Improving Elbow Torque Output of Stroke Patients with Assistive Torque Controlled by EMG Signals

This paper develops an assistive torque system which uses homogenic surface electromyogram (EMG) signals to improve the elbow torque capability of stroke patients by applying an external time-varying assistive torque. In determining the magnitude of the torque to apply, the incorporated assistive torque algorithm considers the difference between the weighted biceps and triceps EMG signals such that the applied torque is proportional to the effort supplied voluntarily by the user. The overall stability of the assistive system is enhanced by the incorporation of a nonlinear damping element within the control algorithm which mimics the physiological damping of the elbow joint and the co-contraction between the biceps and triceps. Adaptive filtering of the control signal is employed to achieve a balance between the bandwidth and the system adaptability so as to ensure a smooth assistive torque output. The innovative control algorithm enables the provision of an assistive system whose operation is both natural to use and simple to learn. The effectiveness of the proposed assistive system in assisting elbow movement performance is investigated in a series of tests involving five stroke patients and five able-bodied individuals. The results confirm the ability of the system to assist all of the subjects in performing a number of reaching and tracking tasks with reduced effort and with no sacrifice in elbow movement performance. [DOI: 10.1115/1.1634284]

Introduction

Hemiparesis is a common deficit which causes stroke patients to be less efficient in lifting and supporting loads in the affected side. Conventionally, a physical rehabilitation approach is adopted to improve the muscle strength of patients with milder deficits, while functional electrical stimulation offers promise for patients with more severe deficits [1–3]. However, it has been observed that the results of rehabilitation are often less than satisfactory for patients with moderate to severe deficits. Accordingly, researchers have developed a variety of mechanical systems capable of increasing elbow torque as a means of assisting patients who are unable to regain adequate muscle force after rehabilitation.

Reinkensmeyer et al. [4,5] proposed the use of an assistive mechanical guide for brain-injured individuals to improve the active range of their arm motion and to compensate for the passive properties of the arm, including gravitational load and passive tissue stiffness. The results of this study demonstrated the effectiveness of the proposed counterpoise assistance in increasing the active range of arm motion, but revealed that the mechanical guide did not enable the users to move through the full range of passive motion. Furthermore, since the system was designed to compensate for the passive load of the arm, rather than to improve muscle strength, it was found that individuals were unable to generate sufficient muscle force when the arm was extended beyond a certain range. Cozens [6] employed a servomotor to apply a velocity-dependent assistive torque to the elbow of stroke patients. When the velocity of the patient’s motion was less than a predetermined threshold, the servomotor applied an increasing torque until either the torque or the velocity reached a preset limit, at which point the torque was gradually reduced. Testing of this robotic assistance technique confirmed that patients were able to perform a large range of extension-flexion movements. However, it was noted that the assistive torque approach failed to adapt adequately to unpredictable situations such as when the patient performed motion under loading or with disturbances. Kiguchi et al. [7] developed an exoskeletal robotic system incorporating a fuzzy-neuro control algorithm which collected surface EMG signals to estimate the degree of elbow torque required. The results of this study demonstrated the effectiveness of the proposed system in facilitating a wide range of elbow motion.

When using surface EMG signals to control the assistive torque, the method adopted to estimate the torque information from the collected EMG signals is an important consideration. It has been shown previously that surface EMG evaluation involves a complex nonlinear function of numerous physiological and experimental factors [8], including the number and spatial distribution of muscle fibers and motor units, the length of the muscle fibers, the contraction velocity, the location of the sensing electrodes, etc. Researchers have developed a variety of processing methods for EMG feature evaluation which permit the measured EMG signals to be related to the muscle force for a single muscle [9–11].

The control algorithm adopted in the present study is simplified by the assumption that the voluntary elbow torque can be determined solely from the elbow angle and the dynamic EMG signals of the biceps and triceps. Therefore, the ratios of the unilateral EMG signals to the elbow torque under isometric contraction at various elbow angles and torque levels are calculated, and the applied assistive torque is proportional to the difference between the weighted EMG signals of the biceps and triceps.

