Development of a Closed-loop FES System Using a 3-D Magnetic Position and Orientation Measurement System

Kenji Kurosawa, Takashi Watanabe, Ryoko Futami, Nozomu Hoshimiya and Yasunobu Handa

Abstract—We have developed a closed-loop FES system using a magnetic 3-D position and orientation measurement system (FASTRAK, Polhemus Inc.). The purpose of this development was to resolve some experimental difficulties involved in our previous goniometer-based experimental system. The new system enabled us to perform FES control experiments on the multi-joint musculoskeletal system of the upper limbs including forearm pronation/supination. In this paper, we evaluated the system by some single-joint tracking tasks in order to compare its control performance with that of the previous system. Four muscles (ECRL(B), ECU, FCR, and FCU) of neurologically intact subjects were stimulated to control the wrist joint’s two degrees of freedom movement. Stimulation currents were determined by a multi-channel PID controller that was designed for a musculoskeletal system with redundancy (i.e. the number of muscles stimulated is more than that of the degree-of-freedom of the movement). The results showed that the system had sufficient control performance on tracking desired trajectories. Moreover, the system could compensate for unwanted external disturbances.

Index Terms — Functional Electrical Stimulation (FES), closed-loop control, PID controller, magnetic sensor.

I. INTRODUCTION

For individuals with spinal cord injuries, Functional Electrical Stimulation (FES) is an effective and reliable approach to restore motor functions. So far, many FES control strategies have been proposed to control paralyzed limbs: look-up tables [1,2], open-loop artificial neural network controllers [3,4], adaptive neural network controllers [5], model reference adaptive controllers [6], PID controllers [7,8], combinations of closed-loop and open-loop controllers [9,10], and so on. In comparison with open-loop schemes, closed-loop controllers have advantage in high stability and in high performance on disturbance compensation [7]. However, few attempts have so far been made for closed-loop multi-joint and multi-degree-of-freedom FES control.

We have been developing a closed-loop FES controller for a musculoskeletal system with redundancy (i.e. the number of muscles stimulated is more than that of the degree-of-freedom of the movement). In our prior work, we had developed a closed-loop FES system incorporating a multi-channel PID controller that could deal with such redundant control targets [8]. The FES system used an electric goniometer to sense joint angles. We have carried out experiments in controlling the wrist joint’s 2 degrees of freedom movement by stimulating 4 muscles, and reported that the PID controller had high potential for accurate control of the joint angles [8]. However, the system was insufficient to be applied to multi-joint targets and to measure forearm pronation/supination because of some physical problems with the angle sensor. For example, the sensor itself sometimes limited the range-of-motion of joints. Moreover, we needed the specific goniometer for measuring the forearm movement and several variation of the goniometer in size should be prepared for multi-joint control: for fingers, the wrist, the elbow, etc. The goniometer could sense only one or two degrees of freedom of a joint. It was troublesome for preparing and setting up several kinds of sensors. Therefore, we have developed a new system that can measure the forearm and multi-joint targets movement to progress the study on the FES controllers.

This paper describes our new closed-loop FES system. The new system utilizes a magnetic 3-D position and orientation measurement system (FASTRAK, Polhemus Inc.) to sense multi-joint angles including the forearm pronation / supination. Because the body motion is measured using the electromagnetic coupling, the joint movement is not interfered with. Furthermore, the system can measure the movement of any joints. As the first preliminary evaluation...
of our new system, we controlled movement of neurologically intact subjects’ wrist joint by stimulating 4 muscles. The control performance of the system will be compared with that of our previous goniometer-based system in Section 4. In addition, we examined disturbance compensation ability of the controller by giving external force or a static load to the FES-controlled wrists.

