

Wireless wearable controller for upper-limb neuroprosthesis

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Abstract—The objective of this project was to develop a wireless, wearable joint angle transducer to enable proportional control of an upper-limb neuroprosthesis by wrist position. Implanted neuroprostheses use functional electrical stimulation to provide hand grasp to individuals with tetraplegia. Wrist position is advantageous for control because it augments the tenodesis grasp and can be implemented bilaterally. Recently developed, fully implantable multichannel stimulators are battery-powered and use wireless telemetry to control stimulator outputs. An external wrist controller was designed for command signal acquisition for people with cervical-level spinal cord injury to control this implantable stimulator. The wearable controller, which uses gigantic magnetoresistive sensing techniques to measure wrist position, is worn on the forearm. A small dime-sized magnet is fixed to the back of the hand. Results indicate that the device is a feasible control method for an upper-limb neuroprosthesis and could be reduced to a small “wristwatch” size for cosmesis and easy donning.

Key words: control, functional electrical stimulation, gigantic magnetoresistance, hand grasp, joint angle sensor, neuroprosthesis, rehabilitation, spinal cord injury, tetraplegia, wireless, wrist angle.

INTRODUCTION

One of the most debilitating effects of a spinal cord injury (SCI) at the cervical level is the loss of hand function. According to a recent study [1], almost 50 percent of all persons with tetraplegia surveyed indicated that regaining arm and hand function would most improve their quality of life. Loss of hand function can severely

limit one’s ability to live independently and retain gainful employment postinjury. Thus, the development of treatments that lead to some functional recovery for the patient has the potential to significantly impact quality of life [1].

Functional electrical stimulation (FES) can be used to successfully restore hand grasp in someone with an SCI at the cervical (C) level [2–3]. The implantable hand-grasp neuroprosthesis, developed at Case Western Reserve University, uses voluntary movement retained by the subject to proportionally control the degree of hand opening and closing as well as grasp force. The device electrically activates paralyzed muscles by using electrodes that are either implanted within or sutured to the muscles in the hand and the forearm to provide two types of grasping patterns: a palmar grasp and a lateral pinch. Use of the neuroprosthesis provides patients with increased grasp strength, enabling them to manipulate

Abbreviations: 2-D = two-dimensional, BR = brachioradialis, C = cervical, DC = direct current, ECRB = extensor carpi radialis brevis, FES = functional electrical stimulation, GMR = gigantic magnetoresistive, GRT = grasp and release test, MES = myoelectric signal, MICS = Medical Implant Communication Service, SCI = spinal cord injury, SD = standard deviation, SOIC = small-outline integrated circuit, T1 = time 1, T2 = time 2, UECU = Universal External Control Unit.

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objects of different sizes and weights, and thus increases independence in activities of daily living [2,4].

The specific neuroprosthetic hand-grasp system discussed here requires two different types of control: a logical command signal and a proportional command signal. The logical command signal turns the device on and off; cycles through a set of predefined grasp patterns, such as lateral and palmar grasp; and locks or unlocks the device at certain grasp strengths. A continuous command signal is required to proportionally control the degree of hand-grasp position and force. Ideally, the continuous command signal is intuitive to the intended movement of the user [5].

A variety of command sources have been used successfully to control hand grasp. For patients who can extend their wrists, either through retained movement or a tendon transfer surgery, wrist position is an effective command source [6–8]. Wrist position is advantageous because it allows for bilateral implementation of a hand-grasp system and provides a more natural extension of the user's intact motor system by augmenting the tenodesis grasp [5,7].

In order to be used as a control method, the command source must be accurately detected and measured. Wrist position has been measured with external sensors as a method for controlling hand grasp [2,7–10]. Placing the sensor outside of the body is noninvasive and easy to both fix and adapt. However, it requires daily donning and doffing and is not cosmetically appealing. Wrist position has also been detected to control hand grasp with an internal sensor [6]. Implantable transducers have the advantages of being cosmetically acceptable and of reducing the possibility of inconsistent signal quality associated with donning and doffing [6]; but certain power, size, and material restraints are associated with implantation of a transducer inside the body.

The recent development of implantable stimulator technology has prompted the design of a wearable external controller. The Micropulse (NDI Medical; Cleveland, Ohio) is a small, rechargeable, wirelessly controlled implantable stimulator that has reached clinical application. The Networked Neuroprosthesis, under development by the Cleveland FES Center, is a modular, scalable, fully implantable technology that will also be able to accept an external wireless signal for control [11]. Thus, the wearable controller must be able to wirelessly communicate command signals to an implanted stimulator that can be translated into stimulation parameters for functional hand grasp. Wrist position was selected as an

appropriate command signal source. In this system, the controller will be worn on the wrist and wirelessly communicate with the implant, as shown in **Figure 1(a)**.

Some general functional and technical specifications are associated with the design of the device and are given in **Table 1**. These performance measures have been adapted from specifications for past successful control methods [5–6,12]. Functional specifications define certain tasks the device must perform. Technical specifications provide performance measures against which the controller can be measured. The device will be used to proportionally control hand grasp; thus, the sensor used to measure wrist position must provide a continuous monotonic signal over a patient's complete range of wrist movement. Covering the range of $\pm 40^\circ$ should be more than sufficient given active range of motion measurements on individuals with SCI [7]. The device must be easy to don and doff, be cosmetically acceptable, and have no physical connection across the wrist joint (**Figure 1(b)**). The device should be reliable and require calibration only with each daily placement. With regard to accuracy, there are various opinions concerning the necessary resolution of joint angle measurement for motor control purposes [12–14]. Typically, applications regarding feedback have higher resolution specifications than those used to measure a command source for control. In our experience using joint angle as a command source, accuracy and resolution requirements are lenient; thus, the accuracy specification for this transducer is $\pm 5^\circ$. Battery power must be sufficient for daily use, such that recharging is necessary nightly at most. Given commercial rechargeable battery options, the power consumption should be less than 20 mW. This specification assumes that a primary cell battery will be changed

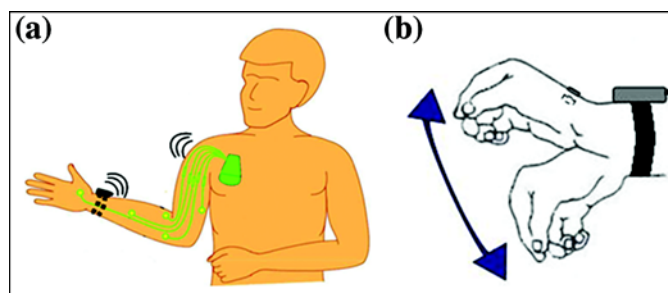


Figure 1.

