

Durability of implanted electrodes and leads in an upper-limb neuroprosthesis

Kevin L. Kilgore, PhD; P. Hunter Peckham, PhD; Michael W. Keith, MD; Fred W. Montague, MS; Ronald L. Hart, MS; Martha M. Gazdik, BS; Anne M. Bryden, OTR/L; Scott A. Snyder, PhD; Thomas G. Stage, BS

Department of Orthopaedics, MetroHealth Medical Center, Cleveland, OH; Department of Biomedical Engineering, Case Western Reserve University, Cleveland, OH; Department of Veterans Affairs, Rehabilitation Research and Development Service, Cleveland, OH

Abstract—Implanted neuroprosthetic systems have been successfully used to provide upper-limb function for over 16 years. A critical aspect of these implanted systems is the safety, stability, and reliability of the stimulating electrodes and leads. These components are (1) the stimulating electrode itself, (2) the electrode lead, and (3) the lead-to-device connector. A failure in any of these components causes the direct loss of the capability to activate a muscle consistently, usually resulting in a decrement in the function provided by the neuroprosthesis. Our results indicate that the electrode, lead, and connector system are extremely durable. We analyzed 238 electrodes that have been implanted as part of an upper-limb neuroprosthesis. Each electrode had been implanted at least 3 years, with a maximum implantation time of over 16 years. Only three electrode-lead failures and one electrode infection occurred, for a survival rate of almost 99 percent. Electrode threshold measurements indicate that the electrode response is stable over time, with no evidence of electrode migration or continual encapsulation in any of the electrodes studied. These results have an impact on the design of implantable neuroprosthetic systems. The electrode-lead component of these systems should no longer be considered a weak technological link.

Key words: electrodes, functional electrical stimulation, implants, neuroprosthesis, spinal cord injury, survival analysis.

INTRODUCTION

Implanted neuroprosthetic systems have been successfully used to provide motor function in spinal cord injury (SCI). To date, the majority of these systems have used multiple muscle-based electrodes and a central implanted pulse generator located in the torso. The design of these systems, therefore, puts stringent requirements on the electrode, lead, and connector assembly [1].

Abbreviations: AbPB = abductor pollicis brevis, EDC = extensor digitorum communis, EIP = extensor indicis proprius, EPL = extensor pollicis longus, FDP = flexor digitorum profundus.

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Address all correspondence and requests for reprints to Kevin L. Kilgore, PhD; MetroHealth Medical Center, Hamann 601, 2500 MetroHealth Drive, Cleveland, OH 44109; 216-778-3480; klk4@cwru.edu.

Although electrodes and leads are common components of implanted systems, such as pacemakers, cochlear implants, respiratory assist devices and bladder/bowel stimulators, these systems do not place the same requirements on the electrode and lead as motor neuroprotheses. First, multiple leads are typically required, because multiple muscles are activated and electrodes are distributed over a relatively large area of the body. Second, the leads are required to cross many joints so that electrodes reach some of the more distal muscles. Third, fine control over the degree of contraction is required to produce highly coordinated movements.

In this paper, we report retrospectively on the reliability and stability of electrode, lead, and connector systems used in a neuroprosthesis providing upper-limb function [1–4]. These systems used 8 or 10 functioning electrodes and required some leads to cross the shoulder, elbow, and wrist joints to stimulate muscles that must be activated in a graded fashion. The clinical performance and acceptance of first- and second-generation upper-limb neuroprotheses have been reported previously [3,4]. These systems provide a stringent test for implanted electrode-lead systems.

METHODS

Neuroprosthetic System Description

The implanted neuroprosthetic system used in this study consisted of an implanted stimulator, electrodes, leads, and connectors, as well as external components, as shown in **Figure 1**. A radio frequency inductive link was used to communicate with, and power, the implanted device. The implanted stimulator had 8 or 10 leads connected to stimulating electrodes. An in-line connection was used to connect the implant device to each lead intra-operatively. Each electrode was implanted in a separate location, typically in the muscles of the hand, forearm, and upper arm. In some cases, an electrode was placed in the supraclavicular region to provide sensory feedback.

Muscle activation was accomplished through electrical stimulation. Stimulus pulses consisted of a constant current balanced charge waveform. The stimulating phase was a square cathodic pulse of 0 to 200 μ s in duration and 20 mA in amplitude. In infrequent cases, a stimulation amplitude of less than 20 mA was used. Stimulus frequency was 12 to 16 Hz for all muscle-based

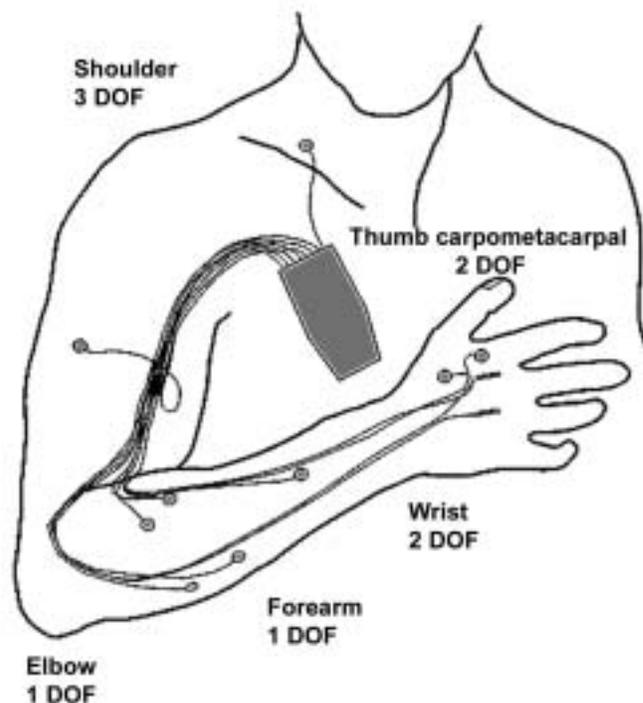


Figure 1.