II. THE CLOSED-LOOP FES SYSTEM

A. Hardware

Figure 1 shows the hardware block diagram of the system. The system is composed of a magnetic 3-D position and orientation measurement system (FASTRAK, Polhemus Inc.), a personal computer (PC), a pulse modulator, and isolators. The FASTRAK system consists of a fixed magnetic-dipole transmitting antenna called a transmitter and freely movable magnetic-dipole receiving antennas called receivers [11]. The outputs of the FASTRAK are digital data showing the position and orientation of the receivers, which are transferred to the PC through the RS-232C serial port at 115.2k-baud rate. The PC calculates joint angles in real-time from the outputs of the FASTRAK (details are described in the following section). Then, multi-channel amplitude of electrical stimulation is determined on the PC from the difference between the desired and the actual movement trajectory by the PID-controller. The multi-channel amplitude data for electrical stimulation is transferred to the pulse modulator through a digital I/O board (PIO-48D (PCI), Contec Inc.). The pulse modulator was newly designed for our new system. The pulse modulator consists of a one-chip microcomputer IC (H8/3048), a 12bit D/A converter IC, and an analog demultiplexer. The pulse modulator converts the digital data describing the stimulus amplitude into trains of analog stimulation pulse (the Pulse Amplitude Modulation (PAM) waves, the pulse frequency is 20Hz, the width of each pulse is 200µs). We adopted this master (PC) – slave (pulse modulator) strategy in order to make the system keep the accurate stimulation pulse frequency and the pulse width while the PC is communicating with the FASTRAK. The outputs of the pulse modulator are applied to subjects’ each muscle through the isolators (5384, NEC Medical Systems Inc.) and Ag/AgCl surface electrodes. The system can stimulate up to 8 muscles independently (16 muscles, if some ICs are added to the pulse modulator), and the maximum angle measurement frequency is 40Hz (2 joints) or 30Hz (3 joints). It must be noted that there is a delay of
several milli-seconds in the FASTRAK for each receiver to sense and calculate the 3-D position and orientation of the receiver.

B. Calculation of Joint Angles
The FASTRAK measures the relative position and orientation of the receivers on the basis of the transmitter in 3-D space. The static accuracy is 0.08cm RMS for the receiver position and 0.15 degree RMS for the receiver orientation, when the receivers are located within 76cm from the transmitter [11]. The FASTRAK can handle up to 4 independent receivers, and it outputs 6 parameters for each receiver (i.e. three are for the position and three for the orientation). The three output parameters showing the Euler orientation angles (azimuth, elevation, and roll) of each receiver are used to calculate joint angles, as follows. As shown in Fig.2(a), the transmitter is fixed at arbitrary position near the subject’s body, and each receiver is attached on subject’s skin of each body segment (the upper arm, the forearm, the hand, fingers, etc.). In Fig. 2(a), three coordinate systems are defined: the transmitter coordinate system T (X, Y, Z), the receiver-1 coordinate system Rec1 (x1, y1, z1), and the receiver-2 coordinate system Rec2 (x2, y2, z2). The three out of six output parameters of the FASTRAK describes the orientation (azimuth (ψ), elevation (θ), and roll (φ)) of the Rec1 (Rec2) coordinate system on the basis of the T coordinate system (Fig. 2(b)), and only the rotational parameters are used to derive the relative angular parameters (ψ12, θ12, φ12). The valuables, ψ12, θ12, and φ12 are the relative angular parameters between Rec2 and Rec1. Here, we define three rotation matrices as follows:

\[ R_1 \]: The rotation matrix of the Rec1 coordinate system around the axes in the T coordinate system
\[ R_2 \]: The rotation matrix of the Rec2 around the axes in the T coordinate system
\[ R_{12} \]: The rotation matrix of the Rec2 around the axes in the Rec1.

The rotation matrix is expressed by [12]:

\[
R_n = \begin{bmatrix}
\cos \psi_n \cos \theta_n & \sin \psi_n \cos \theta_n & -\sin \theta_n \\
-\sin \psi_n \cos \phi_n + \cos \psi_n \sin \phi_n \sin \theta_n & \cos \psi_n \cos \phi_n + \sin \psi_n \sin \theta_n \sin \phi_n & \sin \phi_n \sin \theta_n \\
\sin \psi_n \sin \phi_n + \cos \psi_n \sin \theta_n \cos \phi_n & -\cos \psi_n \sin \phi_n + \sin \psi_n \sin \theta_n \cos \phi_n & \cos \phi_n \sin \theta_n
\end{bmatrix}
\]

Here, \( \psi_n \), \( \theta_n \), and \( \phi_n \) are the azimuth, elevation, and roll angles of the Rec n coordinate system around the axes in the T coordinate system, as defined in Fig.2(b).

