Influence of Muscle Geometry and Stimulation Site on Neuromuscular Electrical Stimulation of the Biceps Brachii

Eric J. Gonzalez, Ryan J. Downey, Courtney A. Rouse, and Warren E. Dixon

***Abstract*—Functional electrical stimulation (FES) can help individuals with physical disabilities by evoking or assisting limb movement; however, the change in muscle geometry associated with limb movement may affect the response to stimulation. The aim of this study was to quantify the effects of muscle geometry and stimulation site on muscle torque production. Contraction torque about the elbow was measured in 12 healthy individuals using a custom elbow flexion testbed and a transcutaneous electrode array. Stimulation was delivered to 6 distinct sites along the biceps brachii over 11 elbow flexion angles. Flexion angle was found to significantly influence the optimal (i.e., torque-maximizing) stimulation site (P-value < 2×10^{-16}). Similarly, the biceps stimulation site was found to significantly influence the flexion angle at which peak torque occurred (P-value = 2.9×10^{-15}). This study demonstrated that the change in muscle geometry associated with functional limb movement impacts the muscle's response to stimulation. Since maximizing muscle force produced per unit stimulation is a common goal in rehabilitative FES, future efforts could examine methods which compensate for the shift in optimal stimulation site during FES-induced limb movement (e.g., flexion angle-based electrode switching within an electrode array).**

***Index Terms*—Electrode Array, Functional Electrical Stimulation (FES), Neuromuscular Electrical Stimulation (NMES)**

I. Introduction

Neuromuscular electrical stimulation (NMES) is the application of an electrical stimulus to activate motor neurons thereby eliciting a muscle contraction. Functional electrical stimulation (FES) is the use of NMES to specifically yield functional limb motion (e.g., grasping, walking, reaching, and cycling). Both NMES and FES are often used in rehabilitative settings to increase muscle strength and function [1]–[5]. Furthermore, a common application of FES is the restoration of limb function in individuals post stroke, spinal cord injury or other neurological disorders, with the aim of improving daily function via external stimulation [6]–[10]. Traditionally, the implementation of FES involves the use of 2 surface electrodes per muscle group to deliver current and produce a desired response. It is known, however, that muscle length varies with limb flexion/extension [11], [12], and electrode position relative to the underlying muscle can impact contraction strength [13]. Since functional limb flexion and extension through a wide range of motion is desired, understanding the influence of changing muscle geometry on muscle response to stimulation may lead to improved methods of delivering FES.

One drawback of traditional FES (i.e., implemented with 2 typically large surface electrodes) is the overflow of stimulation to nearby muscles that do not contribute to the functional goal, resulting in unnecessary discomfort [14] and imprecise motor control in patients [15], [16]. Hence, proper electrode size and placement relative to the underlying skeletal muscle should be considered. In particular, the proximity of stimulation to muscle motor points has been shown to increase muscle force output per unit stimulation intensity while reducing patient discomfort [17]–[19], thereby yielding more efficient and comfortable limb movement. Here, the term motor point is used to describe the skin area superficial to the muscle where the motor activation threshold is minimized for a given electrical input [18]. Furthermore, it is known that increased muscle force output during NMES/FES can improve muscle strengthening [20]. Since improved muscle strength and patient comfort are common goals in rehabilitative settings, targeting optimal stimulation sites such as motor points during FES may lead to improved rehabilitative treatments.

Muscle geometry (i.e., muscle shape, length, and size) may vary considerably during dynamic, functional limb motion (e.g., reaching or lifting). Consequently, the position of a static electrode placed on the skin may shift relative to the underlying muscle as limb motion occurs. Therefore, the aforementioned benefits of targeting precise points during FES may be enhanced by utilizing stimulation methods which increase selectivity and allow for flexibility in stimulation site.

In recent years, the use of multi-channel electrode arrays has emerged as a method to increase selectivity [21]–[23] and lower the rate of fatigue [24]–[26] during FES. Typically a one- or two-dimensional distribution of small surface electrodes, multi-channel electrode arrays are commonly used in applications requiring fine motor movement, such as grasp restoration [22], [27]–[32]. While much of the work involving multi-channel arrays has been successful for muscle contractions in stationary limbs [21], [22], [27]–[33], there

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has been little investigation into the use of an electrode array to maximize muscle contraction throughout the full range of motion of a limb. This is particularly significant because muscle activation using an electrode array has been suggested to depend on limb orientation and position, due to the relative shift of underlying muscle tissue beneath the skin [21].

