terminates on the skin. This method, although feasible, requires a cumbersome connector protruding from the skin surface, presents the hazard of wire breakage, and can provide a pathway for bacteria.

Additional problems of improving the mechanical design and dynamics of a prosthesis with the available-power constraints as well as that of devising a suitable controller are common to all multiple degrees-of-freedom prostheses. These problems are not peculiar to the approach of neuroelectric control.

## Current developments

Of those problems peculiar to neuroelectric control, the primary concern is the development of a suitable electrode-nerve interface. A suitable recording electrode should have the following characteristics: a stable mechanical bond with the nerve to minimize the relative movement between the nerve and recording electrode; an acceptable signal-to-noise ratio; and the capability to disregard concurrent myoelectric signals from adjacent muscles. The design of such an electrode presents two basic problems. First, the electrode must be placed close to the nerve with minimal physiological restrictions and damage to the nerve. Second, all the materials used to construct the electrode must be biocompatible.

When considering the problem of designing an electrode capable of detecting neuroelectric signals from a severed nerve (as in the case of amputees), several possibilities arise. We chose to investigate a design that used a cuff electrode to record the neuroelectric signals from the surface of the nerve. Our motivations for this approach were two-fold. First, the effect of surgery was an important consideration: an electrode of this type requires a relatively simple implantation procedure. Second, neuroelectric signals detected from different locations around the perimeter of a nerve trunk might supply information concerning more than one limb function.

In a preliminary investigation [25] Lichtenberg and De Luca have shown that a cuff electrode containing four, pairs of wires arranged ninety degrees apart in a plane perpendicular to the longitudinal axis of a rabbit sciatic nerve can detect distinguishably different signal patterns. These patterns can be associated with functionally distinct groups of nerve fibers within the nerve. The experiments were performed by electrically stimulating the six branches of the sciatic nerve in sequence and recording the signal pattern at four points around the perimeter of the sciatic nerve. The signal amplitudes evoked by stimulation of the peroneal nerve branch (flexor nerve) were always distinguishable from those evoked by stimulation of the extensor nerve branches. In addition, one could distinguish between signal amplitudes evoked by stimulation of the different extensor nerves. Although it remains to be demonstrated that voluntarily elicited neuroelectric signals associated with specific functions of a limb can be distinguished, these results indicate that further investigations in this direction may prove useful.

We proceeded to design a recording electrode that could be attached to a severed nerve. The original version was initially reported in March, 1974 [26] and was later described in detail by De Luca and Gilmore [27]. It had five constituent parts: the recording contacts, consisting of 90% platinum—10% iridium wires, a knitted Dacron cloth tube to support the recording contacts, a Silastic cover to insulate the recording contacts from the myoelectric signals, a cable, and a transcutaneous device that passed the wires through the skin.

The most crucial parts of the recording electrode were the cloth tube to be placed around the severed nerve and the Silastic cover to be placed on top of the cloth tube. An investigation [28] was performed to examine the effects of various tube parameters on the enclosed nerve and its immediate surrounding tissue. A histological analysis revealed that the presence of the cloth tube aggravates the degenerative process that may occur when a nerve is severed. Degeneration of the enclosed nerve was greatest at the point of severance and diminished in the proximal direction. A minimal space of approximately 0.2 mm between nerve and tube was necessary for at least one-half of the nerve fibers surrounded by the proximal one-third of the cloth tube to retain intact myelin sheaths. Additional space did not improve the results. The type of material (Teflon or Dacron) used to construct the cloth tubes, and the presence or absence of a Silastic cover around the cloth tube appeared to have no significant effect on the enclosed nerve. These results were also valid for the longest implant which lasted 681 days.

The space between the tube and the nerve fibers became filled with fatty tissue and connective tissue. The internal connective tissue was continuous (through the interstitial spaces of the cloth) with the connective tissue which developed outside the tube. Furthermore, the external connective tissue arrangement stabilised the enclosed part of the nerve to the surrounding tube. This is a requirement for obtaining faithful and consistent neuroelectric signals.

