Abstract—The inability to effectively activate and control skeletal muscles is a common impairment following a variety of neurological conditions or injuries. One common approach to restoring or augmenting this impairment is the use of external electrical stimulation of the muscles, called functional electrical stimulation (FES). Typically targeted directly at the anatomical muscle belly, existing methodologies often involve high current amplitudes, limited superficial muscle activation, and early onset of muscle fatigue. We have recently explored the capabilities of a non-invasive peripheral nerve stimulation method for the dexterous control of finger and hand muscles. Further development of our stimulation system has enabled us to manually search across a variety of stimulation locations with increased consistency and efficiency. This study examined the preliminary results in two subjects of an automated stimulation system which can rapidly characterize a large combination of stimulation electrodes. Our preliminary findings suggested that the stimulation grid was able to produce a number of clustered EMG activities and finger forces. This robust ability to flexibly generate different grasp patterns demonstrates the promise of the methodology in future applications for FES and rehabilitation.

I. INTRODUCTION

Weakness of the hand is a greatly debilitating impairment which impedes an individual’s self-sufficiency and ability to complete many activities of daily living[1]–[3]. Typically related to an inability to voluntarily activate their own muscles, various methods are utilized in clinical and research settings to attempt to rehabilitate this lost function. One of the major methods used for motor rehabilitation is functional electrical stimulation (FES) of the hand muscles[4]–[7]. FES enables activation of muscles by electrically stimulating the nerve before it enters the muscle and can be used to augment force generation. FES in the clinic is often restricted to large diameter electrode pads which activate a broad range of superficial muscles, requires higher stimulation currents for functional force levels, and leads to non-specific activation of all the fingers[8]–[10]. FES has also been shown to result in rapid onset of muscle fatigue due to the non-physiological synchronized recruitment of motor units, and therefore is unable to be used continuously for an extended period[11], [12].

Current research in FES of the hand seeks to reduce these issues of non-specificity and easy fatigue. One such example of this is by Malsevic et al, who have developed a multi-pad electrical stimulation system targeting the hand muscles[13].

This system is able to improve the effectiveness of conventional FES by utilizing multiple small electrode pads placed on the skin over the hand muscles. Stimulation could be automatically generated at different electrodes while simultaneously obtaining data on the movement of the fingers induced by the stimulation. This automated design could then identify spatial regions of muscle activation which corresponded to specific fingers. Despite these advances, the major issues of conventional FES involving limited superficial muscle activation and higher current amplitudes still have not been resolved. Alternatively, other studies have utilized implantable stimulation electrodes to peripheral nerves which are able to selectively activate precise finger movements at low current amplitudes [14], [15]. This enables high dexterity of finger movements while also allowing deeper muscle activation. However, the invasive nature of the procedures limits the widespread accessibility of this FES method.

Recently, we have explored the capabilities of a non-invasive peripheral nerve stimulation approach in eliciting a variety of individual and multi-finger movements[16]. By stimulating the ulnar and median nerves where they were superficially accessible in the upper arm, we demonstrated a broad spread of hand grasp patterns which could be generated through the manual movement of a standard bar electrode around this area. Differences in the spatial placement of the stimulation electrode likely induced varied electrical fields around the two nerves and selectively activated certain efferent fibers resulting in isolated finger motions. However, the major limitation of the prior study was the inefficiency in searching for electrode locations that were able to elicit different movements of the hand. To improve the consistency, stability, and efficiency of this stimulation approach, we developed an electrode grid and switch system which could programatically route the stimulation output channels to different stimulation locations. This approach combined with custom-made software enabled us to quickly switch between active electrode pairs while removing the need to physically place and replace the stimulation electrode. Several currently unpublished studies have used this system to aid in the initial manual searching of stimulation locations prior to other experimentation.

The purpose of this study was to explore the potential of an automated searching method through the many available combinations of stimulation electrode pairs. This would enable the rapid identification of the available finger grasp patterns as well as the overall characterization of the stimulation grid’s specificity and redundancy. To better quantify the functional output of each stimulation, both individual finger flexion forces as well as high-density electromyography (HDEMG) was obtained alongside each commanded stimulation electrode pair.

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II. METHODS

A. Subjects

Two healthy control subjects (Male, Ages 30 and 34) with no known neurological or neuromuscular conditions were recruited for this study. All participants gave informed consent via protocols approved by the Institutional Review Board at University of North Carolina at Chapel Hill.

B. Experimental Setup

Subjects were seated at a table with their forearm resting in front of them at a neutral position (90° pronation). Standard EMG preparation of the anterior forearm and palm was used to reduce the skin impedance. A 4x8 HDEMG array (ELSCH064NM3, OT Bioelettronica) was placed on the hand and a 16x8 HDEMG array was placed on the anterior forearm over the finger flexors. Each finger (excluding the thumb) was individually secured to a force transducer (SM-200N, Interface Inc.) via a 3D-printed attachment interface (Figure 1, Top). Each force transducer was adjustably attached to a vertical base so that each finger could be held at a natural abduction angle. Once each finger was securely immobilized above the MCP joints, large foam blocks were placed around the palm and wrist to restrict any translation of wrist movement to the finger transducers. Following the force and HDEMG setup, the medial side of the upper arm was cleaned using alcohol wipes in preparation for the stimulation electrode array. The subjects were asked to “flex” their bicep, and the experimenter palpated the concave space below the muscle bulge for the brachial artery. Sixteen custom-cut round button electrodes (~1 cm diameter, Kendall H59P Cloth Electrodes, Covidien Inc.) were placed alongside the biceps to form a 2x8 grid running parallel to the median and ulnar nerve bundles (Figure 1, Bottom).