Since the frequency content of a surface EMG signal is considerably higher than that of human movement and is of a broader bandwidth, it is necessary to process the raw EMG data in order to generate a lower frequency signal for the assistive device control system. Various methods have been proposed for deriving control signals from the measured surface EMG signals. These methods include taking the integral of the average rectified value, and variance and autoregressive models [11]. However, the low-pass filtering which results from these methods prevents the system from performing rapid movements. Accordingly, past studies generally only employed EMG signals for switch control purposes or as a means of identifying the direction of movement [1,2,12]. In order
The intention of this present study is to develop a control system which provides an assistive torque to the elbow joint whose magnitude is proportional to the processed homogenic EMG signals of the biceps and triceps. The proposed assistive system aims to improve the elbow torque capability of stroke patients without sacrificing elbow movement performance. The study assesses the performance of the developed assistive system against two design objectives, namely its ability to reduce the voluntary effort required to support loading, and the requirement that the tracking movement when using the assistive device should be at least as accurate and smooth as the performance when the system is not used for stroke patients and able-bodied users alike.

Methods

Experimental Setup. Figure 1 presents the experimental setup adopted in the present investigation, in which a servomotor-driven manipulator provides an assistive torque to the subject’s elbow in order to help him or her support a predetermined load. A torque sensor is located between the motor and the elbow holder, and a weight is suspended around a disk of radius 25 cm to provide the loading effects during the reaching and tracking experiments. Since the elbow torque capability varies from subject to subject, the load was specified to be approximately 40% of the minimum of the maximum voluntary elbow torque in the flexion and extension directions for each particular subject. The monitor shown in Fig. 1 was used to provide visual feedback during the reaching and tracking experiments when the subjects were requested to support the applied load and to follow the predefined trajectory displayed on the monitor.

EMG Processing. The EMG signals of the biceps and triceps brachii were measured using standard Ag-AgCl surface cup electrodes. Prior to sampling, the signals were amplified with a gain of 10,000 and low-pass filtered in 300 Hz. The signals were all amplified with a gain of 10,000 and low-pass filtered in 300 Hz. The signals were all sampled at 500 Hz/channel, band-passed in 10 and 200 Hz and then calculated using the formula proposed previously by Hogan and Mann [9], i.e.

\[ w_j = \left( \frac{1}{n} \sum_{i=1}^{n} M^2_{ij} \right)^{1/2} \]

where \( j \) is an integer representing the number of the processed EMG signal, \( n \) is the number of samples in the moving window (100 msec) used for evaluating the EMG feature, and \( m \) is the original sampled EMG signal. The processed EMG signal can then be normalized as follows:

\[ M_j = \frac{w_j - w_{rel}}{w_{rel}} \]

where \( w_{rel} \) is the processed EMG signal in the muscle-relaxed state, and \( M_j \) is the normalized EMG signal. From this expression, it can be seen that subtracting \( w_{rel} \) from \( w_j \) causes \( M_j \) to tend toward zero when the muscle is in the relaxed state.

As shown in Eq. (3) below, the assistive torque was estimated on the basis of the difference between the weighted EMG signals measured at the biceps and triceps. Since the direction of the subject’s voluntary exertion is indicated by the sign of the difference, it is possible to apply the torque to the elbow in such a way that it is always in the “assistive” direction, i.e., it always reinforces the subject’s voluntary effort. Although the difference between the weighted EMG signals provides an approximation of the net torque acting on the joint as a result of the subject’s effort, it ignores the joint stiffness considerations arising from co-contraction. Therefore, in order to reflect various degrees of joint stiffness, a nonlinear damping term was added to the control command as a function of the summation of the biceps and triceps EMG signals and a one-third power of the elbow angular velocity [13]. From the form of this function, it can be seen that the physiological damping of the elbow joint is mimicked by modeling a damping effect which is relatively larger for lower velocities of motion and for larger co-contractions. Accordingly, the total estimated torque output (\( T_c \)) of the system is expressed in terms of the difference between the weighted EMG signals of the triceps and biceps and the additional nonlinear damping effect term, i.e.,

\[ T_c = G \left( K_t M_t - K_b M_b - C_0 (M_t^2 + M_b^2)^{1/2} \right)^{1/3} \]

where \( G \) and \( C_0 \) are constants, \( K_t \) and \( K_b \) are determined from measured data of the elbow angle and the EMG level. These gains are estimated under isometric contraction at four joint angles (i.e., 65, 90, 115, and 140 deg, where full elbow extension is taken to be 180 deg) and three torque levels (i.e., 16.7%, 33.3%, and 50% of maximum voluntary torque) in both the flexion and the extension directions. The values of the gains are calculated as the ratios of the elbow torque to the processed EMG signal of the agonist. For example, the value of \( K_t \) at 90 deg with a 50% torque level is calculated as the ratio of the measured elbow torque to the processed triceps EMG when the subject is requested to support a load corresponding to approximately 50% of the maximum voluntary torque in the extension direction. Interpolation and extrapolation techniques were utilized to derive appropriate gain values for system control purposes at arbitrary torque levels and angles in the reaching and tracking experiments.

