Elicited Finger and Wrist Extension through Transcutaneous Radial Nerve Stimulation

Yang Zheng and Xiaogang Hu

Abstract— Individuals with neurological disorders, such as stroke or spinal cord injury, often have weakness and/or spasticity in their hand and wrist muscles, which can lead to impaired ability to extend their fingers and wrists. Functional electrical stimulation can help restore these motor functions. However, the conventional stimulation method can lead to fast muscle fatigue and limited movements due to a non-physiological recruitment of motor units and a limited recruitment of deep muscles. In this study, we investigated the feasibility of eliciting various hand opening and wrist extension movement patterns through a transcutaneous electrical stimulation array, which targeted the proximal segment of the radial nerve bundle proximal to the elbow. Wrist and finger joint kinematics were used to classify the different movement patterns through a cluster analysis, and electromyogram signals from the wrist and finger extenders were recorded to investigate the muscle activation patterns. The results showed that finger and wrist motions can be elicited both independently and in a coordinated manner, by changing the stimulation intensity and stimulation location. H-reflex activity was also observed, which demonstrated the potential of recruiting motor units in a physiological order. Our approach could be further developed into a rehabilitative/assistive tool for individuals with impaired hand opening and/or wrist extension.

Index Terms— Transcutaneous electrical nerve stimulation, Radial nerve stimulation, Hand function, Hand opening, Wrist extension,

I. INTRODUCTION

A majority of individuals with injuries to the central nervous system (CNS), such as a stroke or a spinal cord injury, have impaired motor function, which limits their activities of daily living. The impairment is mainly manifested as paralysis in skeletal muscles [1-3]. Among the different motor and sensory functions for individuals with paralysis, regaining arm and hand function is the first priority to improving the quality of life [4]. In addition, spasticity is a common clinical condition after CNS injuries [5], which can interfere with movement and hamper the recovery process. For the upper extremity muscles, the most frequent pattern of spasticity leads to hyper-flexed wrist and fingers [6]. As a result, a combination of extensor muscle weakness and flexor spasticity severely limits hand opening [7, 8]. Therefore, approaches that can restore the function of hand opening are needed urgently.

Functional electrical stimulation (FES) can be used as an assistive and therapeutic approach to help restore motor function [9, 10]. Recent studies have utilized FES to assist individuals to perform hand opening and/or grasp motions using multi-electrode stimulation technique [11-13]. In these studies, stimulation electrodes are placed on the muscle belly near the motor points. Muscle contractions are elicited by activating the distal branches of the motor axons. In this case, muscle activation patterns are non-physiological, which limits wide application of FES. First, superficial muscles are activated first [14], and strong stimulation intensity is needed to activate deep muscles, which can cause discomfort. The limited capability to activate deep muscles can also constrain movement kinematics. In addition, motor units (MUs) are recruited in a non-physiological order, with the large and fast-fatigable MUs recruited earlier [15] or different MUs recruited in a random order [16]. This can result in fast decline of elicited muscle forces during FES.

Different techniques have been developed to address these issues. For example, multiple electrodes are placed at different locations of the muscle to sequentially activate different muscle fibers [17-19], which can delay the force decline. Instead of activating the muscle directly through motor axons, stimulation of the afferent fibers to induce reflex activation can lead to physiological recruitment order of MUs [20, 21]. Recently, transcutaneous electrical stimulation at the proximal segment of the median/ulnar nerve bundles has been used to elicit hand grasp patterns [22, 23], and different finger and wrist flexion movements can be elicited by changing the stimulation location and intensity. Muscle fatigue is also significantly delayed compared with conventional FES [24]. The median and ulnar nerve bundles travel superficially along the humerus, thereby making them readily accessible from the skin surface. In contrast, the radial nerve bundle wraps around the humerus deep into the muscles with only a short segment relatively close to the skin surface, which imposes a challenge to activate the radial nerve through transcutaneous electrical stimulation.