(a) Schematic illustrating implementation of external controller that communicates wirelessly with implanted hand-grasp neuroprosthesis. (b) Suggested wrist-controller design illustrating wearable size and no physical connection across joint.

Table 1.

Associated functional and technical specifications of wireless, wearable controller for upper-limb neuroprosthesis.

Functional Specification	Technical Specification
1. Continuous Proportional Control	Monotonic signal with joint angle. Bandwidth of 30 Hz.
2. Ipsilateral to Arm Receiving Motor Function	—
3. Cover Complete Range of Wrist Motion	Specific range $\pm 40^\circ$.
4. Easy to Don and Doff	Subject can independently don and doff controller.
5. Cosmetically Acceptable	Prototype: Wearable, further miniaturization possible to “wristwatch-size” device. No physical connection across joint.
6. Reliable	Resolution $\pm 5^\circ$. Stability such that recalibration is only required after each placement. Measurements do not drift appreciably over time. No physical connection across wrist joint.
7. Onboard Power	Power consumption less than 20 mW.

once a week at most and a rechargeable cell will last for at least 24 hours before recharging is necessary. The signal should have a bandwidth of at least 30 Hz, which has been shown sufficient for hand-grasp control [6].

The objective of this work is to demonstrate the feasibility of an external wearable controller that measures wrist position to control a hand-grasp neuroprosthesis. Thus, the controller discussed here must be small enough to wear, with further miniaturization possible for commercial implementation. The device being designed is external; however, implantation of certain aspects may be desirable for further implementation as well. If so, the implanted components should be passive and capable of being implanted in a minimally invasive outpatient procedure. This article discusses the design of the controller and the initial testing completed to demonstrate feasibility for control of an upper-limb neuroprosthesis.

METHODS

Controller Design

Eventual implementation of the external controller will involve direct communication with an implant; however, this project was designed to illustrate the feasibility of hand-grasp control with an external wireless device. Thus, the controller designed for this study is composed of two major components that communicate via commercial wireless transceivers (DR3000, RF Monolithics Inc; Dallas, Texas) in one direction only. Although communication in either direction is possible, “handshaking” was not implemented for this specific study. The first component is the transmitting unit, or the wearable aspect of the

device. The transceiver within this component was fixed in transmit mode. The transmitting unit has several elements, including the sensors used for position detection and the processing and wireless communication components (**Figure 2**). The second major component is the receiving unit, which accepts the wireless signal from the transmitting unit and provides an analog output representative of wrist position. The receiving unit will be eliminated with the implementation of direct communication to an implant. It was developed specifically to communicate with the prototype wearable unit in order to demonstrate feasibility of control with use of a specific sensing technique.

Various transducers were initially investigated, including bend sensors, accelerometers, and position sensors (both inductive and magnetic). Because of power restraints and the desire to prevent any component from spanning the wrist joint, magnetic position sensors were chosen for this application. The specific transducers, manufactured by NVE Corporation, use gigantic magnetoresistive (GMR) sensing techniques to measure magnetic field strength. GMR sensors are noncontact, low power, and can withstand a large variation in gap distance. GMR sensing has certain advantages over more traditional Hall-effect sensing methods [15]. Advantages include increased sensitivity, temperature stability, and a larger signal level.

To measure wrist position, we integrated three GMR sensors (AAL002, AA004, and AA005, NVE Corporation; Eden Prairie, Minnesota) into a controller that can be worn on the wrist and fixed a disc-shaped rare earth magnet (Magcraft: D12.7 mm \times T1.6 mm, National Imports LLC; Vienna, Virginia) to the back of the hand.

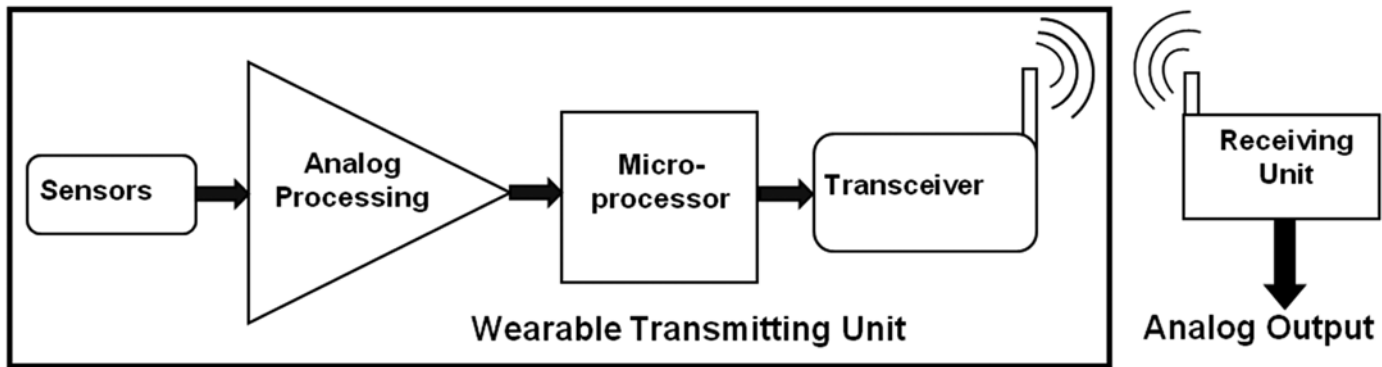


Figure 2.

Block diagram of wearable aspect of controller (transmitting unit) communicating with receiving unit. Receiving unit converts digital value representing wrist position to analog output that can control current upper-limb neuroprosthesis system.

Each sensor responds to a different range of magnetic field strengths (1.5–10 Oe, 5–35 Oe, and 10–70 Oe). Thus, by incorporating all three sensors, the controller is sensitive across a large range of motion. The output from each of the three GMR sensors is differentially amplified and processed in a microcontroller (PIC16F88, Microchip Technology Inc; Chandler, Arizona). The microcontroller communicates directly with the transmitting unit's transceiver. The transceiver within the receiving unit detects the wireless signal and converts it to an analog signal using a second microcontroller and a digital-to-analog converter (MAX518, Maxim Integrated Products Inc; Sunnyvale, California). This analog signal can then be recorded by a computer for data acquisition or used to command a current upper-limb system.

For the majority of tests, the prototype was powered by a standard direct current (DC) power supply. The wearable aspect of the controller has a maximum supply voltage of 3.3 V. The receiving unit requires power at both the 5 and 3 V levels. A low drop-off 3.3 V regulator was used in the receiving unit to obtain both signal levels from a 5 V supply. The prototype design of the wearable aspect of the device can also be battery-powered. Battery power was implemented when the controller was used to command an upper-limb neuroprosthesis.