Since the proposed system uses a servomotor to apply the assistive torque to the elbow, rapid changes in the applied torque might cause the subject to experience operating difficulties. Therefore, an adaptive filter was employed to reduce the rate of change of the torque. The torque output of the system, \( T_c \), was low-pass filtered with this adaptive filter [14] to obtain \( T_{c,j} \), i.e.,

\[ \frac{T_{c,j} - T_{c,j-1}}{S} + T_{c,j-1} = T_{c,j} \]
where $S$ is the sampling period, $T_{cj}$ is the $j$th output of the low-pass filter which is provided to the servomotor as a control command, and $z_j$ is a variable time constant adjusted in accordance with the following expression:

$$z_j = \beta \left( \frac{(p_j - p_{j-1})/S}{p_j} \right)^{2/3}$$

(5)

where $\beta$ is a constant, and $p_j$ is the output of $T_j$, which then passes through a second order low-pass filter (cutoff frequency $= 2.5$ Hz). The function of the adaptive filter is to perform an appropriate tuning of the cutoff frequency given in Eq. (4) so as to maintain a rapid response at the onset of movement and to dampen the oscillation of the control signal in order to maintain a stable condition. As the value of $\beta$ increases, the time constant increases and the cut-off frequency decreases. In the current study, $\beta$ was specified as being equal to 1, which allowed the cutoff frequency to vary within the range of 0.25 and 2.5 Hz when the absolute value in Eq. (4) varied from 0.125 to 5 [14].

Experimental Procedures. In order to demonstrate the feasibility and effectiveness of the proposed assistive system, the Bioethics Board of National Cheng Kung University, Tainan, ROC, authorized the recruitment of five able-bodied subjects and five stroke patients to participate in a series of experimental studies. Table 1 presents the relevant data pertaining to each of these ten individuals. It is noted that the stroke patients were all male, aged between 45 and 70 years old, and with their right sides affected.

As shown in Fig. 2, during the experiments, the subjects lay in a supine position with their forearm attached to the manipulator by means of an elastic strap. Before the main reaching and tracking experiments were conducted, isometric contraction measurements were performed to estimate the gain mapping functions ($K_t$ and $K_b$). Individual maps of the gains $K_t$ and $K_b$ were plotted as functions of the EMG signals and the joint angle using interpolation and extrapolation techniques. Figure 3 presents the example of the map derived for $K_t$. Meanwhile, Table 2 provides the full set of triceps and biceps gains and processed EMG signals for subject S1 over a range of elbow angles at different degrees of voluntary torque.

The current study performed two basic experiments, namely reaching and tracking. During each of these experiments, a monitor was used to display both the required and the actual elbow movement trajectories to the subject. The reaching experiments were only performed by the able-bodied subjects, and were designed to investigate the effects of nonlinear damping and adaptive filtering on the system stability. In this series of experiments, the subjects followed the predefined trajectory presented on the monitor by performing a step-up movement within a range of 60 deg with a preset load, and then maintaining the elbow at the terminal angle for 5 s. As stated previously, the load was specified as approximately 40% of the individual’s maximum voluntary torque capability. The gain constant ($G$) of the assistive torque was set at 100%, which implied that the torque applied by the manipulator was approximately equal to the voluntary torque supplied by the subject. The Integrated EMG magnitude (IEMG) and Mean Path Length (MPL) were utilized as performance indicators throughout these experiments. The IEMG indicator, which was