Implanted upper-limb neuroprosthesis used in this study. Implanted stimulator, leads, connectors, and electrodes are shown. Stimulator device is located in pectoral region. Single-line interconnects are located in mid-humeral region. Leads to electrodes placed in thenar intrinsic muscles must pass across joints with a total of 9 degrees of freedom (DOF).

electrodes. For sensory feedback, frequency varied from 4 to 60 Hz.

Electrode-Lead Design

Stimulus output was delivered through lead wires that traveled subcutaneously to the distal electrode termination. Each lead had an in-line connector. Two types of electrodes have been used. These components are shown in **Figure 2** and are described in the following paragraphs.

Lead

The lead cable consisted of seven Type-316LVM stainless steel wires (each was 0.034 mm in diameter, Fort Wayne Metals, Fort Wayne, Indiana), organized into a single seven-filament strand and insulated with PFA-Teflon™ (Dupont, Wilmington, Delaware) by Temp-Flex (S. Grafton, Massachusetts). The lead was fabricated by winding two lengths of cable in tandem, forming a double helix of two conductors electrically insulated along their

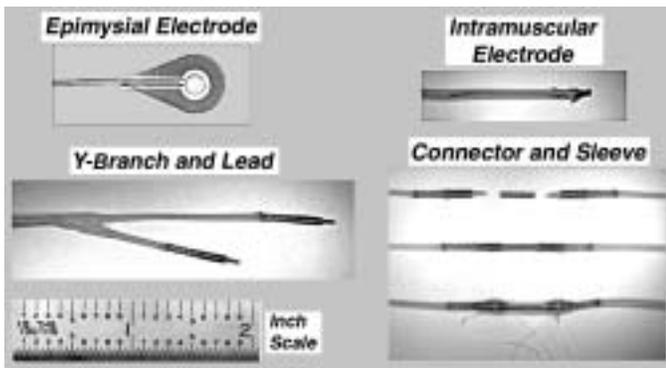


Figure 2. Electrode, lead, and connector components analyzed in this study. All components are shown on same scale.

length but shorted to each other at the electrode and connector end. The coiled lead was placed inside medical-grade SilasticTM (Dow Corning, Midland, Michigan) tubing. The lead outer diameter was approximately 1.3 mm.

Connector

A single-conductor in-line connector was used to independently connect each electrode to the implanted stimulator. The connector consisted of two male plugs that mate with a center spring and were enclosed by an external insulating cuff. The male plug and the associated strain relief spring were integral parts of the terminal end of both the implant lead and the electrode lead. The connector measured 30 mm long from end to end of the strain relief springs. The outside diameter measured 3.5 mm at the suture cuffs (widest point) [5].

Electrodes

Implanted electrodes were used to direct the stimulus current so that it activated the desired neural structures. Two types of electrodes were used in upper-limb implanted systems during this study. They were epimysial and intramuscular electrodes [6,7].

Epimysial Electrode. This electrode was placed on the muscle surface. The electrode was a Pt-Ir disk embedded in a silastic backing. The lead wire was welded to the back of the disk with a resistance welder in an inert gas environment. The silastic backing was reinforced with Dacron. The electrode was sewn onto the muscle epimysium with sutures through the Dacron backing. With 4-0 Dacron suture, five sutures were placed around the

perimeter of the electrode backing, each tied with four knots. The diameter of the stimulating disk was 5 mm.

Intramuscular Electrode. This electrode was inserted into the muscle belly. The electrode consisted of 316LVM stainless steel wire coiled around the outside of the lead tubing. The stimulating surface was 2 mm long and had an approximate surface area of 14.5 mm². A 2 mm-long polypropylene barbed anchor on the tip of the electrode maintained the position of the electrode in soft tissue. The electrode was inserted into the muscle with a probe and cannula system, as described by Memberg et al. [7].

Y-Branch

The total number of leads coming from the implant can be reduced by taking advantage of the two separate conductors in each lead wire [8]. A Y-branch termination was constructed that terminated in two separate male plugs to obtain separate access to each conductor. The individual cables were split during fabrication to form a Y-junction or Y-branch. Each single conductor cable was formed into a separate lead wire with the use of the same fabrication methods used for the two conductor leads. This Y-branch had no splice or connection between conductors of the tandem and single helix leads. The Y-branch was reinforced with a molded silastic strain relief and was only used in devices with 10 electrodes.

Subjects

Data were collected from a single clinical series conducted at two sites: the Louis Stokes Department of Veterans Affairs (VA) Medical Center and MetroHealth Medical Center, both in Cleveland, Ohio. All subjects had sustained a traumatic SCI and were tetraplegic at the C5 or C6 level. All subjects were implanted by one surgeon (MWK), all procedures and technology received appropriate Institutional Review Board (IRB) and Food and Drug Administration (FDA) approvals prior to usage, and all subjects gave appropriate written consent.