Accordingly, the purpose of the current study was to develop a transcutaneous nerve stimulation approach to activate the radial nerve. Specifically, current was delivered to the radial nerve bundle proximal to the elbow near the radial groove, which was different from conventional FES with stimulation current delivered to the muscle belly that targeted the distal
branches of the motor nerves. The elicited finger and wrist kinematics were measured, and high-density electromyography (HD-EMG) were used to capture the activation patterns of finger and wrist extensors. With the stimulation array electrodes placed near the radial nerve bundle, different axons of the radial nerve can be activated, thereby leading to different hand opening and/or wrist extension patterns. Our results showed that the stimulation technique was able to generate different finger and wrist movements both independently and in a coordinated manner. Muscle activities can be elicited through the activation of both the motor axons (M-wave) and the sensory axons (H-reflex). The latter can lead to a more physiological recruitment order of MUs. In addition, the same motion can be elicited by activating different muscles or portions of muscles. This redundancy can help reduce muscle fatigue by alternating activation of different muscle fibers. These findings demonstrated that the proximal radial nerve stimulation can improve muscle activation and can facilitate the development of non-invasive nerve stimulation techniques to help restore functional hand/wrist movements of individuals with muscle weakness/spasticity.

II. METHODS

A. Subjects

Eight subjects (21–34 years of age, one female and seven males) without any known neurological disorders were recruited in the study. All subjects gave informed consent with protocols approved by the Institutional Review Board of the University of North Carolina at Chapel Hill.

B. Experiment

1) Apparatus and data recording

Twenty-five reflective markers with a 6.4-mm diameter were attached on the dorsal side of hand to measure the angles of the wrist (W) flexion/extension, the metacarpophalangeal (MCP) and proximal interphalangeal (PIP) flexion/extension of the index (I), middle (M), ring (R), and pinkie (P) fingers. Four 9.5-mm markers were attached on the dorsal side of wrist as the base. The markers were tracked using an 8-camera OptiTrack System (Natural Point, Inc, Corvallis, OR), and their 3D position data were recorded using the accompanied Motive software with a sampling frequency of 100 Hz.

An 8×16 HD-EMG electrode array, with a 3-mm electrode diameter and a 10-mm inter-electrode distance was used to record the activities of the extensor digitorum communis muscles and the extensor carpi ulnaris muscle during finger and wrist extension (Fig. 1). Monopolar EMG signals were recorded with the reference electrode placed at the wrist, and the ground electrode placed at the elbow via EMG-USB2+ (OT Bioelettronica, Torino, Italy). The EMG signals were amplified with a gain of 200 and a pass band of 10-900 Hz, and sampled at 5120 Hz.

Monophasic current pulses were generated with a multi-channel stimulator (Multichannel Systems, Reutlingen, Germany). The anode and cathode of the stimulator channel were connected to the rows of a switch matrix (Agilent Technologies, Santa Clara, CA), of which the columns were connected to 11 stimulation electrodes. The cathode electrodes were arranged in a 2×4 array and placed at the radial groove, and the three anode electrodes were placed beneath the short head of the biceps brachii at the medial side of the upper arm (Fig. 1). The stimulator and the switch matrix were controlled using a custom MATLAB user interface, and stimulation trains can be delivered to any cathode and anode pair. Clustered 80 μs subthreshold current pulses with 10-kHz carrier frequency (termed pulse clusters) used in our previous study [25] were delivered to activate the radial nerve, because reduced muscle fatigue was observed for both intact subjects [25, 26] and stroke survivors [27]. Each 80-μs current pulse can only induce subthreshold depolarization of the axon membrane potentials, and temporal summation effect of multiple pulses were needed to induce an action potential. The effective stimulation frequency was 30 Hz, i.e., 30 pulse clusters per second.

2) Stimulation paradigm

During the experiment, the subjects were seated in an arm chair. The forearm was supported in the pronated position with a soft foam pad on the desk (Fig. 1). Before data recording, the cathode and anode pairs that can elicit different finger and/or
wrist motions with no pain sensation were identified through a searching procedure across all cathode and anode combinations. For a given electrode pair, the stimulation amplitude was adjusted to induce moderate or strong muscle contractions, and fifteen 3-second stimulation trains were delivered with a 2-second resting interval for each trial. The wrist and fingers flexed naturally before stimulation onset. A rest time of 1 minute was provided between trials to avoid possible muscle fatigue. The recorded angles of individual joints and the EMG signal with the maximum amplitude from a representative trial are illustrated in Fig. 2.