Experimental Procedure

The evaluation of the controller was accomplished in three stages. First, bench testing verified the performance of the sensor in ideal conditions. Second, the sensor was characterized by mounting it on the arms of nondisabled volunteers and measuring its performance in more realistic

conditions. Third, one user who has an implanted neuroprosthesis was fitted with the controller. The user performed a series of manipulation tasks by using this controller in conjunction with his neuroprosthesis.

Bench Testing

Initially, the current draw properties of the wearable aspect of the device were evaluated to examine power usage. The transceiver may draw up to 12 mA of current during transmission; thus, the average current draw is related to the frequency at which a new position signal is transmitted to the receiving unit. The transceiver is placed in a current saving mode between each transmission. To evaluate current draw properties, the position signal was transmitted at a variety of frequencies ranging from 10 Hz to 1 kHz. Each position value is represented by an 8-bit number. The frequency of transmission refers to how often an updated 8-bit value was wirelessly communicated. For consistency, the maximum value, 255, was transmitted at each frequency specified. The transmitting unit was powered by a standard 3 V supply. A Hall-effect sensing current probe (TM502A, Tektronix; Beaverton, Oregon) evaluated current dynamics and measured the peak-to-peak current range. A Fluke 79 Series Multimeter (Fluke Corporation; Everett, Washington) was used in DC mode to measure the average current draw. The average current draw and supply voltage were then used to calculate power consumption of the wearable unit.

A mechanical model with two degrees of freedom was designed for initial validation of the controller as a position transducer. Movement along one degree of freedom is

similar to flexion/extension movement of the wrist. Movement along the other is similar to radial/ulnar deviation movement. The transmitting unit was placed on one side of the joint and a small disc-shaped magnet was placed on the opposite side. The receiving unit was placed near the transmitting unit, and the output of the controller over the model's complete range of movement was recorded at a frequency of 30 Hz with use of a National Instruments Data Acquisition card and LabVIEW software (National Instruments Corporation; Austin, Texas). During these trials, the transmitting unit sent an updated position measurement to the receiving unit every 4 ms. An Optotrak camera system simultaneously measured the position of the model joint (Northern Digital Inc; Waterloo, Ontario, Canada).

The model allowed isolation of movement along either the flexion/extension axis or the radial/ulnar deviation axis. Thus, the effect of each axis on controller output was measured. The resolution of the device was also calculated in these trials. To measure resolution, we determined the flexion/extension angle ranges associated with specific values of controller output. Another aspect considered in testing with the mechanical model was the stability of the signal over time. To study the stability of the signal, we initially placed the controller on the two-dimensional (2-D) model and collected data over the two axes of movement (time 1 [T1]). Power remained turned on for 2 hours and data were collected again (time 2 [T2]). The controller and magnet remained in the same position.

Nondisabled Testing

The repeatability of the controller was tested on five nondisabled participants. The controller was placed on each participant five different times, and the output of the controller and the position of the wrist were measured. The Optotrak camera system was used to measure actual joint angle. Initially, two rigid bodies were placed on the subject: one on the forearm and one on the distal region of the back of the hand. Four points were defined with respect to the rigid body on the forearm: the radial process, the ulnar process, the medial epicondyle, and the lateral epicondyle. The coordinate system representing the forearm was defined with use of these bony landmarks, as suggested by the International Society of Biomechanics Recommendations [16]. Three points on the hand were used to define the coordinate system corresponding to the hand. Each metacarpal bone can be defined separately for analysis of hand movement; how-

ever, since this specific study only considered global wrist position, the four medial metacarpal bones (excluding the thumb) were assumed to move as a rigid unit. The three bony landmarks used to determine the coordinate system of the hand were the distal head of the third metacarpal, the distal head of the fifth metacarpal, and the base of the third metacarpal. Euler principles were used to calculate the three angles of rotation between the forearm and the hand. The Optotrak components used to measure wrist position remained in place for the duration of the experiment with each subject.

The wearable aspect of the controller was attached to the subject's distal forearm with a Velcro elastic armband. The small disc-shaped magnet was taped to the back of the hand, approximately at the base of the third metacarpal bone. The magnet remained in place for the duration of the experiment with each subject. The controller was donned and doffed five times. With each placement, two trials were performed. After each placement, the subject was asked to move his or her hand through its full range of flexion/extension movement with the wrist either ulnar deviated, radial deviated, or held at a neutral position along the deviation axis. These tests were performed with the forearm held in pronation or in a neutral position. Each trial was 60 seconds in duration. The receiving unit was placed adjacent to the subject and received position information from the transmitting unit every 4 ms. The controller output and the locations of the bony landmarks determined by the Optotrak system were recorded simultaneously.

Data were analyzed offline to consider the effect of donning and doffing of the controller. The angles of the wrist were calculated with use of the positions recorded by Optotrak. On each subject, the data collected for each placement were binned according to flexion/extension angle value across the range of -40° (extension) to $+40^\circ$ (flexion) at 2° increments. All values within $\pm 1.0^\circ$ of the target values (-40° , -38° , -36° , etc.) were used to calculate the mean and standard deviation (SD) of the controller output at the target flexion/extension angles for each different placement of the controller and across the data collected at five placements combined for each subject.

Feasibility Study with Neuroprosthesis User

A feasibility study was completed with one current neuroprosthesis user to determine whether the device could be effectively used to control an implanted hand-grasp system. The subject was a male with a diagnosis of

tetraplegia due to SCI resulting from a fracture/dislocation of the C6 vertebrae that occurred in October 2000. His injury is classified as American Spinal Injury Association A (both motor and sensory complete) at the C6 level on both his right and left side [17]. The subject has retained active wrist extension, which is typical at this level of injury [18]. Further details, including active and passive ranges of wrist motion, are described in **Table 2**. Extension of the wrist results in passive finger and thumb flexion, thus providing some functional ability to grasp small, light objects [7].

The subject had neuroprosthetic hand-grasp systems implanted bilaterally. Each system is composed of a receiver-stimulator-telemeter implanted on the respective side of the chest and 12 electrodes placed on or in paralyzed muscles in the ipsilateral arm and hand [3]. In normal operation, the subject uses myoelectric commands for control of hand function. Two electrodes were implanted to record myoelectric signals (MESs) from muscles under voluntary control. The recorded signals are telemetered outside of the body and processed externally. The external unit, known as the Universal External Control Unit (UECU), contains the power for the system as well as the processing capabilities and is generally mounted on the wheelchair. A cable with a coil at one end extends from the UECU, and the coil is taped to the chest over the implant. It telemeters power and control commands across the skin to the implant, and the implant telemeters MES outside of the body to the UECU to be processed for control.