Preoperative Preparation

Before surgery, each subject underwent a period of muscle conditioning using surface stimulation for at least 1 month [2]. This produced stronger, more fatigue-resistant muscles in preparation for intraoperative placement of electrodes.

Implantation Procedure

The implant device, leads, and electrodes were placed during an operative procedure with the patient under general anesthesia, as described in detail elsewhere [1]. Epimysial electrodes were sutured on the muscles in an open surgical procedure. Intramuscular electrodes were inserted into the muscle belly with the use of a probe and cannula. Leads were tunneled from the electrodes up the arm to a connector in the mid-humeral area. Tunneling was accomplished with a blunt plastic probe that was 6 mm in diameter and approximately 30 cm long (Scanlan tunneler, Scanlan International, St. Paul, Minnesota). Suction was placed on one end of the tunneler, and the leads were fed into the opposite end. Sterile saline was used to cause the leads to be drawn into the tunneler. When the leads were completely drawn into the tunneler, it was withdrawn, leaving the leads in place. Up to 10 leads can be passed through a single tunneler with this method.

The stimulator unit was implanted in the pectoral region with the leads tunneled subcutaneously to the humeral connector site, where the spring connector was used to connect the lead from the stimulator unit to the lead from the electrode. Where leads crossed joints, the surgeon routed the lead near the neutral axis whenever possible to try to minimize the stress on the leads. Postoperatively, the patient was placed in a long arm cast for 3 weeks for electrode stabilization. Elective modifications to the postoperative care were made if the patient received additional surgical alterations to the upper limb, such as tendon transfers, with the result that some patients remained in a cast for as long as 4 weeks. After cast removal, muscle conditioning with the use of the neuroprosthesis was initiated at a low level for the muscle strength to be rebuilt. This condition typically continued for at least a month before the patient was trained to use the system for functional activities. Patients underwent 1 to 3 weeks of rehabilitation training, after which they were released home to use the device for functional activities.

Monitoring of System Integrity

We monitored the implanted neuroprosthetic system to verify the biological safety and integrity of the implanted components. The mechanical and electrical integrity of the implanted components was also monitored. These tests included intraoperative verification, X rays, electrode thresholds, surface potential measurements, and patient and staff reports of technical and medical incidents.

Intraoperative Verification

Prior to final wound closure, the entire system was tested with a sterile radio frequency (RF) link to the implanted stimulator. The surgeon tested each electrode individually to ensure that the proper response was obtained by visual observation of joint movement.

X rays

X rays of the entire upper limb and shoulder were taken at the time of surgery. Subsequently, if an unexplained change in electrode response was identified by the subject or investigator, the images were repeated. Leads, electrodes, and connectors were easily identified on a standard X ray, but only gross separation of wires could be identified with this method (i.e., the fractured ends need to be separated by at least 1 mm).

Electrode Thresholds

The electrode threshold was defined as the lowest stimulus level at which a visible response was obtained [9]. In all cases, the stimulus pulse amplitude was 20 mA and the frequency was 12 Hz. We recorded thresholds by slowly increasing the stimulus pulse duration in 1 μ s increments to each electrode individually while observing each upper-limb joint for movement. Thresholds were recorded at 1 to 2 months postimplant and at 6 and 12 months postimplant. Thresholds were repeated any time the subject reported a possible change in grasp response.

Surface Potential Measurements

The presence of an electrical stimulus from an implanted system can be accurately recorded with the use of surface electrodes on the skin [10]. A reference electrode was placed over the implant capsule where the anode was located. The sensing electrode was placed at variable distances along the arm (typically 10 cm intervals). A voltage potential was recorded during the delivery of the stimulus. The voltage increased as a function of the distance between the two electrodes and could be as high as 6 V. A broken lead was confirmed by changes in the expected surface potential map and by changes in the stimulus waveform recorded on the skin surface [10].

Medical and System Incident Reports

Physiological and technical incidents were recorded and evaluated as reported by each subject. Subjects were regularly contacted and queried regarding incidents during

their first 2 years postoperative, and followed at approximately yearly intervals thereafter. Subjects were instructed to contact the research staff before having surgery to allow antibiotics to be prescribed.

Statistical Methods

Threshold Stability

Electrode threshold measurements were recorded at irregular intervals for subjects once they were beyond 1-year postimplant. To determine whether some electrodes demonstrated a long-term increase in threshold as a function of time, we examined each electrode that met the following criteria:

1. Electrode threshold recorded at least 60 days postimplant.
2. At least four threshold values, each recorded on different days.
3. Threshold data must span at least a 2-year interval.

A line was matched to the threshold data points with the use of a linear least squares estimation. We evaluated the slope of the estimated line to determine if it was statistically significantly higher than zero at $p = 0.05$. Where the slope was statistically higher than zero, the projected increase in threshold over 50 years was estimated. If the threshold increase over that period was projected to be less than $50 \mu\text{s}$ (i.e., slope $< 1 \mu\text{s}/\text{yr}$), the slope was determined to be insignificant clinically. A threshold increase of this rate would not be functionally noticeable to the user and, in the worst case, might require reprogramming of the neuroprosthesis approximately every 20 years.

Electrode data during the first 60 days postimplant were not considered in this analysis because electrode thresholds undergo a settling period during that time. Typically thresholds are slightly higher immediately postoperative and then settle to a consistent value at 2 to 3 months postimplant [1].