Based on the above information, the recording electrode underwent a series of improvements [29,30] resulting in the current version [24]. In the current design, the recording electrode was formed from six Teflon-coated stranded wires (Medwire 10 1R 9/49T) helically wound around 2-0 surgical silk suture. The entire cable was enclosed in medical grade Silastic. The six wires in the cable formed three pairs of electrode contacts (three channels) which were located on the inner surface of the cloth tube. Approximately five milli-

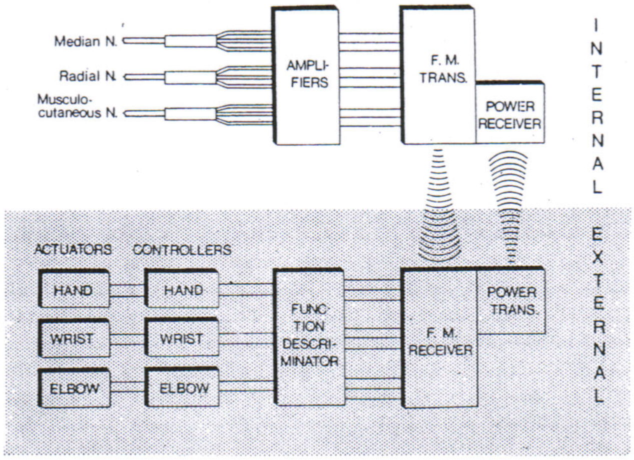


*Figure 3 A block diagram of a neuroelectric control scheme for duplicating the principal degrees-of-freedom: flexion-extension of the elbow, rotation of the wrist, and opening and closing of the hand. The neuroelectric signals are amplified and frequency modulated within the stump and then transmitted externally. The implanted devices are powered by an external radio-frequency transmitter. A function-discrimination device amplifies the signals and selects the appropriate controller(s) for the particular set of neuroelectric signals that are detected. The controller(s) activate the appropriate actuator(s) to duplicate the desired movement.*

meters of the Teflon insulation was removed from the distal end of each wire so that the exposed surface formed the electrode contact area. The proximal ends of the six wires in the cable were soldered to a seven-pin connector (Microtech Model ER7 5-4H) cemented with medical grade Epoxy in the center portion of a bio-carbon transcutaneous device from Bentley Laboratories, Inc. The underside of the bio-carbon device was covered with a thin coat of Silastic adhesive. A 28 mm long tube constructed from knitted Dacron fabric having a thickness of 0.2 mm and a water porosity approximately 10,000 ml/min/cm² (at 1 atmosphere) formed the structural component of the electrode unit. The inside diameter of the tube was 3.5 mm which was approximately 0.6 mm greater than the average diameter of the sciatic nerve in the popliteal fossa region of the rabbits that were used. The fabric was cut to size and the three pairs of electrode contacts from the cable were woven into the cloth and bonded 0.5 mm above the surface of the fabric with Silastic adhesive. This arrangement placed the exposed surfaces of the electrodes in intimate contact with the surface of the nerve. The proximal and distal electrode contacts of each pair were located five millimeters from the respective ends of the tube. This arrangement positioned the recording contacts over the nerve portion in which at least 50% of its nerve fibers have been found to retain structural integrity after axotomy. The electrode pairs were spaced so that they were situated 120 degrees apart on the inner surface of the finished tube. The fabric patch with the attached electrode contacts was then wrapped around an appropriately sized mandrel (3.5 mm) and the mating edges of the cloth were bonded together with Silastic adhesive, forming a tube. The silk suture strain relief was secured to the cloth tube.

The Silastic cover consisted of a thin-walled tube molded from Silastic 382 medical grade elastomer. The cover was sealed at the distal end. The inside diameter of the cover was three millimeters greater than the outside diameter of the cloth tube.

A recording electrode unit was implanted around each of the severed left sciatic nerves of six rabbits. Motor neuroelectric signals of physiological origin as well as electrically evoked neuroelectric signals were detected from the surface of the severed nerve. Neuroelectric signals recorded while the rabbit was hopping indicated that the amplitude of the signal can be modulated, as can be seen in Figure 2. This behaviour is similar to that seen in the myoelectric signal. It may be possible that neuroelectric signals can be implemented in a similar fashion as myoelectric signals for the purpose of controlling prostheses.

The results showed a definite chronological increase in the time duration that the neuroelectric signals could be recorded, indicating that the results were related to the experience gained in performing the implantations. To date, the longest time that a motor signal has been successfully recorded is 142 days. Analysis of the time dependent behaviour of the amplitude of detected neuroelectric signals and the impedance of the electrode unit was performed, but even in conjunction with histological observations no adequate explanation of the limited duration of successful recording could be found.

We have yet to distinguish the effects of the animal's disposition and experimental artifacts from the physiological effects of the recording electrode and the trauma of the implementation procedure. Experiments that will attempt to answer some of these questions are planned.

The limited information available from current studies suggests that the nerve-electrode interface may provide the necessary man-machine interface for controlling the forward path of prostheses control. It should be emphasized that the results described in this section investigate only one possible approach. Alternative approaches should also be considered.