In the neuroprosthesis on his right side, which was the hand used for this specific study, the two electrodes recording voluntary MES activity were placed in the ipsilateral extensor carpi radialis brevis (ECRB) and the ipsilateral trapezius. The subject uses his wrist extensor muscle for proportional control of his right hand grasp.

Table 2.

Active and passive ranges of wrist motion and Manual Muscle Test (MMT) score for right side of feasibility test subject (current neuroprosthesis user).

Wrist Movement	Range of Motion (°)		MMT Score
	Active	Passive	
Extension	46	71	—
Flexion	—	70	—
Radial Deviation	19	23	—
Ulnar Deviation	—	25	—
Wrist Extensor Muscles	—	—	4-

The relationship between the level of the command signal and the stimulus level is defined in look-up tables known as “stimulus maps” [19]. This subject’s stimulus maps associated with two grasp patterns, palmar and lateral, were determined when he was initially trained to control the system using MES from his wrist extensor muscle during a rehabilitation stay after implantation surgery in November 2004.

To demonstrate the feasibility of hand-grasp control, we substituted wrist control for his myoelectric control. The wearable aspect (transmitting unit) of the controller and the magnet were fixed to the subject as described earlier for the nondisabled experiments and shown in **Figure 3**. The analog output of the receiving unit was connected to the subject’s UECU. An averaging filter was implemented within the transmitting unit, and a new position signal was transmitted to the receiving unit every 20 ms. To establish proportional control using wrist position, the subject was asked to move through his active range of wrist motion. Approximately 30 percent of his active range of wrist extension was used for control. The correlation between input voltage to the UECU from the controller and grasp pattern command was determined by software in the UECU.

A grasp and release test (GRT) was given to the subject to determine a measure of hand control performance. This test is described in detail in Wuolle et al. [20] and was one of the three primary measures of effectiveness used to evaluate patients’ performance of the Freehand System, an implanted neuroprosthesis that received Food and Drug

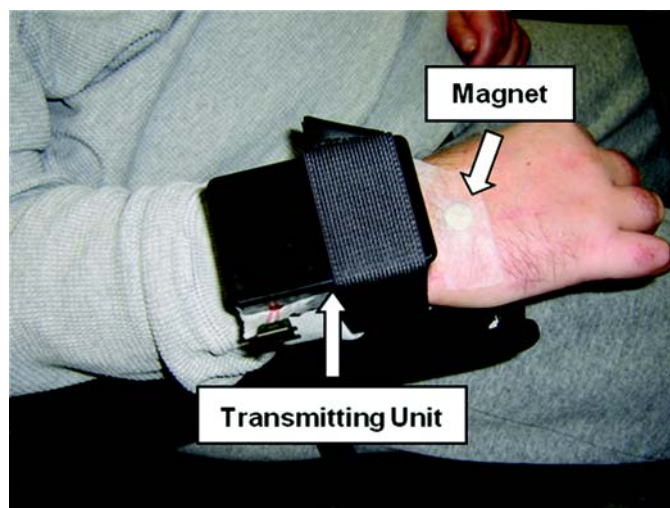


Figure 3.

Wearable aspect of controller (transmitting unit) and magnet placed on subject’s forearm and hand, respectively.

Administration premarket approval [21]. During the test, the subject was asked to grasp, move, and release six standard objects in a specific period of time. As described in Wuolle et al. [20], the specific objects used were:

1. Block, 2.5×2.5 cm, 0.1 N.
2. Can, 9.1×5.4 cm, 2.1 N.
3. Videotape, $20.4 \times 12.0 \times 3.0$ cm, 3.49 N.
4. Peg, 7.6×0.6 cm, 0.02 N.
5. Paperweight, 5.0×1.4 cm (disk mounted vertically), 2.59 N.
6. Fork, nylon handle attached to a spring loaded piston; requires 4.4 N to depress to indicated position.

The GRT board was placed on a table in front of the participant and measures 23×26 cm. Half of the board is a $23 \times 23 \times 4$ cm-high box with a removable top that also serves as a platform. In order to completely move an object, the subject must pick up the object at the start location, 10 cm from the edge of the board, and transfer the object either into the box (peg and block) or onto the platform (can, videotape, and paperweight). For each completion of the fork, the subject must grasp the fork and press down until a line on the handle passes a certain point. For each object, the number of completions in 30 seconds was the quantity used to measure hand and control performance.

To measure the effectiveness of the wrist controller, we asked the subject to complete the test using either his tenodesis grasp (no FES) or FES with the wrist controller. At the start of the test, the subject was given sufficient practice time with each object and control method. He was then tested three times with each object for each control method. During each of the three sets of trials, the order of the objects and the control methods were randomized. Within each 30-second trial, the number of completions and errors was recorded. Between each trial, the subject was allowed at least a 30-second rest period.

The mean number of completions and SD for each object were calculated for post-GRT analysis. A Student *t*-test was then used to determine significant differences in mean number of completions between the tenodesis grasp (no FES) and grasp with the neuroprosthesis using wrist control.

All human studies protocols received approval from the institutional review board at MetroHealth Hospital, and all subjects provided consent and release in accordance with the protocol.

RESULTS

Bench Testing

Results obtained with the Hall-effect current probe indicated that, as expected, current draw peaked during signal transmission. Peak-to-peak current was measured at a variety of signal transmission frequencies and did not vary significantly. Mean \pm SD peak-to-peak current was 10.2 ± 0.2 mA. Because the highest current level only occurs during signal transmission, average current draw varies with transmission rate, as shown in **Figure 4**. The maximum frequency tested represents a new position value transmitted every 1 ms. At this speed, the average current draw was 6.56 mA. Considering a supply voltage of 3 V, the power consumption is 19.68 mW. As evident from the plot, the average current draw decreased at a rapid rate initially as transmission speed decreased and then leveled off. Consider the transmission rate of 100 Hz; at this frequency a new position was transmitted every 10 ms, which is sufficiently fast for a control application [6]. Power consumption at frequencies at or below 100 Hz is approximately 12 mW.