Correlation of Intraoperative and Postoperative Electrode-Muscle Output Characteristics

It is desirable to determine if the electrode-muscle output characteristics (i.e., the muscle forces as a function of stimulus level) that are observed intraoperatively correlate to the postoperative output characteristics [9,11,12]. A poor correlation would result in many electrodes needing to be repositioned surgically. However, obtaining direct measurements of the electrode-muscle output characteristics intraoperatively is difficult because

quantitative measurements of muscle force and joint position are time-consuming and instrumentation-intensive. Therefore, we have established four easily recorded criteria to estimate this correlation. These are—

1. Accurate recruitment of targeted muscle. The postoperative electrode response should demonstrate that the first muscle recruited by the electrode is the muscle that was targeted intraoperatively.
2. Isolated response. Stimulus delivered to the electrode should only recruit the muscle or muscle groups that were targeted intraoperatively. If the electrode recruited a second muscle within a pulse duration change of $2 \mu\text{s}$ from the target muscle, then it was defined as nonisolated [11].
3. Low threshold. The threshold should be below $50 \mu\text{s}$.
4. Electrode moved or replaced. An electrode that was moved (surgically) or replaced indicates a possible failure of identifying the desired response intraoperatively.

Survival Analysis

A Kaplan-Meier analysis was used [13], which enables data from surviving and failed electrodes to statistically predict the longevity of a population. The proportion surviving is equal to the geometric sum of one minus the ratio of the number of failures to the number of electrodes at risk for failure. The predicted surviving proportion is constant between points of failure. Therefore, a meaningful confidence interval for survival can only be calculated at points of failure. The hazard rate was also calculated at each failure point, with the use of the formula of number at risk divided by the interval between failures.

RESULTS

Between August 1986 and December 1999, 28 arms in 27 patients received the upper-limb neuroprostheses. Twelve arms had C5 motor function and the remaining sixteen arms had C6 motor function. Patients were at a median age of 32 years at the time of implantation (range 21 to 47 years). The active use of the neuroprosthesis varied considerably among the patients, from regular daily use to occasional use [3,14]; therefore, the total stimulation time experienced by each electrode varied from approximately 1,000 hours to over 50,000 hours. Overall, 238 electrodes were implanted, with an average follow-up time of 7.1 years (range: 3.2 to 16.4 yr). There were 204 epimysial electrodes and 34 intramuscular electrodes.

Intramuscular electrodes were not introduced until the tenth subject in the series (1995) and were not used extensively until 1997. One subject has nine intramuscular electrodes and one epimysial electrode.

Electrodes were grouped into various “regions” of the body, depending upon the location of the motor point or skin area (for sensory electrode) of the muscle to be excited. The electrode locations are shown in **Table 1**. Total lead length (from stimulator package to electrode termination) varied from 28 to 83 cm and depended on electrode placement, lead routing, and subject size.

No cases have been reported where failure of a component of the neuroprosthesis resulted in the inability of the subject to use the neuroprosthesis for functional activities. Of the 238 electrodes in the series, 234 (98.3%) remained intact throughout the study. Three (1.3%) were broken and one (0.4%) was infected. These electrodes are discussed in detail subsequently. Three subjects died during the study at 3.3, 6.0, and 9.4 years postimplantation (representing 25 electrodes). Therefore, 209 electrodes continue to be used in functioning neuroprosthetic systems.

Survival analysis using Kaplan-Meier showed that there was a 98.7 percent probability for an electrode to be intact at 16 years, as shown in **Figure 3**. At the latest failure point (1.9 years), the 95 percent confidence interval for the survival probability is 97.3 to 100.0 percent. The hazard rate at each of the three failure points ranged from 0.6 percent a year to 0.8 percent a year. At present, 31 electrodes are older than 10.0 years, 102 electrodes older than 7.5 years, and 191 electrodes older than 5.0 years.

Correlation of Intraoperative and Postoperative Electrode-Muscle Output Characteristics

Less than 1.5 percent of the electrodes exhibited a lower threshold for a muscle that was not the targeted muscle intraoperatively. An additional 2 percent of the electrodes did not have an isolated response (second muscle

Table 1.

Electrode locations and lead length.

Location	Distribution (%)	Average Total Lead Length (cm)
Chest	6	32
Upper Arm	4	38
Forearm	64	55
Hand	26	75

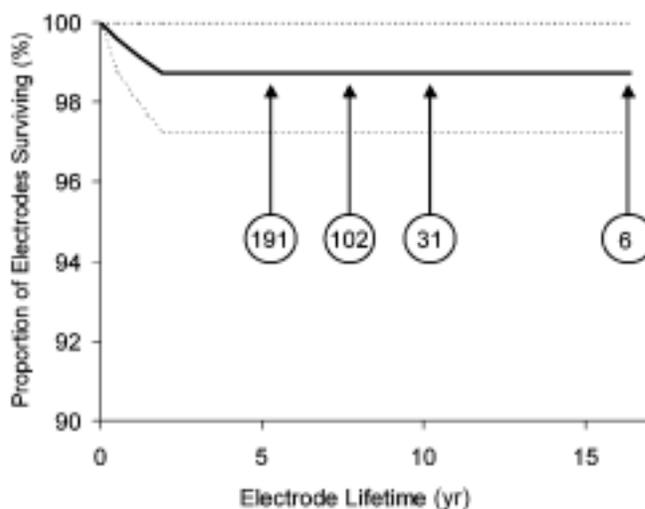


Figure 3.