The shape/length of the human biceps has a clear dependence on upper limb orientation (i.e., elbow flexion angle), making the biceps brachii a good candidate for the study of stimulation-induced muscle response as a function of changing muscle geometry. While some studies have explored the control of planar arm motion using closed-loop control of FES [34]–[36], the relationship between force production in the biceps, stimulation site, and upper limb orientation remains unclear. In theory, limb position information could be used to adjust the stimulation site in real-time via electrode switching, potentially leading to more efficient muscle recruitment, less fatigue, and reduced patient discomfort during FES.

The aim of this study was to examine the relationship between human arm orientation (i.e., elbow flexion angle), stimulation site, and stimulation-induced torque production in the biceps brachii. The optimal (i.e., torque-maximizing) electrode in a 1 x 6 array was determined at 11 flexion angles throughout the complete range of motion of the arm in an effort to quantify the effect of changing muscle geometry on torque production about the elbow. Flexion angle was found to significantly influence the optimal stimulation site, and similarly, stimulation site was found to significantly influence the flexion angle at which peak torque occurred. These findings indicate an opportunity for future efforts to examine methods which compensate for the shift in optimal stimulation site during FES-induced limb movement.

II. Methods

A. Subjects

Twelve able-bodied individuals (9 male, 3 female, ages 21–44 years) participated in the study. Written consent was obtained from all individuals prior to participation, as approved by the institutional review board at the University of Florida. Participants had no history of joint issues in the upper body and did not report any significant soreness in the biceps of either arm prior to participating in the study. Because the left and right arms of any participant may vary in strength, muscle mass, and response to stimulation, the left and right arms of each participant were considered to be distinct; thus, a total of 24 unique arms were tested in this study.

B. Apparatus and Materials

All testing was performed using the custom arm flexion testbed depicted in Figure 1. The testbed consists of 1) a 12 VDC gear motor (Allied Motion PLA25) which governs flexion of the participant's arm, 2) an optical encoder (US Digital HB6M) which measures the participant's arm flexion angle, 3) a torque transducer (FUTEK TFF350) which measures the net torque about the participant's elbow axis, 4) a hinged aluminum frame to which the participant's arm is secured, and 5) a personal computer which controls flexion angle via the gear motor and collects sensor information via a data acquisition device (Quanser Q8-USB) and a compiled Simulink diagram. The arm of each participant was fit with a custom surface electrode array consisting of six self-adhesive 0.6″ x 2.75″ PALS® Flex-Tone electrodes (cut from an original size of 0.6″ x 6″) placed over the biceps and one 3″ x 5″ Valutrode® electrode placed over the triceps, used as the reference for each biceps electrode.[1] Stimulation to each of the six electrode subgroups (consisting of an individual biceps electrode and the shared reference electrode on the triceps) was delivered via a current-controlled 8-channel stimulator (RehaStim, Hasomed GmbH, Germany), which was controlled by a personal computer via ScienceMode (compiled as a Simulink block).