Figure 5 shows the average controller output measured using the model with two degrees of freedom. In **Figure 5(a)**, the deviation angle was fixed at left, right, or neutral deviation and the model was moved through a range of flexion/extension angles, which are shown along the *x*-axis. (Extension is represented with the negative degree measures.) The *y*-axis is the output of the controller in volts. The plot shows mean values calculated at angle increments of 2° across trials completed at each deviation location. The left deviation was at an angle of 27° from neutral and the right deviation was at an angle

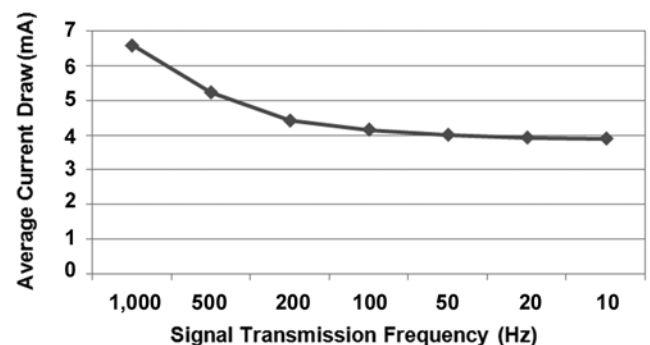


Figure 4.

Decrease in average current draw as frequency with which new position is transmitted decreases.

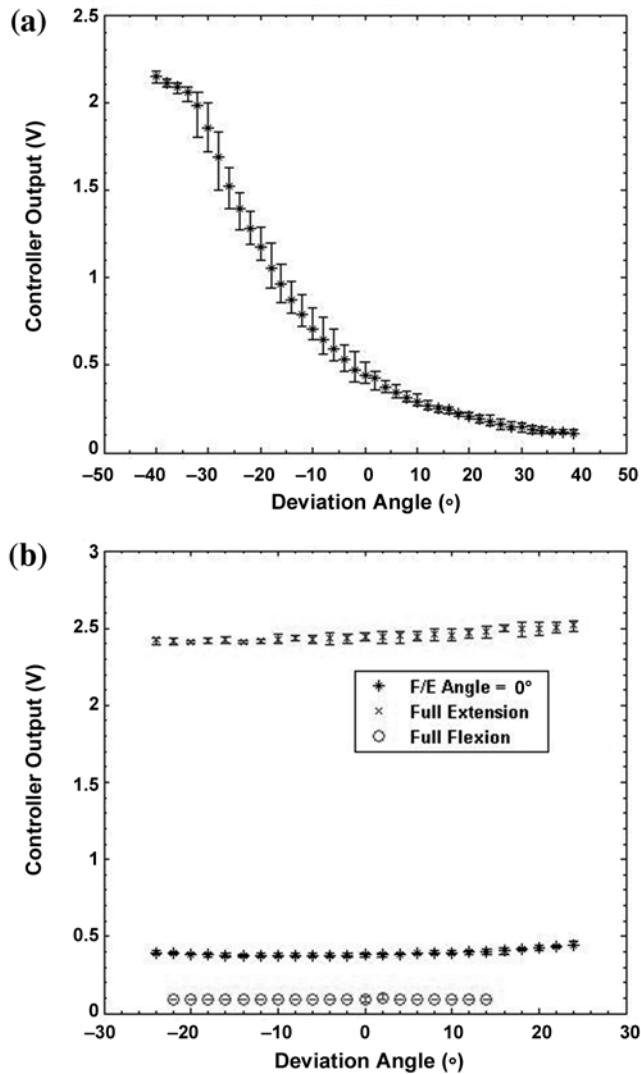


Figure 5. (a) Average controller output measured and plotted against flexion/extension (F/E) angle calculated from positions determined by Optotrak camera system. Error bars represent maximum and minimum controller output values measured at specific F/E angle. (b) Average controller output measured and plotted against deviation angle at three fixed F/E angles.

of 22°. The other two trials were collected when the model was at right deviation or left deviation. (Angle values are calculated after data collection.) Error bars on the graph represent the maximum and minimum value recorded at each flexion or extension value.

In **Figure 5(b)**, the flexion/extension axis was fixed and the model was moved along the deviation axis. Negative angle measures represent right deviation. Three fixed flexion/extension angles were considered: 0°, an extreme

flexion angle of approximately 58°, and an extreme extension angle of 53°. Again, mean values at 2° angle increments are plotted as well as error bars representing maximum and minimum values found. As evident from the figure, the controller output depends on flexion/extension movement. The 2-D model deviation movement does not significantly influence controller output.

Figure 6(a) illustrates the controller output characteristics over time. The mean controller output values as well as the maximum and minimum controller output at each specific angle value across eight different trials are plotted. Four of these trials were completed at T1 and four were completed at T2. Within the four trials measured at each time, two were completed at a constant neutral deviation angle, one at a left deviation angle, and one at a right deviation angle.

The resolution of the controller at specific output values is shown in **Figure 6(b)**. Here, the mean angle value associated with specific controller output increments of 0.03 V is shown. The error bars represent the maximum and minimum angle values with which that specific controller output was associated. The average and maximum resolutions across this range were approximately $\pm 2.5^\circ$ and $\pm 4.3^\circ$, respectively.

Nondisabled Testing

Nondisabled testing looked at the effect of controller placement multiple times on one subject and the variability of controller output between different subjects. Results, as shown for subject 4 in **Figure 7(a)**, indicate that controller output is consistent across different placements on an individual. **Figure 7(a)** is a plot of the mean and SD calculated at each 2° angle increment from +40° to -40° for one subject across all five placements. Negative and positive angle values represent the degree of wrist extension and flexion, respectively. **Figure 7(b)** illustrates the similar data collected for each of five subjects. As evident from this plot, controller output varies between each individual, as would be anticipated.

Feasibility Test Results

The GRT was used to assess the effectiveness of the wrist controller. The mean and SD for number of completions of each object are shown in the bar graph in **Figure 8(a)**. The first two bars represent the results from the initial GRT completed in November 2004 (GRT1) and show the mean number of completions for tenodesis grasp (no FES) and MES control (FES), respectively. The

third and fourth bars for each object illustrate the results of the GRT performed specifically for this study (GRT2) and show the mean number of completions for tenodesis grasp (no FES) and wrist control (FES), respectively.