C. Data processing

1) Joint kinematics

Feature extraction. The PIP, MCP and wrist joint angles were defined as shown in the bottom right insert of Fig. 1. The time courses of individual joint angles were segmented into 4.5-second segments between 0.5 second prior to the onset and 1 second after the end of individual 3-second stimulation trains. Then, the average trajectories of individual joints were obtained by averaging across all the 15 segments to represent the hand/wrist motion of each trial. To standardize the motions from all subjects, feature vectors were extracted from the joint trajectories of individual trials through the following procedures. First, the initial joint angles \( I_j \) \((j=1, 2, \ldots, 9)\) were calculated as the mean value within the first 0.5 second for individual joints, and the initial joint angles were subtracted from individual joint trajectories. Then, the joint angle \( X_j \) \((j=1, 2, \ldots, 9)\) corresponding to the maximum value of the trajectories (absolute value) was selected. According to the definition of joint angles (Fig. 1), if \( X_j > 0 \) \((X_j < 0)\), the corresponding joint extended (flexed). The features for individual joints were then calculated by normalizing the joint angle \( X_j \) with the corresponding joint range of motion:

\[
A_j = \left(\frac{\text{sign}(X_j) + 1}{2}\right)X_j / |I_j - E_j| + \left(\frac{\text{sign}(X_j) - 1}{2}\right)X_j / |I_j - F_j|
\]

(1)

where \( F_j \) and \( E_j \) \((j=1, 2, \ldots, 9)\) were the joint angle corresponding to full flexion and full extension, respectively. \( \text{sign}(\cdot) \) was the signum function. In this case, the feature values were within \([-1, 1]\), where -1 meant full flexion, 1 meant full extension, and 0 meant no motion. Lastly, these values from all 9 joints were combined to construct the feature vector for individual trials.

Cluster analysis. The feature vectors from all subjects were pooled together before cluster analysis. The hierarchical clustering algorithm was then used to categorize the feature vectors into different clusters. The Euclidean distance was used to measure the similarity between different vectors, and then the vectors were grouped into a binary, hierarchical cluster tree by linking pairs of vectors in close proximity. The cluster tree was then naturally divided into distinct, well-separated clusters by setting a threshold of the inconsistency coefficient, which was calculated as the difference between the height of a given link and the mean height of all the links below the given link, normalized by the standard deviation. A larger inconsistency coefficient meant the similarities between the two clusters connected by the link decreased abruptly.

Motion variability. In order to quantify the variability of the motions elicited, the variation of the induced joint range of motion was calculated for individual trials. Specifically, the joint range of motion was first calculated as the absolute difference between the average joint angle 0.5 second prior to the stimulation and the average joint angle from 2 to 3 second after stimulation onset (maximum change in angles) for individual stimulation trains and individual joints. The coefficient of variation (CV) of the joint range of motion was calculated for individual joints. Then, the CV was averaged across all joints to represent the overall variability of motions for individual trials.

2) EMG activity

The EMG segments that started 5 ms prior to and ended 33 ms after the onset of individual stimulations (i.e., pulse clusters) were extracted, and the segments that contained motion artifacts were excluded from further analysis. Stimulation artifacts were then identified manually, and the EMG segments between two stimulation artifacts were extracted to calculate the EMG amplitude.

To investigate the change of the M-wave and H-reflex amplitude over successive stimulations within the 3-second stimulation trains, the extracted EMG segments were then separated into M-wave and H-reflex activity with the time division point of 17 ms after the stimulation onset. The amplitude of the M-wave or H-reflex was calculated as the difference between the maximum of local maxima and the minimum of local minima of the M-wave or H-reflex. The M-wave amplitude and the H-reflex amplitude was averaged across all 128 channels and then averaged across all stimulation trains and all trials for individual stimulations.