Kaplan-Meier survival analysis for all electrodes and leads. There is a 98.7% probability of electrode survival to 16 years. Dotted lines indicate 95% confidence interval. Numbers in circles indicate number of electrodes currently surviving at 5.0, 7.5, 10.0, and 15.0 years, respectively.

recruited within 2 μ s of the first muscle) and another 2 percent had thresholds higher than 50 s. Therefore, 94.5 percent of the electrodes met our criteria for correlation between the targeted response in surgery and the response achieved after surgery. Many of the electrodes with a less than a desirable response were those placed in weak, partially denervated muscles.

Three epimysial electrodes have been moved as part of a subsequent surgical procedure. In two cases, the surgeon repositioned the electrode on the muscle in an attempt to obtain better recruitment properties. One electrode was placed on the extensor digitorum communis (EDC) and was moved after 2 years. The difference in response was minimal. The second electrode was placed on the flexor digitorum profundus (FDP) and was moved after 10 months. The surgeon moved a third electrode, which was originally implanted on the pronator quadratus muscle, to the ulnar nerve near the flexor carpi ulnaris tendon to obtain a new function. In all cases, after the electrode was moved, the patient was recasted for 3 weeks. No other incidents have occurred with these electrodes (at least 6 years of follow-up).

Functional Stability

Stability of physiological responses obtained from the electrodes was also excellent. Of all 238 electrodes, the

average pulse width for threshold was $15\mu\text{s}$ (range of $2\mu\text{s}$ to $82\mu\text{s}$). The threshold depended on the implanted muscle, as shown in **Table 2**. Larger muscles, such as the FDP and triceps, tended to have higher thresholds.

A more extensive analysis was performed on 81 electrodes from 11 subjects that had at least four recorded threshold values spanning at least 2 years, with the first data point being at least 60 days postimplantation. We used a linear fit to identify whether the thresholds showed a trend that was significantly different than zero. Seventy-six of these eighty-one electrodes (93.8%) had slopes that were not significantly different from zero. Of the remaining five electrodes, all had an estimated slope of less than $0.8\mu\text{s}$ a year, which was well below the rate that was considered clinically significant. Therefore, 100 percent of the electrodes demonstrated long-term stability in their threshold responses.

Thresholds for the first group of electrodes implanted have been recorded regularly over the course of 16 years [1]. These data are shown in **Figure 4**. There is session-to-session variability in the observed threshold, but the estimated slope for each electrode is less than $0.74\mu\text{s}$ a year.

Surface potential mapping was recorded regularly in the first three subjects, but the response was determined to be quite predictable and therefore only needed to be measured if a suspected lead fractured. An example of a typical surface potential map is shown in **Figure 5**.

Table 2.
Electrode thresholds for upper-limb muscles.

Target Muscle or Nerve	n	Average Threshold (μs)	Standard Deviation (μs)
Ulnar Nerve	7	4	2
Third Dorsal Interosseous	5	10	4
Adductor Pollicis	24	10	10
Extensor Pollicis Longus	27	11	6
Abductor Pollicis Brevis	28	12	9
Extensor Carpi Ulnaris	9	14	8
Extensor Digitorum Communis	24	14	11
Second Dorsal Interosseous	6	16	12
Flexor Pollicis Longus	20	17	15
Flexor Digitorum Profundus	22	21	21
Triceps	9	21	22
Flexor Digitorum Superficialis	29	25	18

n = number of electrodes implanted

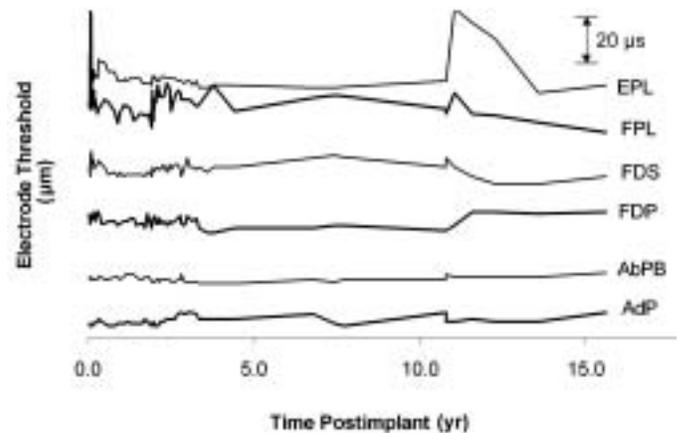


Figure 4. Epimysial electrode threshold measurements for subject 1J, first subject to receive implanted upper-limb hand system. Electrodes are located on the following muscles: abductor pollicis brevis (AbPB), adductor pollicis (AdP), flexor digitorum profundus (FDP), flexor digitorum superficialis (FDS), flexor pollicis longus (FPL), and extensor pollicis longus (EPL). Each electrode has been used daily for past 16 years. Average thresholds vary from 5 s (AbPB) to 21 s (FPL). Note that baseline for each electrode has been shifted in plot for clarity. Cause of transient increase in threshold for EPL at 11 years is unknown.