As evident from **Figure 8(a)**, results from GRT2 indicate a significant increase in number of completions for both the tape and the fork. The subject was also able to pick up the paperweight using the wrist controller and his

neuroprosthesis, a task he was not able to complete with his tenodesis grasp alone. **Figure 8(b)** shows the mean and SD of number of completions for different groups of objects. The objects were divided based on weight into heavy and light groups, as well as based on grasp mode into palmar and lateral-pinch groups. The peg, block, and can were all considered light objects. Heavy objects were the fork, paperweight, and videotape. The peg, block, paperweight, and fork were all grasped using lateral pinch. The videotape and can were both maneuvered with palmar grasp. In **Figure 8(b)**, the top line shows the sum of the mean and SD for all objects; below that are lines representing the completion differences for the four different

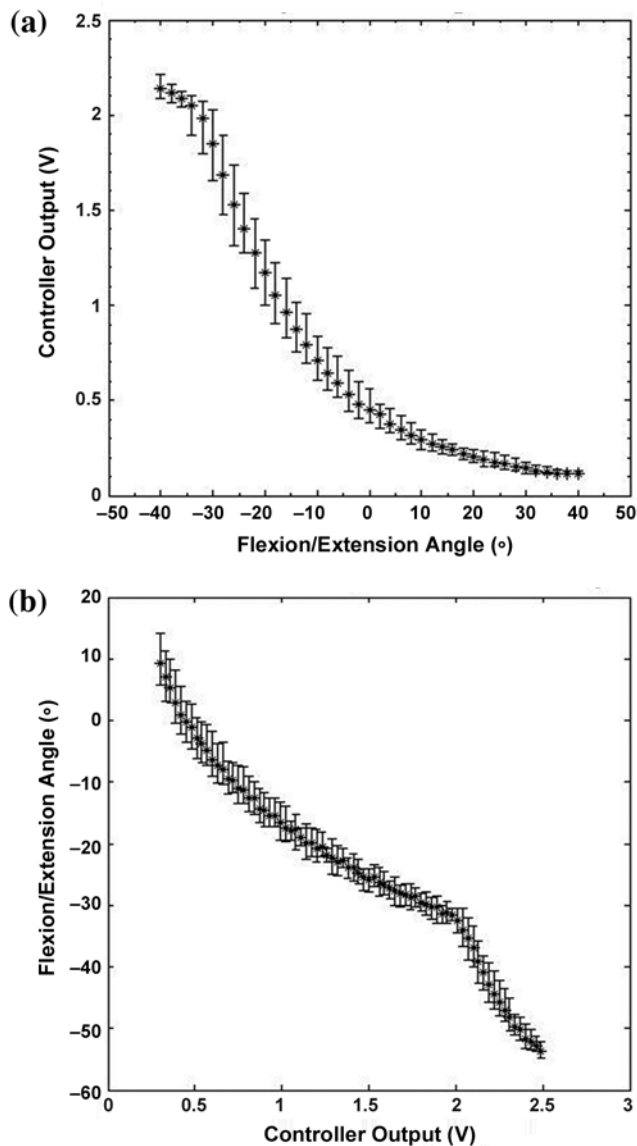


Figure 6. (a) Average controller output measured against flexion/extension angle. Trials were recorded across 2 hours at fixed deviation angles. Error bars indicate maximum and minimum values at each flexion/extension angle. (b) Resolution measured at specific controller output values. Error bars represent range of angle values associated with specific output value.

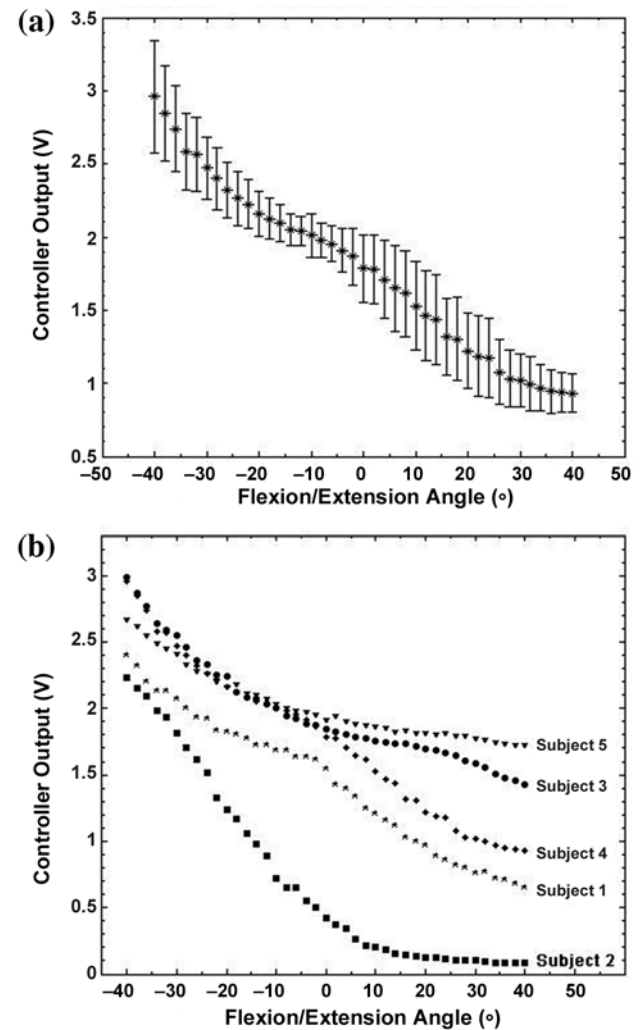


Figure 7. (a) Mean \pm standard deviation controller output calculated across range of $\pm 40^\circ$ at 2° increments for subject 4. (b) Mean controller output calculated again at 2° increments for all five subjects.

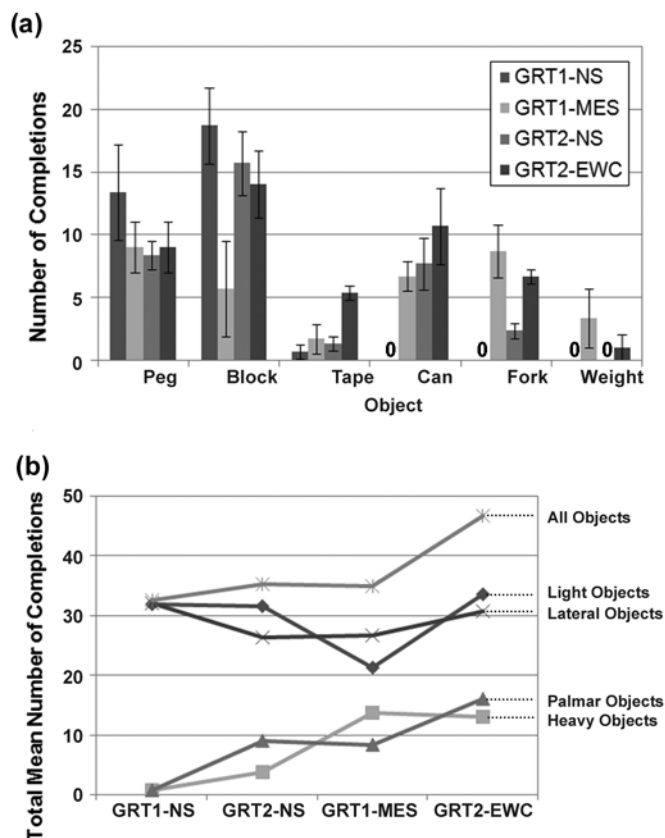


Figure 8.