These rotation matrices have the relation:

\[ R_{12} = R_2 R_1^{-1} \]  \( \text{(1)} \)

Thus, the orientation of the Rec2 coordinate system on the basis of the Rec1 is calculated as follows:

\[
\begin{align*}
\psi_{12} &= \begin{cases} 
\mu : u \in \cos^{-1} (r_1 / \cos \theta_{12}) \quad \text{and} \\
\nu : v \in \cos^{-1} (r_2 / \cos \theta_{12}) 
\end{cases} \quad (180 \leq \psi_{12} < 180) \\
\theta_{12} &= -\sin^{-1} r_3 \quad (-90 \leq \theta_{12} < 90) \\
\phi_{12} &= \begin{cases} 
\mu : u \in \cos^{-1} (r_6 / \cos \theta_{12}) \quad \text{and} \\
\nu : v \in \cos^{-1} (r_5 / \cos \theta_{12}) 
\end{cases} \quad (180 \leq \phi_{12} < 180)
\end{align*}
\]

These parameters can be obtained uniquely. Therefore, we can obtain 3-D joint angles between segment 1 and 2.

C. Feedback Controller
In this study, the PID controller defined as equation 3 was used to determine the stimulation current vector \( I_s \):

\[ I_s = I_{th} + K_P e_n + K_I \sum_{i=0}^{n} e_i + K_D (e_n - e_{n-1}) \]  \( \text{(3)} \)

where \( I_{th} \) is the threshold vector. The error vector \( e_n \) is defined by \( e_n = \theta^{\text{desired}} - \theta^{\text{measured}} \) (\( \theta^{\text{desired}} \) and \( \theta^{\text{measured}} \) are the desired and the measured joint angle vector at time \( n \)). The modified CHR (Chien, Hrones and Reswick) method are used to determine the PID parameter matrices \( K_P \), \( K_I \), \( K_D \) as equation 4 [8]:

\[
K_{PI} = 0.6 \frac{L}{T_i} m_{ij}, \quad K_{II} = 0.6 \frac{L}{T_i} m_{ij}, \quad K_{DI} = \frac{0.37}{\Delta t} m_{ij} \]  \( \text{(4)} \)

where \( L_i \) and \( T_i \) are the latency and the time constant of the step response of muscle \( i \), and \( \Delta t \) is the sampling period. The coefficient \( m_{ij} \) is an element of a generalized inverse matrix of a transform matrix \( M \). The matrix \( M \) is the Jacobian matrix describing the relationship between a stimulation current vector and a joint angle vector of a musculoskeletal system [13]. The relationship between the small displacement in the joint angle vector (\( d\theta \)) and in the stimulation current vector (\( dI \)) can be expressed using the numerical constant matrix \( M \) by the linear approximation [8,13]:

\[ d\theta = M dI \]  \( \text{(5)} \)

The matrix \( M \) is obtained experimentally by applying the ramp stimulation to each muscle and by measuring its response. The inverse of \( M \) means the transform matrix from the joint angle vector into the stimulation current vector. However, the inverse matrix does not exist in general because \( M \) is not a square matrix (i.e. usually, the number of muscle is more than that of the degree-of-freedom of the movement). Therefore, we introduced a generalized inverse matrix \( M^{-1} \) instead of \( M^{-1} \). The \( M^{-1} \) can be calculated uniquely under the limitation of the sign of its elements and the constrained least square criterion [14]. The matrix \( M^{-1} \) is used in Eq.4, \( m_{ij} \) is the element of the \( M^{-1} \). Thus, this coefficient works
as the transformer from the error into the current. The PID parameters were kept constant throughout the experiment.