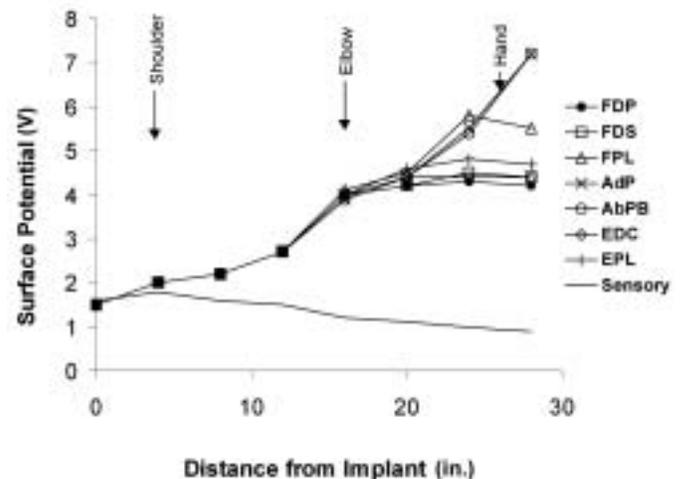


Figure 5. Typical surface potential map. Surface potential reaches a maximum over each respective electrode termination. Electrodes located in hand are abductor pollicis brevis (AbPB) and adductor pollicis (AdP). Electrodes located in forearm are flexor digitorum profundus (FDP), flexor digitorum superficialis (FDS), flexor pollicis longus (FPL), extensor digitorum communis (EDC), and extensor pollicis longus (EPL). Sensory electrode is located in chest above clavicle.

Mechanical Durability

Across all 238 electrodes, only three electrode-lead mechanical failures occurred. Each of these incidents occurred within 2 years postimplant. We have identified that repeated use is the most likely cause of failure in only one of the three breakages. No failures of the lead occurred more than a few centimeters proximal to the electrode tip, and no failures or separations of the connectors occurred. None of the mechanical failures appear to have been the result of the lead being flexed across the joint, because no failures occurred near the joints.

Failed Electrode—Case 1

In subject 2K, the epimysial electrode used for sensory feedback broke secondary to the implant device rotating within the body. The rotation of the device caused the sensory lead to become wound around the remaining leads. The lead eventually pulled apart in tension and broke near the electrode termination. At the time of repair, we determined that the implant had made 16 complete revolutions within the body. The implant was unwound and we determined that all the leads from the implant device were intact and had not been damaged. The distal lead of the sensory electrode was disconnected at the connector site and a new sensory electrode implanted in the same location. This device has now been operational for 10 years without further incident. Because of this early incident, the surgeon now sutures the implant device in place to prevent rotation or migration within the body, and no additional incidents have occurred.

Failed Electrode—Case 2

In subject 1W, an epimysial electrode placed on the abductor pollicis brevis (AbPB) muscle broke after 2 years of regular use. The break was identified through routine electrode threshold measurements and verified with surface potential measurements. The surface potentials for the AbPB electrode, as compared to the nearby AdP electrode, are shown in **Figure 6**. The surface potentials indicated that the breakage did not include the lead insulation and/or that the break was near the electrode tip. The electrode was surgically removed and examined. The break was in the lead wire just proximal to the electrode disk. We suspect that this fracture occurred from repeated pressure on the thenar eminence. This patient pushed a manual wheelchair and used the palm of his hand against the rims of the wheelchair to accomplish propulsion, undoubtedly placing repeated high stresses directly on

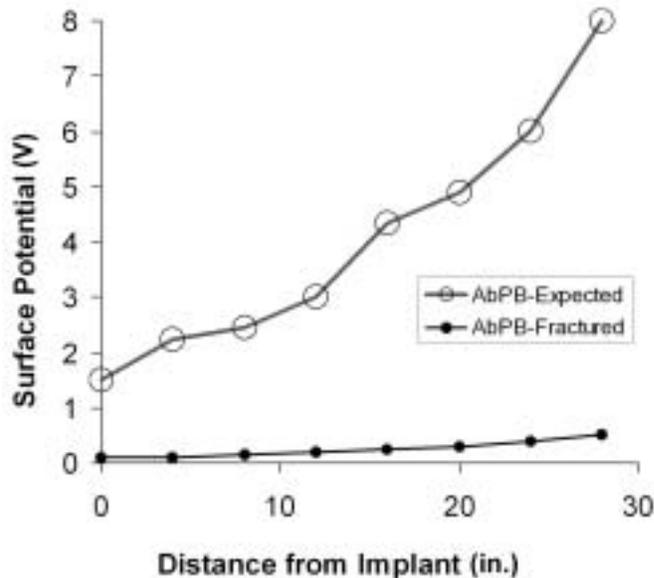


Figure 6.

Surface potential map for failed abductor pollicis brevis (AbPB) electrode in subject 1W. Expected AbPB surface potential map is based on surface potential map of adductor pollicis electrode for subject 1W. Lead wire fractured at electrode termination, but insulation remained intact. However, this type of failure is easily identified by significant reduction in surface potential recording when compared to expected values. Maximum surface potential voltage was recorded at most distal site, indicating that stimulus current was escaping at or near electrode termination.

the electrode. Because of this incident, we now place the AbPB electrode in a more medial location so that it is more protected from pressure to the palm.