In order to investigate the overall amplitude of EMG activities, the EMG segments were averaged across all stimulations within each trial to obtain the average EMG of individual channels. The average EMG between two stimulation artifacts were extracted to calculate the EMG amplitude as the difference between the maximum and minimum of the average EMG. The amplitude of the M-wave and the H-reflex was also extracted from the average EMG with the same method described above.

III. RESULTS

A. Motion types and variability

The average initial joint angle for the wrist was 158°±13° (mean ± standard deviation) across all trials from all subjects. The average initial joint angle of the MCP joint of the index, middle, ring, and pinkie fingers were 151°±9°, 150°±9°, 156°±9°, and 158°±8°, respectively. The average initial joint angle of the PIP joint of the index, middle, ring, and pinkie fingers were 145°±11°, 136°±13°, 135°±13°, and 143°±12°, respectively.
Ten motion types (clusters) were obtained through the cluster analysis using the feature vectors from all subjects (Fig. 3). The lines in the polar coordinates represent the normalized joint angles by the range of motion of different motion types. The origin of the coordinates represents full flexion of joints, and the inner and the outer rings represent the initial position and full extension of joints, respectively. Video clips showing the different motion types are included in the Supplementary Material. A number of motion types involved both finger and wrist movements. For example, in motion type 5, wrist extension was elicited with combined finger flexion. In motion type 10, the wrist and MCP joints extended while the PIP joints flexed. Motions of independent joints/fingers and coordinated multiple joints were also elicited. In motion types 4 and 6, extension of independent pinkie and index fingers was elicited, respectively. In motion type 1, index and middle fingers were mostly extended. Motion type 3 involved full hand opening and wrist extension.

![Fig. 3](image_url)

Fig. 3. Normalized joint angles by the full range of motion (feature values) of different motion types after the cluster analysis (a). The corresponding postures at the end of the stimulation (b). Wrist-flexion/extension (W-FE), index (I), middle (M), ring (R), and pinkie (P) fingers.

Fig. 4a illustrates the motion types of individual subjects. The average number of motion types that can be elicited for individual subjects was 3.63±0.92 with a range from 3 to 5. The average number of subjects for individual motion types was 2.90±1.37 with a range between 1 and 5. Both motion types 3 and 7, which involved wrist and all finger extension, can be observed in 5 subjects. Motions of simultaneous wrist and at least one finger extension can be observed in all subjects except for subject 5. Fig. 4b illustrates the CV of the elicited joint range of motion for individual subjects. The CV were largely observed in the range of 0.1-0.2. The average CV across all subjects was 0.14±0.07. Fig. 4c illustrates the stimulation intensity used for individual subjects. The current amplitude was largely within the range of 4-8 mA. Most of the subjects had current range within 1-2 mA; however, a large current range (>4 mA) was observed in subjects 5 and 8.

B. EMG activity

1) M-wave and H-reflex

Fig. 5a illustrates the EMG signals of the channel with the maximum amplitude from a representative trial. The stimulation onset was at time 0. M-wave started approximately 7 ms after the stimulation onset, and the H-reflex activities started approximately 22 ms after the stimulation onset. Compared with the amplitude of the M-wave, the amplitude of the H-reflex was small. The comparison of the average amplitude across all trials between the M-wave and the H-reflex of individual subjects is shown in Fig. 5b. The results of the signed-rank test showed that the amplitude was significantly different between the M-wave and the H-reflex activity for subject 1, 2, 4, 6, 7, and 8 (p<0.05), while the differences were not significant for the other two subjects.

Fig. 5c shows the amplitude distribution of the M-wave and the H-reflex from a representative trial. The amplitude of the H-reflex was smaller than that of the M-wave. Meanwhile, compared with the amplitude distribution of the M-wave, the location with the strongest H-reflex activity shifted towards the left, which demonstrated that different muscles or muscle portions were activated through direct stimulation of the motor axons and the stimulation of the sensory axons. In order to quantitatively compare the muscle regions activated through the two pathways, the 2D correlation coefficient of the amplitude distribution between the M-wave and the H-reflex was calculated for individual trials. Fig. 5d illustrates the correlation coefficient values of all trials with a median of 0.25 and the 25th and 75th percentile at 0.0644 and 0.5176, respectively.