(a) Mean number of completions for all six objects across two different grasp-and-release tests (GRT1 and GRT2). GRT1 was completed in November 2004 and measured completions with use of tenodesis grasp (no functional electrical stimulation [NS]) and myoelectric signal [MES] control of the neuroprosthesis. GRT2 was completed in January 2007 and measured completions with use of tenodesis grasp and external wrist control (EWC) of neuroprosthesis. Zero in place of bar indicates object subject was not able to complete. (b) Sum of mean number of completions for objects grouped either by weight (light or heavy) or by grasping mode (lateral or palmar) used for GRT1 and GRT2.

groups. A significant increase in number of completions occurred in both the heavy-object group and the palmar-grasp object group within GRT2. Note the differences between MES and wrist control that are evident in the light-object group and the palmar-grasp group.

DISCUSSION

An external wireless controller has been designed that measures wrist position to command an upper-limb neuroprosthesis. The device meets the specifications

given in the “Introduction.” The controller provides a continuous and monotonic signal over the full range of joint motion without a physical connection across the joint. The resolution of the controller is approximately $\pm 2.5^\circ$ and was sufficient for proportional command of the hand-grasp neuroprosthesis used by the subject in the feasibility study. The prototype device is small enough to wear on the wrist, and further miniaturization of the wearable aspect is possible. The magnet is easily fixed to the back of the hand; however, because this component does not require power, implantation of the magnet under the skin is a feasible option. The device has power consumption less than 20 mW. Verification results indicate that the controller output is stable across five different placements on an individual. The study with a current neuroprosthesis user showed that measurement of wrist position with this specific controller is a feasible control method for an upper-limb neuroprosthesis.

Wrist Position as Command Source

Wrist position is a viable command source for someone with adequate voluntary control of wrist extension. Active wrist extension is typically retained in someone with a C6 or lower SCI and can be returned to someone with an injury at the C5 level by using a tendon transfer of the brachioradialis (BR) muscle to the insertion point of the ECRB [22–23]. In this study, as well as a previous study in which wrist position was used as a command source, all subjects had a Manual Muscle Test score of at least 4– in the muscle used to extend the wrist (either external carpi radialis longus or a transferred BR) [7]. The range of motion required depends on the levels of command desired for proportional control and the resolution of the sensor. The approximate resolution of this sensor is $\pm 2.5^\circ$. Thus, for someone with at least a 25° range of wrist extension, five distinct levels could be set on the proportional command scale. A larger range of motion allows for more distinct levels within the proportional command scale and more precise control of hand position.

If adequate wrist extension is retained, using wrist position for a command source is beneficial because it augments tenodesis grasp and is thus easily learned by neuroprosthesis users. This ease of learning is evident from the lack of training required for the feasibility study discussed in this article and the results discussed in Hart et al. [7]. Minimizing the training necessary for volitional control of functional assist devices, such as a neuroprosthesis, can lead to subconscious control [5]. Subconscious

control allows the user to focus more intently on complex tasks and reduces the cognitive effort necessary to perform simple tasks. Because wrist extension is a method traditionally learned by people with tetraplegia to grasp light objects by using their tenodesis grasp, it can easily be converted into a command source for a hand-grasp neuroprosthesis to increase grasp strength, thus allowing manipulation of heavier objects as well.

The results of the GRT in the feasibility study illustrate the success of wrist-position control coupled with a neuroprosthesis to improve manipulation of heavy objects. Objects such as the peg and block may be picked up by tenodesis grasp alone, but GRT2 also indicates that FES with wrist control does not hinder a user's manipulation of these lighter items. No significant difference was found in the number of lighter items completed in GRT2. Heavier objects, on the other hand, are difficult to grasp and maneuver without electrical stimulation. The mean number of heavy objects completed with use of the neuroprosthesis with wrist control significantly increased compared with tenodesis grasp alone.

The GRT also helps describe the inherent success and limitations of wrist control when used in the two different grasp modes. The number of objects completed with use of the palmar grasp in GRT2 significantly increased. Both the videotape and can were manipulated with the hand in palmar-grasp mode. This increase could be explained by two factors. One factor to explain the increase in completions is the weight of the objects. Although the can is considered a light object, it is larger and heavier than both the peg and the block. The videotape is considered a heavy object, and a statistically significant difference was seen between the mean number of completions of the videotape alone with and without the neuroprosthesis in GRT2. A second factor that may explain palmar-grasp success could be the ease of using wrist position for control in a neutral pronosupination position. The objects in the palmar group are grasped with the forearm in this neutral position. The neutral position allows movement of the wrist without much effect on finger placement. Thus, the neuroprosthesis user could position his fingers around the object with the hand in an open position and then isolate extension of the wrist to grasp the object. The objects manipulated with use of lateral pinch were all grasped with the forearm in a pronated position. With the forearm pronated, as with tenodesis grasp, extension of the wrist can cause the fin-

gers to pull away from the object, making the object more difficult to grasp [7].

The ease of wrist control in the neutral pronosupination position also offers a possible explanation for the differences that are evident between MES and wrist-position control. This study was not designed to directly compare MES control and wrist position; results from GRT1 are included to show that wrist control with use of the wireless device is comparable to an acceptable control method. However, noticeable differences between the two control methods help to illustrate pros and cons associated with wrist-position control. The subject used voluntary movement of his wrist extensor muscle to control his hand-grasp neuroprosthesis in GRT1. He could contract this muscle without changing the position of his wrist and thus could place his fingers around an object such as the fork in the pronated forearm position and fire his ECRB muscle without causing subsequent movement of his fingers. Wrist control showed an increase in the number of palmar-grasp objects compared with MES control. Grasping with the forearm in a neutral pronosupination position lessens the effect of wrist extension on finger placement; thus with wrist control, the subject could grasp more quickly those objects that he held with his hand in a neutral forearm position.

A significant difference was also seen in the number of light objects grasped with use of wrist control versus MES control. Wrist control augments the tenodesis grasp. The user could pick up light objects with his tenodesis grasp alone. Thus, tenodesis-like control of his hand-grasp neuroprosthesis is an intuitive control method for grasping these objects, as evident from the similar number of light objects completed both with and without FES in GRT2 (wrist control).