The 32+128 EMG signals were band-pass filtered at 10-900 Hz, amplified with a gain of 500, and sampled at 5120 Hz (EMG-USB2+, OT Bioelettronica). Each stimulation electrode was connected to a switch matrix (34904A, Agilent Technologies), the output of which was connected to a multi-channel programmable stimulator (STG40008, Multichannel Systems). The force signal was acquired at 2000 Hz using a NI USB-6225.

C. Procedures

A custom MATLAB (2016b, MathWorks Inc.) graphical user interface (GUI) was utilized to combine the control of the switch matrix, stimulation generation, and force acquisition. The HDEMG was displayed and recorded using OT BioLab (v3.2, OT Bioelettronica). The stimulation output was designed as a 40 pulse per second (pps) train of 200us biphasic square pulses over a half second duration (20 pulses per train). An additional half second of rest was given following each stimulation while the system automatically switched to the next randomized electrode pair. The start of each stimulation train also output a 3V synchronization pulse which triggered the acquisition of a 1 second duration of force, and was recorded alongside the HDEMG as an additional auxiliary input. The electrode pair, stimulation parameters, and 4 finger forces for each trial were saved together in MATLAB, and the HDEMG were recorded continuously during each searching session.

A preset 4 mA of current was used as the standard amplitude for all stimulations. An initial sequential stimulation of all the electrode pairs was conducted to identify if any of the stimulation channels resulted in noxious sensations at the local stimulation site. Any painful electrode channels were excluded and the remaining combinations of potential stimulation electrode pairs (16 choose 2 = 120) were randomly ordered and repeated 3 times each. Only a single searching session at 4 mA was run for each subject for this preliminary study.

D. Data Analysis

The compound muscle action potential (CMAP) of each trial set and electrode pair repetitions (20 pulses x 3 repetitions) was averaged to obtain the mean CMAP. The Area Under the Curve (AUC) of each absolute CMAP was calculated as an estimate of the EMG activity for each channel, and subsequently, these AUC values for each channel on the arm...
electrode pad were arranged into a 16x8 matrix to match the spatial orientation of the EMG array. This 2D AUC Map represents the spatial EMG activation pattern of the underlying arm muscles corresponding to a single stimulation pair (Figure 2). The 2D correlation coefficient between each AUC Map was calculated to form a correlation matrix between all of the electrode pairs. Hierarchical clustering was used to identify natural clusters of EMG activity, and the stimulation pairs corresponding to these clusters were grouped together. An averaged peak finger force corresponding to each stimulation pair was also calculated and grouped according to the EMG clusters. The hand EMG data were excluded as there was insufficient activity between different electrode pairs to result in well-separated clusters.

III. RESULTS

The EMG clustering results and the grouped finger forces from Subject 1 are shown in top half of Figure 3. In subject 1, seven EMG clusters were found which were activated by at least 2 electrode pairs. Upon visual inspection, each of the EMG clusters did qualitatively show distinct EMG patterns across the recorded muscle. The corresponding peak finger forces were also grouped together, the mean and standard error of each shown in the bottom of Figure 3. All the cluster groups showed activation of the middle finger, while having different levels of activations for the other fingers for each stimulation cluster.

Figure 4 shows the 6 stimulation clusters obtained from Subject 2. Like the first subject, the EMG clusters for subject 2 also showed visually distinct regions of EMG activity, albeit more concentrated in the distal-ulnar side of the EMG pad. The 4-finger peak force averages for each cluster also showed different groupings of force activity. However, the relative differences of the force levels between each finger appeared to be less distinct.

IV. DISCUSSION

This preliminary study characterized the different patterns of muscle activity and force output elicited from our novel non-invasive stimulation method. A stimulation electrode grid placed near the short head of the biceps muscle was used to selectively activate different portions of the ulnar or median nerve which resulted in variable EMG activity as well as individual finger forces. The CMAP AUC was used to estimate the spatial EMG activity across the HDMEG array, and each of these AUC Maps were clustered to determine which electrode pairs activated similar EMG patterns. Our
results on two control subjects show that our stimulation method was able to produce at least 6 distinct EMG clusters which also corresponded to differing levels of individual finger forces. These findings provide preliminary insight into the range and specificity of our stimulation electrode grid in eliciting unique repeatable muscle activations and grasp patterns.

As shown in our previous study, multiple unique muscle activation patterns are transcutaneously elicitable through variations in the induced electric field along the ulnar and median nerves. Although, our current stimulation grid was not able to reproduce the exact level of specificity of the previous study’s manual searching of stimulation locations, the overall ease of setup and reliability of the current methodology is a positive tradeoff for moving towards a more universal and widely applicable stimulation approach. Moving forward, it is important to determine how the different stimulation electrode pairs respond to varied current amplitudes as well as to different combinations of simultaneous stimulation.

One of the limitations of the current study was that a fixed current intensity was used for all the stimulation locations. This can limit the activation patterns observed, as different electrode pairs are likely to have differing sensitivities to varied current amplitudes. A broader characterization of multiple current levels would be able to better predict which stimulation channels best matched a desired movement. Future directions for the stimulation grid should involve automated searching of the electrode pairs across several current levels, as a way to quantify which pairs produce the best range and resolution of muscle activation. Additionally, further exploration of the activation of finger extensors must be investigated as both flexion and extension are necessary for a complete neurorehabilitative application.

REFERENCES