2) M-wave and H-reflex amplitude over time

Fig. 6 illustrates the average amplitude of the M-wave and H-reflex over successive stimulations within the 3-second stimulation trains across all subjects. During the initial several stimulations, the M-wave amplitude fluctuated and then increased to a stable level until the end of the stimulation train. As for the H-reflex activity, the amplitude first decreased
Fig. 5. The EMG segments and their average of the channel with the maximum EMG amplitude from a representative trial (a). The comparison between the M-wave amplitude and the H-reflex amplitude of individual subjects (b). The error bar represents the standard error. *, p<0.05. The amplitude distribution of the M-wave and the H-reflex from a representative trial (c). The box plots of the 2D correlation coefficient of the distribution between the M-wave and the H-reflex from all trials (d).

(though not significantly) and then increased to a relatively stable level until the end of the stimulation train. Therefore, the amplitude of the first 6 stimulations were extracted to perform a repeated measures ANOVA. Results showed that the H-reflex amplitude varied significantly over stimulation (F(5,35)=3.5026, p=0.0114) but no significant difference was observed regarding the M-wave amplitude (F(5,35)=1.1438, p=0.3560). Further post-hoc test with Holm–Bonferroni correction showed that the H-reflex amplitude after the fifth stimulation was significantly larger than that of the third and the forth stimulation (p<0.05). A slow rise of the H-reflex amplitude was evident until 50 stimulations.

3) EMG amplitude distribution

After obtaining the clustering results and the EMG amplitude distribution of individual trials, two different 2D correlation coefficients of the EMG amplitude distribution were calculated between trials. The first was the ‘within-types’ correlation coefficient. Specifically, for a given subject, if a certain motion type was observed in more than one trials, the correlation coefficient was calculated between all possible combinations of the trials. The second was the ‘between-types’ correlation coefficient, which was calculated between trials that belonged to different motion types. Both within-types and between-types correlation coefficients were calculated within individual subjects, because the placement of the EMG array might differ between subjects.

The coefficient values from all subjects were pooled together as shown in Fig. 7a. The median of the within-types coefficients was 0.9104 with the 25th and 75th percentile at 0.8379 and 0.9633, respectively. There were 8 outliers that had relatively small correlation coefficient. As for the between-types coefficients, the median was 0.4428 with the 25th and 75th percentile at 0.1361 and 0.6773, respectively, and no outliers were detected. Fig. 7b illustrates the motion kinematics and the EMG amplitude distribution of two trials with the same motion type. The EMG amplitude distribution shared the similar pattern, which demonstrated that similar muscles or muscle portions were activated in the two trials. On the contrary, even though the two trials shown in Fig. 7c also had the same motion type, the EMG amplitude distributions were significantly different. This meant that the same motion can be elicited by activating different muscles or different portions of muscles.

IV. DISCUSSION

This study sought to investigate the feasibility of transcutaneous electrical stimulation at the proximal segments of the radial nerve in the activation and control of different finger and wrist extension movements. Elicited finger and wrist motions were recorded using a motion capture system, and the activation patterns of finger and wrist extensors innervated by the radial nerve were captured with a HD-EMG grid. The results showed that this non-invasive stimulation technique can
activate the wrist and finger extensors and elicit various finger and wrist joint extension motions both independently and in a coordinated manner. Varying the stimulation location (electrode pairs) can induce different finger and wrist motions. Our findings demonstrate that this stimulation technique can potentially be used as a rehabilitation or assistive approach to elicit different hand opening and/or wrist extension motions for individuals with muscle weakness/spasticity after CNS injuries.

Conventional FES with electrodes placed on the muscle belly largely activates the superficial muscles [14], resulting in limited movements due to the limited accessibility of deep muscles. For example, independent index finger extension requires the activation of the deep muscle extensor indicis proprius, which is rarely activated through conventional FES. In order to access the deep muscles, strong stimulation intensity is required, which typically leads to more diffused recruitment of muscles, further limiting the selectivity of muscle activation. For example, even with a multiple-electrode array and a searching procedure for the optimal stimulation configuration, co-activation of finger and wrist extensors is inevitable [28].