III. EXPERIMENTAL METHOD

We performed experiments to examine the developed system and the controller. Although the system can handle a multi-joint musculoskeletal system, we simply controlled the wrist joint’s 2 degrees of freedom movement (the dorsi/palmar-flexion and the radial/ulnar-flexion) to compare its control performance with our experimental results obtained with the previous goniometer-based system. Four muscles (the flexor carpi radialis [FCR], the flexor carpi ulnaris [FCU], the extensor carpi radialis longus/brevis [ECR], and the extensor carpi ulnaris [ECU]) of 3 or 2 neurologically intact subjects were electrically stimulated. The stimulation was applied to each muscle with Ag/AgCl surface electrodes (F-150M, Nihon Koden). The stimulation frequency was 20Hz, and the pulse width was fixed at 200μs.

Three FASTRAK receivers were attached on the subjects’ left upper arms, the forearms, and the back of the hands to measure not only the wrist joint angles but also the forearm pronation/supination angle. In this study, the forearm movement was not controlled but measured in order to observe the synchronized movement of the forearm with the wrist. This could not be measured with the previous system. The angles were sampled at 20Hz. The subjects were sat on a chair with their left arms hanging (i.e. to the gravitational direction); the shoulder, the elbow, and the hand were free in position. The subjects were instructed to relax their left arms and the hands as much as possible. The neutral angle of the wrist was defined at this posture. All the subjects were well trained for electrical stimulation before the experiments to avoid interference of the voluntary movement. Three types of experiments were carried out.

A. Trajectory Tracking Test (3 Subjects)
The wrist joints were controlled with the system to track desired trajectories, which were elliptical (the cycle period was 10s or 3s), rectangular (12s), and linear (the direction was both the dorsi/palmar-flexion and the radial/ulnar-flexion, 10s). The elliptical trajectory was a smooth curve, whereas the rectangular trajectory was a combination of straight movement for 2s and positioning control for 1s. The linear trajectories were reciprocating motions. The desired trajectories were defined on the 2-D joint angles plane [cf. the dashed curve in Fig. 3(a)]. The results were evaluated in the mean absolute tracking (MAT) error:

$$MAT = \frac{1}{N} \sum_{i=0}^{N} |p_{i}^{(D)} - p_{i}^{(M)}| [\text{cm}] \quad (6)$$

where \(p_{i}^{(D)}\) and \(p_{i}^{(M)}\) are the desired and the measured end position vector at time \(i\). \(N\) shows the number of samples. The error measured in the angle space was transformed into the spatial displacement of the end point of the middle finger, because the error in the spatial displacement is more important on clinical applications, even though the desired trajectory was given in the angle space. The error was compared with the results obtained with our previous system.

B. Compensation Test for Impulsive External Disturbance (2 Subjects)
As mentioned in the section 1, the closed-loop controllers are robust against unexpectable disturbance. As the second test, impulsive external force was given to the hands while the positioning and the tracking control in order to examine whether or not our controller could eliminate this type of disturbance. The force was given by pushing the hands quickly so as to change the posture. The duration of the force was almost 0.2s.

C. Compensation Test for a Static Load (2 Subjects)
The third test was compensation for a static load. A ceramic cup, whose weight was 300g, was tied to the palm of the subjects’ hands. Under this condition, the wrists were controlled to track the elliptical desired trajectory whose cycle period was 10s. The error values were compared with the results under the load-free condition.

IV. RESULTS

A. Trajectory Tracking Test
The system could control all the subjects’ wrist successfully. The realized trajectories were almost same as the desired trajectories. Figure 3 shows one of the results in the case of the elliptical trajectory whose cycle period was 10s (Subject B). Fig.3(a) is the 2-D trajectory of the wrist, Fig.3(b) shows the stimulus pulse currents, and Fig.3(c), 3(d), and 3(e) are the time response of each angle. From these figures, one can see that the wrist joint could be controlled almost along the desired trajectory. We measured pronation/supination angle of the forearm [Fig.3(e)], which could not have been obtained with our previous system. The synchronized movement of the forearm with the wrist was observed. The error values of all results were summarized in Table I. The error values shown in Table 1 are the spatial displacement at the end point of the middle finger, whose unit is “cm”. We can see that the mean absolute tracking (MAT) errors were less than 1cm in all cases unless the movement was fast (3s/cycle). The errors were tripled when the movement velocity of the elliptical trajectory was increased to 3s/cycle. In the results of 3s/cycle movement, delay in the response was observed. Thus, we think this is the limitation in speed for the controller. These results were almost same as the results obtained with our previous goniometer-based system as shown in Table II [13]. Therefore, the results indicate that the new system is not inferior to the previous system on control accuracy, although oscillation was partly observed in Fig.3. Probably, the oscillation could be repressed if the parameters were optimized by trial and error. In this study,
the PID parameters were determined by equation 4, and they were kept constant throughout the experiments.