Failed Electrode—Case 3

In subject 2C, an intramuscular electrode placed in the second dorsal interosseous muscle developed a high threshold and an altered response. This electrode had functioned normally for 442 days. The electrode exhibited an elevated threshold, and the surface potential measurements indicated that the resistance of the lead/electrode or electrode/tissue interface had increased. We determined the latter by examining the stimulus pulse as recorded from the skin surface, as shown in **Figure 7**. Under normal conditions, the constant-current stimulus pulse delivered by the implant produces a falling edge that gradually reaches a plateau after a few microseconds. When the electrode impedance exceeds the capacity of the stimulator to maintain a constant current pulse, the recorded waveform changes. The falling edge exhibits an overshoot before reaching a plateau. The increased

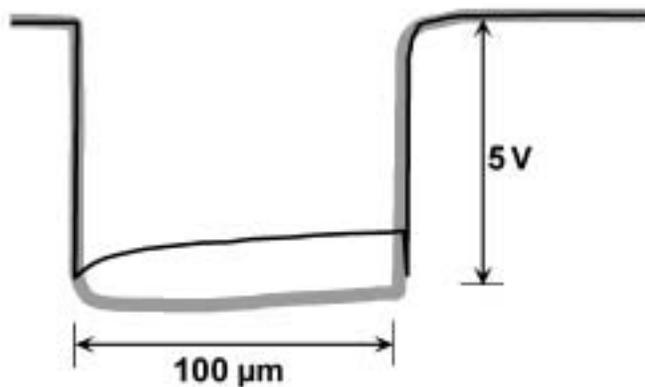


Figure 7.

Recording surface potential for a single stimulus pulse of 20 mA amplitude and 100 s duration. Gray thick line shows typical waveform for a constant current pulse. Black line shows waveform recorded for broken electrode in subject 2C. Waveform indicated that current was no longer maintained because of increased electrode-lead resistance.

threshold and impedance indicated a broken electrode. The electrode was removed during an unrelated surgical procedure and replaced with a new electrode in the same muscle. Examination of the lead uncovered a region of fracture that was approximately 10 cm proximal to the tip of the electrode. The silastic insulation was intact.

We could not determine the cause of the fracture. The subject underwent a surgical procedure near the location of the lead fracture just before the electrode exhibited the altered response. In addition, the subject had accidentally hit his hand on a metal door just before the altered response, causing bruising. The fracture mechanism is consistent with either cause. Based on our analysis, it is unlikely that the fracture occurred secondary to flexion fatigue.

Tissue Response

In a few cases, when an electrode has been replaced or moved, there has been the opportunity to observe the encapsulation response of the body to the electrodes. The encapsulation is similar to what we have observed in the dog model [10], consisting of a 0.5 to 1.0 mm-thick translucent encapsulation that securely holds the electrodes in place. The sutures are still visible and tied.

Infection/Rejection

A single incident of a localized infection at an electrode occurred. In subject 1N, a localized infection developed at the site of the sensory electrode termination near

the mid-clavicle. The infection appears to have been initiated by a suture in the incision site. To ensure that the infection would not track up the electrode lead, we cut the distal lead proximal to the infection site and removed the distal end of the electrode. The infection resolved without further incident. The sensory electrode was not replaced in this subject.

DISCUSSION

This study presents data demonstrating the long-term viability of implanted electrodes, leads, and connectors for use in upper-limb neuroprostheses. Electrode and lead failures or infections have been anticipated to be significant causes of failure in motor neuroprostheses for limb control [15–21], but our results demonstrate that the total failure incidence for any reason was less than 2 percent. This incidence can be reduced by careful electrode placement. The results further indicate that the electrodes remain in the location in which they are placed and that the tissue response consists of a thin encapsulation that is formed within the first 1 to 3 months postimplant and is entirely stable thereafter.

Muscle-based electrodes provide an excellent method of delivering electrical stimulation to paralyzed muscles. These electrodes provide selective stimulation, deliver minimal trauma to the underlying tissue, are easily implanted, and are easily repositioned. There are several situations in which the neuromuscular anatomy dictates the need for muscle-based electrodes. For example, the extensor pollicis longus (EPL), which is the key muscle for providing thumb extension, must be activated strongly and in isolation from the nearby finger extensors. Isolating activation of the EPL from the more superficial EDC and extensor indicis proprius (EIP) is virtually impossible with the use of surface electrodes [22]. The multiple branching of the distal portion of the radial nerve precludes the distal placement of a nerve-cuff style of electrode isolated to the EPL motor branch. Therefore, the only existing method to reliably activate the EPL in isolation is to use a muscle-based electrode. This illustrates why we expect that muscle-based electrodes will be a necessary part of motor neuroprostheses for the near future.

Surgical installation of implanted electrodes does not present any unusual technical difficulties, especially in the upper limbs where all muscles are relatively easy to access and expose. The use of an epimysial-style mapping probe

allows the surgeon to move the electrode over the entire surface of the muscle while observing the stimulated response in real-time. Once the optimum location has been identified, the electrode can be temporarily sutured in place and the response tested again. Evaluation of the electrode response should be performed after the overlying muscles and skin are returned to their natural position. Then by varying the stimulus levels, the physician should perform an intra-operative evaluation of the electrode-muscle output characteristics. Important electrode-muscle output characteristics include the change in muscle force as a function of the muscle length and the change in muscle force output as a function of the stimulus intensity [11]. When these principles are followed, our results show that the postoperative response corresponds to the intraoperative response in 95 percent of the electrodes. In addition, the average threshold for the electrodes in this series was 15 μ s, which is considerably lower than the average threshold of 33 μ s obtained for percutaneous electrodes in the same group of muscles [12], indicating that the fully implanted electrodes can be placed more accurately.