On the contrary, the proximal segment of nerve bundles contains axons innervating both superficial and deep muscles. By adjusting the stimulation intensity and location, different axons innervating different muscles can be activated, resulting in more dexterous and more natural movements. Consistent with the earlier median/ulnar nerve stimulation outcomes [22, 23], our results showed that finger and wrist movements can be elicited both independently and in a coordinated manner by adjusting the stimulation location and intensity, demonstrating the advantage to the conventional FES in the selectivity and accessibility of muscle activation.

Independent index and pinkie finger movements were observed in this study, but no independent middle or ring finger movement was elicited. This was largely determined by the muscle anatomy in the forearm. The contraction of the extensor digiti minimi and the extensor indicis proprius can elicit independent index and pinkie extension, respectively. On the contrary, the extension of ring and middle fingers needs the activation of the extensor digitorum, which generally extends the index, middle, and ring fingers at the same time. Physiologically, the radial nerve at the stimulation location in our study only contains the axons innervating the finger and wrist extensors. However, finger flexion was observed in motion types 5 and 10. It might be caused by the stretch reflex of the finger flexors or passive stretch of the flexors when strong contractions of wrist extensors were induced. In addition, the hand and forearm was placed at the pronated position initially. The effect of segmental gravity might also contribute to the flexion of fingers when no finger extensors were activated. Spasticity is common in the affected arm flexors following a stroke [29], which can arise from a hyper-excitability of the stretch reflex [30] and a reduced agonist-antagonist reciprocal inhibition associated with the development of spasticity [31]. As a result, the electrical activation of extensors can elicit stretch reflex in flexors. Together with a potential contracture in the flexors, it is expected that a larger stimulation amplitude is needed to open the hand or extend the wrist joint. However, since the stimulation intensity needed in our approach is substantially lower compared with conventional FES, it is reasonable to speculate that the discomfort induced by our approach is still much less than conventional FES. On the other hand, since H-reflex was elicited through the nerve bundle stimulation, the flexor spasticity may be reduced with reciprocal inhibition. Clearly, further studies are needed to maximize the H-reflex activity with tuned current amplitude and to evaluate the potentially altered excitability of the reflex in the flexors.

The stimulation of the radial nerve bundle can activate muscles directly through the motor axons (M-wave) and/or through the activation of the afferent fibers (H-reflex). The recruitment order of MUs through H-reflex resembles the physiological conditions [21, 32], and, therefore, the resultant muscle contractions are more sustainable [20]. In our current study, both M-wave and H-reflex activities were observed across subjects, which suggested the potential of our stimulation method in reducing muscle fatigue during elicited movements. It has been shown that the electrically evoked activities in the spinal and supraspinal circuitries through the activation of sensory neurons can lead to short- and long-term plasticity in the neural circuits that control movements [32]. This demonstrated the potential of our stimulation method in enhancing neuromuscular functional recovery by increasing the CNS involvement, which can induce greater overall benefit to the users. However, the amplitude of the H-reflex activity was smaller compared with the M-wave, largely due to the fact that high current intensity was used during the experiment in order to elicit moderate or strong muscle contractions. The high stimulation intensity might activate more motor axons, resulting in more cancellation of the reflex response [33]. Further studies are required to fine tune the stimulation parameters such that H-reflex activities are preferably elicited. A post-activation depression has been observed in H-reflex when stimulating with a high frequency, with a recovery time up to 200 ms for upper extremity muscles [34]. Our results showed that the H-reflex amplitude first decreased mildly for the first 100 ms (3 stimulations), and then increased slowly to a stable level at approximately 1.6 s. This indicated that the H-reflex activity was sustainable with our stimulation method. The slow increase of the H-reflex can come from the increased excitability of the spinal reflex, potentially due to a depolarization of the resting membrane potential with repetitive stimulation. The high-frequency stimulation can also trigger voltage dependent persistent inward current, which can amplify the H-reflex amplitude [35, 36].