Wrist position has cognitive benefits because it is intuitive to the intended motion of the user, but it also has limitations, such as the associated figure movement that has already been discussed. Another limitation associated with both tenodesis grasp and neuroprosthesis control with wrist position is dependence of wrist position on outside forces such as gravity or the weight of an object being grasped. Voluntary control of wrist flexion muscles is not retained after injury at the C5–C6 level. With the hand in a supinated position, gravity fully extends the wrist. A heavy object adds an additional extension force on the wrist when the forearm is supinated. Thus, hand opening cannot be achieved in a supinated position when wrist position is used for control.

Wrist-position control is an attractive candidate for a command source since it is an extension of the user's intact motor system. Wrist position is an ideal command source to measure externally and requires little processing. Results from this article indicate that a wearable wrist-position controller is feasible given current sensor and wireless communication technology.

Implementation Considerations

The device considered in this article was a prototype designed to investigate the feasibility of a wearable wireless controller for use with a hand-grasp neuroprosthesis. Results indicate the success of GMR sensors to measure joint position along one axis of movement. The results also provide a basis for further considerations of implementation of this device. Specific controller aspects such as size and power consumption need to be considered and are associated with user concerns such as the cosmesis, comfort, and convenience.

Further miniaturization of the controller is necessary for commercial application. Miniaturization will not only improve cosmesis but will also improve ease of wear and reduce variability associated with controller movement. The improvement of cosmesis with reduction of size is obvious. Size reduction will also help prevent controller movement on the arm. The greater the distance between the top of the device and the forearm, the more likely the controller is to respond to gravitational forces in a neutral pronosupination position and shift on the arm. A controller similar in size to a wristwatch is expected to reduce the possibility for movement around the forearm and improve cosmesis, thus increasing user acceptance.

The size of the device depends on two factors: the size of the circuit board and the size of the battery required for onboard power. The circuit board in the wearable aspect is approximately 4.5×5.0 cm and stands 1.5 cm high with the antenna mounted. For the studies in this article, the board was mounted in a commercial enclosure ($5.5 \times 8.5 \times 4.0$ cm) along with three AAA batteries. Three sensors are necessary to measure the complete range of wrist motion. Each sensor is encased in a standard 8-pin small-outline integrated circuit (SOIC) package. Bulk manufacturing may permit all three sensors to be placed in one 8-pin SOIC package, reducing the area occupied by the sensors on the circuit board. The single component that covers the largest amount of area on the circuit board is the transceiver. Implementation of wireless communication across the skin would require

the use of the Federal Communications Commission-regulated Medical Implant Communication Service (MICS) band. MICS band transceivers are inherently small and low power because they are designed for use in medical implants. Further optimization of component placement on the board has the capability to reduce board size to the width and length of the battery required.

Battery size is a factor related to both miniaturization concerns and power. The results from the average current draw experiments indicate that power consumption is directly related to the frequency at which a command signal is transmitted. At the maximum speed used in this study, 1 kHz, the power consumption was still less than the given specification of 20 mW. A more typical transmission rate for movement applications (50 Hz) can bring down power consumption to 12 mW. At this rate, a lithium polymer rechargeable battery (UBC581730, Ultralife; Newark, New York) would be sufficient for approximately 50 hours of use before it would be necessary to recharge the device. The battery dimensions are $18.0 \times 31.5 \times 5.8$ mm. With a circuit board design optimized to the same dimensions as the battery, the controller dimensions could be as small as $25.0 \times 35.0 \times 10.0$ mm.

In addition to controller size, calibration is an important concern for user acceptance. Concerns related to daily calibration of the device are reduced by the results of the placement experiments on nondisabled subjects. The results indicate that the controller may not need to be calibrated with each placement. Controller output across the five placements for each subject was similar. However, the output range did vary from subject to subject. This suggests that the controller will need to be calibrated for each individual but that calibration may not need to be performed daily.

Implantation of the magnet underneath the skin may have significant advantages including the reduction of controller output variation between subjects and the improvement of cosmesis. Possible sources of variation between subjects include the size of the back of the hand, the diameter of the distal end of the forearm, and the amount of skin across the wrist joint. Because the magnet was fixed to the back of the hand with medical adhesive, it moved as skin either stretched across the joint during wrist flexion or collected at the wrist during wrist extension. When the wrist was in a neutral flexion/extension position, the magnet was parallel to the back of the hand. However, as the wrist extended and skin began to fold underneath the magnet, its position with respect to the

metacarpal bones in the hand changed. Implanting the magnet underneath the skin and fixing it to the base of the third metacarpal bone may reduce signal variability. Necessarily, the magnet would have to be packaged in a biologically compatible material such as titanium. This approach has been employed in implantation of magnets in other devices, such as cochlear prostheses, and would not be expected to generate unanticipated regulatory hurdles [24–25]. Implantation of the magnet will also improve cosmesis and increase the ease of donning the device, both of which would be expected to lead to increased user acceptance.

Future Applications

The external wrist controller uses GMR position sensors to measure the joint position of the wrist; the basic sensing techniques involved could be incorporated into a variety of possible applications involving measurement of movement. Possible applications include the transduction of the angle at other various joints for feedback and control. This specific project measured position along one axis of movement. The transducer could be easily implanted to measure joints that primarily cover one-dimensional motion, such as the elbow or the knee. Small, low-power sensors are ideal for integration with control systems requiring joint-position feedback. Additionally, the device developed could be used as a component of a motion-capture system for research or clinical purposes. It offers the distinct advantage of wireless communication. A wireless control and data-collection system could be developed to communicate and further process the data collected with various GMR sensing devices placed across joints.

CONCLUSIONS

A prototype of a wireless, wearable device that measures wrist position was developed to illustrate feasibility of control for an upper-limb neuroprosthesis. The device is low-power, cosmetically acceptable, and reliable when measuring position along the flexion/extension of the wrist. Results indicate that the controller must be calibrated for each individual. Wrist position is an effective command source for an upper-limb neuroprosthesis as shown here and indicated in Bhadra et al., Hart et al., and Prochazka et al. [6–8]. As with any command application, advantages and disadvantages are associated with

wrist position. The optimal command source for an individual depends on the neuroprosthesis system as well as the voluntary movement retained postinjury. Advances in sensor technology and wireless communication make wrist position an ideal command source to measure with an external wearable controller. The device described in this article can be used to successfully control an upper-limb neuroprosthesis; thus, it can increase the functional skills of a person who has tetraplegia and may significantly impact the quality of his or her life.

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