B. Compensation Test for Impulsive External Disturbance

Figure 4 shows the typical result of this test. The wrist was controlled to the desired posture (10 degree in the radial-flexion, 20 degree in the dorsi-flexion) in one second at the beginning of the control. Then, the posture was maintained for 29 seconds by the closed-loop controller. After the control became stable, an external force was given to the back of the hand at the time shown with the arrows [in Fig. 4(b), 4(c)]. We can see that the controller compensated the disturbance by changing the stimulation currents. This change achieved the rapid recovery of the wrist angles. However, some overshoot was observed before the response became stable. From Fig.4(b) and 4(e), it is observed that the currents and the forearm angle changed after the impression of the force.

This is discussed in the section V. We also gave the external force to the hands during the trajectory tracking control. The disturbance was compensated well. The wrist joints were pulled back to the desired trajectory within a second.

---

**TABLE I**

<table>
<thead>
<tr>
<th>Trajectory → Subject</th>
<th>Elliptical 10s</th>
<th>Elliptical 3s</th>
<th>Rectangular Linear (dorsi/palmar)</th>
<th>Linear (radial/ulnar)</th>
<th>Elliptical (300g Load)</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>0.61</td>
<td>1.86</td>
<td>0.85</td>
<td>0.55</td>
<td>0.57</td>
</tr>
<tr>
<td>B</td>
<td>0.52</td>
<td>1.42</td>
<td>0.75</td>
<td>0.35</td>
<td>0.46</td>
</tr>
<tr>
<td>C</td>
<td>0.92</td>
<td></td>
<td>0.87</td>
<td>0.76</td>
<td>0.76</td>
</tr>
</tbody>
</table>

**TABLE II**

The mean absolute tracking error [13] (GONIOMETER, UNIT: CM)

Results of two trajectories are shown. These are the averaged values of three trials.

<table>
<thead>
<tr>
<th>Trajectory → Subject</th>
<th>Elliptical cycle period 5s</th>
<th>3s</th>
<th>Rectangular</th>
</tr>
</thead>
<tbody>
<tr>
<td>S</td>
<td>0.79</td>
<td>1.18</td>
<td>1.71</td>
</tr>
<tr>
<td>I</td>
<td>0.57</td>
<td>1.08</td>
<td>1.70</td>
</tr>
</tbody>
</table>

---

Fig.3 The result of the trajectory tracking test. The desired trajectory was elliptical (cycle period 10s). (a) 2-D trajectory of the wrist joint, (b) Amplitude of the stimulus pulse currents, (c) Time response of the wrist angle (radial/ulnar flex.), (d) Time response of the wrist angle (dorsi/palmar flex.), (e) Time response of the forearm angle (pronation/supination). The graph of Fig. 3(a) has 2 axes, which are the radial/ulnar-flexion (horizontal) and the dorsi/palmar flexion (vertical). The horizontal axes of the other graphs are the time in second.
C. Compensation Test for a Static Load
Under the condition of the 300g static load, the controller could compensate the weight of the load. The measured trajectories were close to the desired trajectory. However, overshoot was observed when the wrist was palmar flexed. The error values under this condition were also shown in Table 1. The errors were also less than 1cm, but they were slightly larger than the case of no load.

V. DISCUSSION
In order to progress the study on the FES controller, we have developed the closed-loop FES system incorporating the FASTRAK system. This effort was made because our previous goniometer-based experimental system was not suitable for studying the controller on a multi-joint target including the forearm movement. The new system can measure up to three adjacent joint movements, because the FASTRAK can handle four independent receivers.