\begin{table}[h]
\centering
\caption{Details of 10 test subjects}
\begin{tabular}{lllllll}
\hline
Subject$^*$ & Dominant side & Maximum torque (Nm) & External load (Nm) & Sex & Age \\
& (affected side) & Extensor & Flexor & & \\
\hline
AB1 & L & 23.01 & 32.32 & 8.9 & male & 24 \\
AB 2 & R & 15.65 & 21.45 & 6.12 & female & 25 \\
AB 3 & R & 23.22 & 30.64 & 8.9 & male & 23 \\
AB 4 & R & 28.08 & 42.82 & 11.12 & male & 24 \\
AB 5 & R & 29.4 & 45.23 & 11.68 & male & 24 \\
S1 & R & 25.69 & 29 & 10.01 & male & 49 \\
S2 & R & 11.62 & 17.24 & 4.45 & male & 45 \\
S3 & R & 9.17 & 12.57 & 3.89 & male & 57 \\
S4 & R & 25.83 & 28.72 & 10.01 & male & 48 \\
S5 & R & 14.06 & 15.69 & 5.56 & male & 70 \\
\hline
\end{tabular}
\footnotesize{$^*$AB 1~AB 5: able-bodied subjects; S1~S5: stroke subjects}
\end{table}
calculated by summing the processed EMG signals and then dividing by the movement time (5 s), was taken to represent the voluntary exertion, while the MPL index was used to estimate the smoothness of the movement, and was calculated as the ratio of the total path length of the angle trajectory to the movement time. Having completed the reaching experiments for the five users, the results were evaluated and a set of parameters (i.e., \( C_0 = 0.2 \) and \( \beta = 2 \)) was established for the subsequent tracking experiments.

The aim of the tracking experiments was to investigate the performance of the assistive system. Each subject was requested to move his or her forearm to match the trajectory displayed on the monitor while supporting a preset load. As shown in Fig. 4, the predefined trajectory consisted of a ramp-up stage, followed by a hold stage and then a subsequent ramp-down movement. This series of movements was designed specifically to verify the functionality of the assistive system under both concentric and eccentric contractions. As in the reaching experiments, the range of movements considered in this series of experiments was set to be 60 deg. Furthermore, the ramp-up movement was conducted at a speed of 20 deg per second, while a slower speed of 10 deg per second was adopted for the ramp-down movement. As before, the supported load corresponded to approximately 40% of the subject’s maximum voluntary torque capability, and the assistive torque gain was set at 100% for all subjects. Tests were also performed using a gain of 150%. However, these tests were only performed by the able-bodied subjects since a gain of this magnitude implies that the torque applied by the servomotor was greater than that supplied voluntarily by the subject, and this situation proved difficult for the stroke subjects to control very well.

The reaching and tracking experiments were performed with the load applied in both the flexion and the extension directions. The experiments were repeated six times under each testing condition for statistical analysis purposes (the Mann-Whitney test, \( \alpha = 0.1 \)). In the tracking experiments, the Root Mean Squared error (RMS error) and the Integration of Square of Jerk (ISJ) measures were adopted as performance indicators of the tracking accuracy and the smoothness of movement, respectively. In these measures, the RMS represents the difference between the actual elbow trajectory and the required trajectory, while the jerk referred to within the ISJ measure was the third differential of the angle.

**Results**

Since this was the subjects’ first experience of using an assistive torque device to manipulate a load, most of them were initially unable to operate the system very well. Generally, they tended to exert an excessive force, and the corresponding trajectory of movement oscillated as they then attempted to exert an opposing force to regain the correct trajectory. Sensing that the subject was applying an opposing force, the servomotor also supplied an assistive torque in the same direction, which further exacerbated the problem. However, after performing three practice trials, it was found that all of the subjects were able to master the use of the device, and could proceed to the experimental trials.

**Reaching Experiments.** Figure 5 presents the agonist IEMG results of the able-bodied subjects with and without the assistive torque (i.e., \( G = 0 \% \) and \( G = 100 \% \), respectively). It is noted that the results on the left of the figure relate to movements in the flexion direction, while those on the right relate to movements in the extension direction. It can be seen that the application of an assistive torque decreases the value of IEMG significantly (\( p < 0.1 \)) in all cases other than the extension movements of subject AB1. Furthermore, during the experimental trials, when the assis-

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**Table 2 Values of \( K_t, K_b, M_t, \) and \( M_b \) for test subject S1**