The EMG amplitude distribution revealed the activation pattern of muscles in a large area. The 2D correlation of the EMG amplitude distribution was used to quantify the similarities of muscles activated across different stimulation conditions. Our results showed that the correlation of the EMG distribution from two trials was generally high if they corresponded to the same motion type. However, it was interesting that the similarity between some trials with the same motion type was substantially small compared with the average value (identified as outliers in Fig. 7a). This demonstrated that
different muscles or muscle portions controlling the same motion can be activated separately by changing the stimulation locations or intensity. The inherent redundancy in the stimulation can be helpful for further development of this stimulation technique, in order to reduce muscle fatigue by alternating the activation of different muscles or muscle portions. As for the correlation coefficient of the EMG amplitude distribution between trials with different motion types, the values had a larger range with a median below 0.5. Some trials with different motion types had highly-correlated EMG distributions. One possible reason was that the surface EMG signals can only capture the activity of superficial muscles rather than deep muscles that resulted in different motions.

The average current amplitude used in this study was 6.32±1.51 mA across all subjects, which were substantially lower than the typical values in the conventional FES with electrodes placed on the muscle belly that range from 10-100 mA [28, 37]. The high current intensity is needed because the distal portion of the nerve fibers branches out further into different portions of the muscle. Therefore, a high current intensity is needed to activate sufficient nerve branches, in order to produce meaningful force output. On the contrary, all the axons of the radial nerve are gathered in a bundle with a small cross-sectional area of 3 mm² and with a depth of 1.21 cm [38] at the radial groove. Therefore, a low current intensity is sufficient to activate motor axons that innervate a large number of muscle fibers to produce comparable forces. Compared with our previous study in which the median/ulnar nerve bundles were targeted [26], the average current intensity in this study was higher (6.32 mA vs. 3.7 mA), largely due to the fact that the median/ulnar nerves are more superficial compared with the radial nerve at the stimulation location. The low current intensity can reduce the discomfort or pain sensation induced by the electrical stimulation, which can help improve the adoption of the electrical stimulation systems. In addition, the low current can also potentially benefit future development of a portable stimulation system when power consumption is taken into consideration.

One limitation of the current study was that the movement of the thumb was not recorded, mostly due to the complex degrees of freedom of the thumb. Forearm supination was not recorded either mainly because hand opening motions were targeted in this study. However, both thumb motions and forearm supination motions were observed during the experiment. Since the thumb is also important to most hand functional movements in activities of daily living and limited supination is common in affected arms [39], further studies should take the thumb and the forearm supination into consideration and analyze the motion types that can be elicited through proximal nerve stimulation. In addition, different electrode pairs were manually searched, and the selection of the stimulation intensity was, to some degree, arbitrary. Automatic searching strategies can be developed further to find the optimal stimulation configurations, in order to elicit all functional movements of different joints both independently and in a coordinated manner. Further, the long-term stability of the elicited motion and the day-to-day stability after donning-doffing of the stimulation grid also need to be investigated in future studies before clinical applications. It is expected that the elicited movement patterns for a given electrode pair will vary between days after donning-doffing. An automated recalibration process [23] to identify the available motions can facilitate the utility of the nerve stimulation technique across days.

V. CONCLUSIONS

This study demonstrates the feasibility of eliciting dexterous finger and wrist extension motions via non-invasive electrical stimulation at the proximal segment of the radial nerve bundle. This stimulation method shows promise in generating individual and coordinated joint movements of the finger and wrist. The accessibility to both superficial and deep muscles with low stimulation intensity can lead to more natural movements with less discomfort. The motions elicited through the central pathway can be more fatigue-resistant and can also potentially facilitate neuroplasticity of the CNS circuits. These advantages can potentially alleviate some of the difficulties faced by the conventional FES targeting the muscle belly, and can enhance the functional recovery of individuals with neurological injuries. With further development, this non-invasive stimulation technique has the potential to be a more desirable rehabilitative or assistive approach for individuals with weakened or spastic hand/wrist muscles after CNS injuries.

REFERENCES