A pair of receivers gives the 3-D angular parameters of a joint. The sensor does not interfere with the joints movement because the electromagnetic field is used to sense the movement. The experiments of the FES controller for the multi-joint targets will be easily carried out with the system.

However, the geometrical arrangement of the transmitter and the receivers should be carefully considered, because the noise component produced by the FASTRAK could be magnified through the calculation process of the joint angles under some conditions. This magnification of the noise component may cause unstable control. It is considered that the high frequency component of the realized trajectories shown in Fig.3 and Fig.4 were due to the noise produced by the FASTRAK system, because such high frequency component was not observed with the goniometer-based system.

The disadvantage of the system is the limitation in measurement environment. Volumes of metal such as cabinets may affect the measurement accuracy, because the FASTRAK uses the electromagnetic field to sense the receiver position and orientation. It is recommended by the manufacture that the receivers and the transmitter are located far from a volume of a metal object, at a distance of twice as long as the measurement range. Because we kept this criterion during the experiments, we think the effect was negligible in our results.
From Fig. 4(c) and 4(d), it was confirmed that the controller could compensate for the impulsive disturbance. However, one can see from Fig. 4(b) that the stimulation currents were increased in all channels once the external force was given to the hand. This means that although the controller could regulate joint angles to the desired posture immediately, this control might change the stiffness of the joint. We consider that this was partly because the constrained least square condition for obtaining $M^{-1}$ did not minimize the total amount of the currents but minimized the amount of the change in the currents. Thinking the clinical usage, this problem is not negligible. Development of the controller that can control both joint angles and joint stiffness is required. Moreover, from Fig. 4(e), it was observed that the forearm angle also changed after the impulse of the force. There seemed to be the relation between the change in the current and in the forearm angle. This was one of the important results obtained with the system.

In the trajectory-tracking tests, we prepared the desired trajectories of three different types. They were elliptical, rectangular, and linear. The new system could control the wrist along all the trajectories with the accuracy of mm-orders (less than 1 cm) unless the velocity was 3 s/cycle. This demonstrates that the controller has high potential for accurate control of the musculoskeletal system with redundancy if the motion is slow. However, the error increased three times as the movement velocity increased to 3 s/cycle. Further improvement of the controller will be required for a quick movement. It seems realistic to switch controllers according to the movement velocity and the nature of desired trajectories.

In the PID controller, we introduced the static transform matrix $M$ for conversion of the error vector into the current vector. The matrix was made on the basis of the linearization technique. The controller with the static matrix worked well in our experiments. We think this was partly because the control targets had relatively linear property in their response. However, for a control in a wider range of angles, it may be better to recalculate the matrix in real-time according to the operating point. Moreover, it is desirable to take into account the effect of the other joint’s angle for calculating the parameters, because the change in the elbow angle causes the change in the property of the bi-articular muscles and the effect of the gravity on the wrist changes as the shoulder moves.

The closed-loop FES system described in this paper was designed for laboratory experiments. Thinking the clinical use, more simple and economic sensors should be developed. In the control experiments, we directly gave the reference trajectories to the controller. However, it is difficult for patients to generate and give such detailed trajectories by themselves. Development of a proper user interface and an interpreter that can transform simple patients’ commands into the reference trajectories for the sophisticated FES controller is also one of the important subjects for the FES researchers.

VI. CONCLUSION

We have developed the new closed-loop FES system using the magnetic 3-D position and orientation measurement system (FASTRAK) in order to make it possible to study the FES controller on multi-joint targets. We evaluated the system by some tracking tasks on neurologically intact subjects. The results showed that the system had sufficient control performance as well as our previous goniometer-based system. It was also shown that the system could compensate for the unwanted external disturbances.

Although the system can handle a multi-joint musculoskeletal system and forearm pronation/supination, the preliminary evaluation was performed on the simple wrist joint’s 2 degrees of freedom movement in this paper. Our future works are to examine our controller with the system on the 3 degrees of freedom movement of both the wrist and the forearm, and on a multi-joint musculoskeletal system. The development of the controller that can deal with both the joint angles and the joint stiffness is also an important subject. An evaluation of the methods in subjects with motor disabilities is also required before the clinical applications [15].

REFERENCES