<table>
<thead>
<tr>
<th>Percentage of maximal voluntary torque</th>
<th>Triceps</th>
<th>Biceps</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>( K_t )</td>
<td>( M_t )</td>
</tr>
<tr>
<td>16.7%</td>
<td>3.35</td>
<td>1.28</td>
</tr>
<tr>
<td>90 deg</td>
<td>2.8</td>
<td>1.46</td>
</tr>
<tr>
<td>115 deg</td>
<td>3.62</td>
<td>2.32</td>
</tr>
<tr>
<td>140 deg</td>
<td>3.49</td>
<td>2.38</td>
</tr>
<tr>
<td>50%</td>
<td>2.57</td>
<td>4.55</td>
</tr>
<tr>
<td></td>
<td>3.12</td>
<td>3.84</td>
</tr>
<tr>
<td></td>
<td>3.46</td>
<td>3.67</td>
</tr>
<tr>
<td></td>
<td>3.55</td>
<td>3.41</td>
</tr>
</tbody>
</table>

*Full elbow extension as 180 deg
†Unit: Nm
‡The processed, normalized EMG (dimensionless)

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**Fig. 4** The predefined trajectory in the tracking experiment. It is noted that the trajectory comprises a series of elbow movements including lifting a weight at a constant speed until the target position is attained, and then holding the weight for a specified period of time before lowering the weight at a slower speed and then holding it in position until the end of the trial.

**Fig. 5** The agonist IEMG of able-bodied subjects during the reaching experiment with assistive device activated (\( G = 100 \% \)) and with no applied assistance (\( G = 0 \% \)). It is noted that the results on the left of the figure relate to movements in the flexion direction, while those on the right relate to the extension direction.
tive system was active, the subjects provided subjective comments relating to the need to expend less effort in supporting the load.

Figure 6 indicates the effects of changing two parameters relating to nonlinear damping and adaptive filtering (i.e., $C_0$ and $\beta$, respectively) on the agonist IEMG and MPL. It is noted that each of the sub-figures also indicates the IEMG and MPL results when the assistive device is not activated. As shown in the upper sub-plots of Fig. 6, the agonist IEMG is reduced in all cases when the assistive device is employed. Conversely, from the lower sub-plots, it can be seen that there is a significant increase in the MPL when the assistive device is activated with parameters of $C_0=0$ and $\beta=1$. From the previous definition of MPL, it is clear that an increased MPL value implies a poorer movement performance, i.e., it suggests that a large angular displacement of the elbow is required to complete the reaching trial. Therefore, the results suggest that the proposed assistive system is ineffective in the case where no damping effect is applied and $\beta$ is specified as 1. However, it can be seen that as the damping effect is increased, the value of MPL falls. This implies that the subjects completed the reaching tasks with less effort and with a smaller angular displacement. Similarly, the results indicate that increasing $\beta$ also reduces the value of MPL. However, comparing the MPL results with those of the IEMG reveals that the improvement in smoothness of motion is paid for at the expense of an increased voluntary effort (i.e., the value of the IEMG rises as $\beta$ is increased from 1 to 4). On the basis of the results presented in Fig. 6, compromise values of the two parameters were chosen for the tracking experiments, i.e., $C_0=0.2$ and $\beta=2$.

Tracking Experiments. Figures 7(a) and 7(b) present the tracking results of a representative stroke subject, S3, as he moves his forearm in the extension direction during the ramp-up segment and then supports the load in the flexion direction. Note that Fig. 7(a) presents the results for the case where the assistive torque is not applied, while Fig. 7(b) provides the corresponding results when the device is activated. The upper plots within each figure show the required and actual trajectories of the elbow movement. Meanwhile, the lower plots depict the raw EMG signals of the triceps and biceps. It is noted that both sets of plots represent the average results of the six trials, and that the dotted lines in the upper plots indicate the range of one standard deviation. A comparison of the two figures reveals that the amplitude of the agonist EMG signal is decreased when the assistive device is activated, but that there is no significant difference in the antagonist EMG signal. Furthermore, it is noted that these is a delay at the beginning of both the ramp-up and the ramp-down segments. Comparing the upper plots in Fig. 7, it is noted that the standard deviation is smaller during the ramp-up segment of Fig. 7(b), which indicates that the subject moved his forearm more stably when the assistive torque was applied. In the ramp-down segment, it is observed that the movement of the forearm is smooth even though a significant tracking error exists.

Figure 8 presents a summary of the RMS errors, and the ISJ and IEMG results of the agonist and antagonist in both directions of movement for the ten test subjects. For all the subjects, the results show that the agonist IEMG value with a 100% assistive torque is significantly smaller than the IEMG value when the assistive torque is not applied. It is observed that there is no significant difference in the RMS error and the ISJ results for applied assistive torque gains of 0% and 100% in the case of the able-bodied subjects. However, a gain of 150% applied to the assistive torque results in a clear deterioration of the movement performance, as indicated by the increased RMS error and ISJ value. Regarding the stroke subjects, it is noted that the assistive system
tends to reduce the average RMS error, but has little effect upon the ISJ results. Finally, the results show that the antagonist IEMG increases in both the able-bodied and the stroke subjects as the assistive torque is increased.