Three electrodes were moved during revision procedures to gain improved performance of the electrode. All of these cases occurred during the first 13 subjects in this study. Our experience has been that we have not been able to find a significantly better electrode response in the attempted revision procedures, indicating that the initial placement was probably already close to optimal.

The long-term physiological response of the body to the presence of the electrode materials and to the stimulus current was an important concern in early studies [1,6,10]. The results of this study further confirm that the presence of the electrode materials (Pt-Ir, 316 SS, silicone) are well tolerated by the body and do not pose a source of long-term irritation, even when placed on (or in) contracting skeletal muscles. We found no evidence that daily delivery of electrical current through these electrodes at the stimulus levels used in this study (20 mA, 200 μ s,) electrode surface area: 10 mm²) caused any adverse response. No evidence was found of any progressive muscle weakness as a result of the chronic daily stimulation.

Although histology has not been performed on humans because of the risk of losing viable muscle tissue in patients who are already weak, we have had opportunity to grossly observe the tissue response to the implanted components during revision surgery. In three cases, the implant device itself was exposed after implantation periods of 6 months, 2 years, and 9 years. In all cases, we

found that a well-formed capsule with a thin wall surrounded the implant components. No cases indicated any infection, inflammation, or ongoing encapsulation.

The absence of long-term drift in the thresholds for the stimulating electrodes indicates that substantial encapsulation does not continue beyond the first few weeks after implant. Electrode thresholds have been consistent regardless of patient activity or stimulation time. This result suggests that the stimulation itself has no effect on the electrode-tissue interface.

There is a clinical concern that the presence of any foreign material within the body might be a potential site for infection. The presence of long leads running the length of the arm to the chest might also present a path for migration of infection. In this series of patients, only a single localized infection of one electrode was identified, and this infection was probably due to the suture in the wound above the electrode. The infection did not track along the electrode lead, but the lead was removed as a precaution. Other systemic infections have not tracked to the implantable components in this series of patients. Patients have reported unrelated infections, such as pneumonia, urinary tract infections, and even cellulitis, without any adverse affect from the implanted components. Although patients were instructed to take antibiotics as a precaution with any infection, compliance was difficult to gauge.

The incidence of mechanical failures with the electrodes, leads, and connectors was extremely small in this study. Only three failures were encountered, and most likely, only one of these three failures was an actual failure caused by repeated lead flexure. Including all leads in our analysis, the 98.7 percent survival for 16 years is excellent. This compares favorably to the reported survival rates of pacemaker, defibrillator, and spinal cord stimulator leads [23–28]. The connectors had no failures.

In the single subject where the lead appears to have undergone fatigue failure (subject 1W), extenuating circumstances were found. The electrode was placed on the surface of the AbPB so that it was quite superficial. This electrode was easily palpated and was located in a region that was subject to high stress. People with tetraplegia frequently place substantial body weight on their hands for weight shifts and transfers and stress on their arms and shoulder for wheelchair propulsion. This particular subject also pushed his own wheelchair and was otherwise quite active. All of these activities are likely to put stress on the thenar eminence located directly under the electrode. Because of this experience, we have now modified

the placement of the electrode so that it is more medial, deeper in the tissue and is therefore more protected from external stresses. We have had no further incidents of this type in the 7 years since this issue was first identified.

All three of the lead failures and the single electrode infection occurred within 2 years of implantation. Currently, 191 electrodes have been functioning longer than 5 years, with no failures or infections within this group. The trend of earlier failures is similar to that reported for other implantable leads [24,27,28]. Within the time frame of the present study (15 years), failure is not positively correlated to total time implanted. In addition, we have not had a lead failure occur in this study for the past 3 years, which may indicate that we have identified and rectified some of the possible failure mechanisms.

Even though electrode failure is an infrequent occurrence, identifying these failures quickly and confidently is important so that they can be corrected. We have developed a complete battery of methods that can be used to identify and analyze a possible failed electrode prior to surgical replacement, as described in this paper. In all three incidents where an electrode failed, the subject reported a different response or sensation from the neuroprosthesis. In one case, after the subject described changes in the location of the stimulus sensation, an X ray was used to reveal the problem. However, our experience is that only gross failures can be identified by X ray. In the other two cases, electrode threshold measurements, when compared to previously recorded values, indicated an unusually elevated threshold. The use of surface potential recordings allowed further confirmation and localization of the broken lead. To determine if the electrode impedance has increased above the expected range, one can use the shape of the recorded waveform. Theoretically, surface potential measurements can be used to identify a proximal current leakage through the lead insulation, but we have yet to confirm this in a human because the silastic insulation has produced a sufficient seal to prevent measurable current leakage at the break site. In our experience, yearly electrode thresholds provide the most straightforward means of identifying a failed or failing electrode.

CONCLUSION

In 27 patients who received an upper-limb neuroprosthesis, 238 electrodes have been studied. There is at least 3 years follow-up on all electrodes, with a maximum fol-

low-up time of 16 years. Three lead failures occurred and one localized infection occurred during the study, resulting in an overall failure rate of less than 2 percent. The expected lifetime of the electrodes is over 98 percent at 16 years. The device-tissue interface consists of minimal encapsulation that is stable over time. The results indicate that complications of a neuroprosthesis caused by device failure, electrode breakage, lead breakage, infection, or rejection are extremely low and are not a source of major concern. These results have an impact on the design of implantable neuroprosthetic systems. The electrode and lead component of these systems should no longer be considered a weak technological link.

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