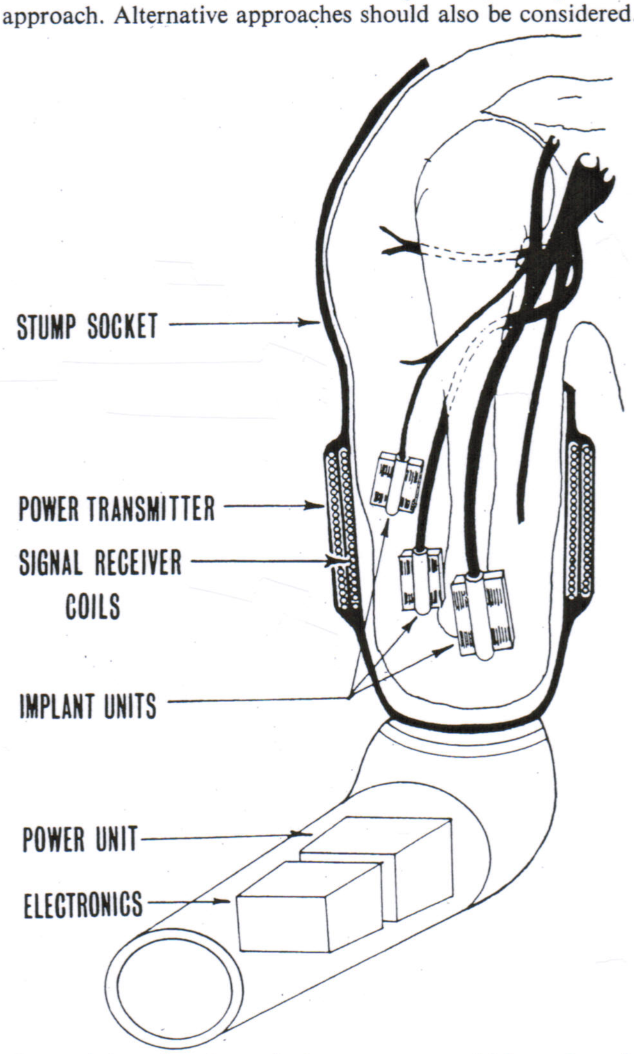


*Figure 4 A preferred prosthesis arrangement which employs the devices illustrated in Figure 2. Each implant unit consists of the recording electrode flanked by two capsules: one capsule contains amplifiers and transmitters; the other contains the radio-frequency power receiver. The coils for the power transmitter and the FM receiver are encased in the wall of the socket. The external electronics and the power unit are contained in the prosthesis.*

## A possible scheme for neuroelectric control

With the development of an acceptable nerve-electrode interface, the scheme presented in Figure 3 would provide a possible arrangement for employing the neuroelectric signal to control an above-elbow prosthesis. In this particular case, the principal functions to be duplicated are flexion-extension of the elbow, rotation of the wrist and opening and closing of the hand. The block diagram is separated into those devices which are to be implanted and those that will be located externally.

The most likely situation is that three electrode-nerve interfaces will be required. The neuroelectric signal(s) must then be amplified and FM modulated prior to transmitting them externally. The power required by the amplifier(s) and transmitter is supplied via an implanted power receiver powered by an external radio-frequency source. The transmitted neuroelectric signal(s) are demodulated by an external receiver. The demodulated signals are fed to a function-discrimination device that determines which of the available controllers are activated according to the particular signal pattern present at any given time. The controllers then drive the actuators to produce the required prosthesis movement.

Figure 4 presents a conceivable physical implementation of the devices in Figure 3. Each implanted unit has two capsules, one on either side of the recording electrode. One of the capsules contains the amplifiers and transmitter(s), the other contains the radio-frequency power receiver. Two coils are located in the wall of the stump socket. One coil is connected to the external radio-frequency power transmitter; the other receives the transmitted neuroelectric signal(s) and is connected to the external FM receiver. This arrangement should provide a convenient and durable implementation of neuroelectrically controlled prostheses for amputees.

## Concluding remarks

Although there is as yet no definitive evidence that neuroelectric signals can replace myoelectric signals as forward-path control signals, pioneering investigations indicate that the approach is technically feasible. The concept of neuroelectric control may be used either independently or in conjunction with myoelectric control. In either case, neuroelectric control should provide a significant improvement in the performance of prostheses and, therefore, deserves consideration in the broad realm of prostheses design.

The concept of transferring the neuroelectric signal outside the body presents exciting possibilities for controlling a wide variety of devices in our environment. It is not restricted to prostheses control. Its future applications are limited only by the thoughts of those who make it their concern to explore the concept.

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