Discussion

The current study uses the static EMG signals to construct the gain (\( K_1 \) and \( K_2 \)) maps. Although there is evidence to suggest that the static and dynamic EMG signals may be different, the tracking movements adopted during the experimental studies are relatively slow, i.e., 20 degress/s, and consequently, the dynamic influence is comparatively small. The results have indicated that the specified velocity is sufficient to improve the elbow capacities of stroke patients (i.e., Fig. 7 indicates that the stroke subject accomplished the tracking movement with less effort). However, it is acknowledged that the performance of the proposed control algorithm with higher velocity movements requires further study.

The results of the reaching experiment show that the assistive system reduces the exertion significantly, but, as shown in Fig. 6, it is clear that increasing the damping effect and decreasing the bandwidth of the adaptive filter results in an increased antagonist IEMG. Nevertheless, the appropriate choice of damping and adaptive filter parameters can increase the stability of the system, such that at the very least, the performance when using the assistive device can match the performance achieved with no applied torque. In the current experiments, the minimum value of \( \beta \) was specified as 1 since individuals were unable to operate the system if the adaptive filter was deactivated. The lower system bandwidth provides the advantage that the system is easier to operate, but has the disadvantage that it permits the subjects to exert an excessive force in an overly rapid movement. Therefore, in order to economize subject effort, the tracking experiments were performed using a value of \( \beta \) equal to 2. Similarly, the nonlinear damping parameter was specified as \( C_0 = 0.2 \) to ensure an improved movement performance.

The results of the tracking experiments reveal that the application of a 100% assistive torque does not cause a deterioration of movement performance in most of the able-bodied and stroke subjects as they lifted, held and deposited the weight. This result implies that the proposed assistive control algorithm is suitable for use under both concentric and eccentric contractions. It has been noted that the antagonist IEMG rises with increasing assistive torque, and this implies that the subject has to generate greater co-contraction effects during tracking in order to stabilize his or her elbow movement. Each subject underwent a maximum of two or three practice trials before completing the main experimental tasks. During this time, the subjects usually had to co-contract their biceps and triceps to maintain a smooth movement. However, during the course of the experiments, the sum of the agonist IEMG and antagonist IEMG was reduced as movement was conducted with the assistance of the applied torque device. Furthermore, it is expected that the performance of the both the able-bodied and the stroke subjects will improve as they gain more experience in operating the system. The results have indicated that when the gain of the assistive torque is increased to 150%, the system becomes increasingly difficult for the able-bodied subjects to control. In this situation, the assistive torque exceeds the torque supplied voluntarily by the subject, and this causes a deterioration in the subjects' tracking performance.

During the experimental trials it was noted that the contraction of the antagonist also caused a non-negligible change in the agonist EMG signal. Despite adding a nonlinear damping structure to the algorithm to increase the damping effect, it was observed that excessive co-contraction could lead to instability. In this situation, the subjects were reminded to relax, and were requested to concentrate upon reducing co-contraction in order to obtain a better performance.

Conclusion

The current study has developed a system to provide an assistive torque to a patient’s elbow which is proportional to the difference of the weighted EMG signals of the patient’s biceps and triceps. The experimental results have confirmed the effectiveness of the proposed system in developing a large apparent torque at the elbow joint if system stability can be achieved and if the patient is allowed sufficient practice time. A series of reaching experiments have been performed in order to determine the optimum values of the nonlinear damping and adaptive filter parameters. Using these parameter values, tracking experiments have confirmed that the proposed assistive torque system can increase the elbow torque output by 100% without impairing movement performance. Hence, the system enables stroke patients suffering from muscle weakness to move heavier objects or to exert force more efficiently. It is the intention of the current authors to extend the performance of the proposed system to applied assistive torques in excess of 100% of the voluntary torque in a future research study.

Acknowledgments

The authors gratefully acknowledge the support provided to this study by the ROC National Health Research Institute and by Taiwan’s NSC under contract numbers NHRI-EX90-9017EP and NSC 90-2213-E-006-070, respectively.

References