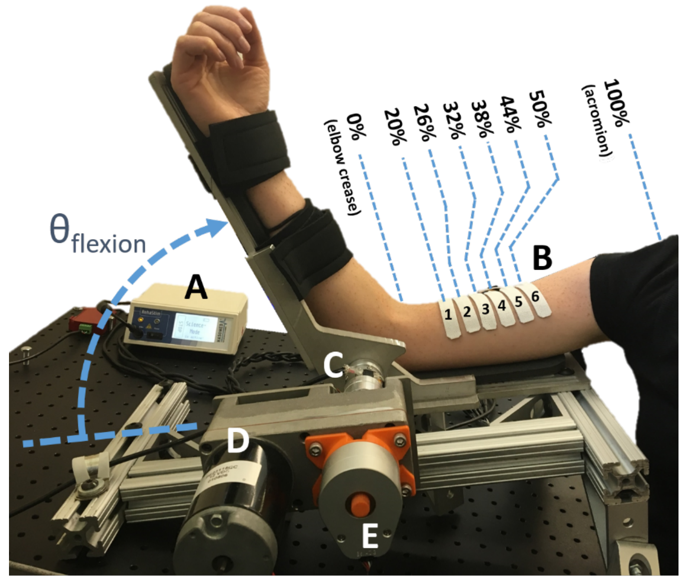

Fig. 1. The experimental setup consists of A) a current-controlled stimulator, B) an electrode array placed on the muscle belly of the participant's biceps, C) a torque transducer, D) a 12 VDC gear motor, and E) an optical encoder. Electrodes 1 through 6 and their relative locations between the elbow crease and acromion are labeled on the array above. The flexion angle of the participant's arm is defined by θ_{flexion}. Mechanical safety limits prevent hyperflexion/hyperextension of the arm.

C. Electrode Arrangement

Each electrode subgroup mimics the electrode arrangement commonly used during muscle motor point identification, in which a small active electrode is used to elicit contraction of the target muscle while a large reference electrode is placed over the antagonist muscle (i.e., monopolar stimulation) [18], [37]. This electrode arrangement causes the current density of the active electrode to be greater than that of the reference electrode (as current density is inversely related to electrode area). Therefore, as the stimulation intensity is increased, motor units in proximity to the active electrode will be preferentially recruited compared to those in proximity to the reference electrode. In the present study, the reference

[1] Surface electrodes for the study were provided compliments of Axelgaard Manufacturing Co., Ltd.

electrode was selected sufficiently large such that a tricep contraction did not occur at the stimulation intensity tested.

Placement of the active electrode array on the biceps wa standardized across participants based on the relative positic of each electrode with respect to the length of the participant upper arm. As illustrated in Figure 1, for each participa Electrode 1 was placed at 20% of the distance from the elbo crease to the acromion and Electrode 6 was placed at 50% with the remaining electrodes spaced evenly between (i.e., relative distance of 6% between electrodes). These relative po sitions were determined through preliminary experimentatic and approximately cover the entire muscle belly of the bicep throughout the entire range of motion of the arm.

D. Warm-up Protocol

All participants completed a brief warm-up protocol prior to the experimental protocol to get acclimated with the sensation of NMES. During this period, each participant received 1 second of stimulation out of every 15 seconds, delivered to a single active electrode in the array. Note that the reference electrode on the triceps also received stimulation since the stimulator delivered symmetric biphasic pulses. However, the large surface area of the reference electrode ensured that current density remained below the excitation threshold and thus triceps contraction was not expected to occur. Participants indicated that they could not sense a triceps contraction, which was also confirmed by visual inspection.

Stimulation was delivered at a frequency of 30 Hz with a current amplitude of 25 mA. Electrical stimuli were delivered at 30 Hz since literature suggests this frequency to be a reasonable compromise between slowing fatigue and eliciting strong contractions. Current amplitude and pulse duration were selected based on preliminary experiments. Specifically, the goal was to use stimulation parameters that would elicit measurable torque in all person-electrode-angle combinations while not inducing contractions so strong that they would cause discomfort.

Prior to data collection, pulse duration was manually incremented from 20 μs to 100 μs over the course of approximately 90 seconds to familiarize the individual with the stimulation sensation. A minimum rest period of 3 minutes was required following the pretrial period prior to beginning the experimental protocol.

E. Experimental Protocol

During the experimental protocol, computer control of the motor was used to cycle the participant's arm through a randomized sequence of 11 target flexion angles (0, 10, 20, . . . , 100 degrees) within ± 2 degrees. At each flexion angle, each of the six biceps electrodes received 1 second of stimulation (30 Hz, 25 mA, 100 μs) in a random order as elbow torque was recorded. The motor was locked such that stimulation-induced contractions of the biceps did not alter the flexion angle of the arm (i.e., contractions were isometric). The order of flexion angles and stimulation sites (within each flexion angle) tested were randomized to prevent systematic bias due to possible carryover effect of fatigue. An example of the stimulation pattern and induced torque measured at one fixed angle is presented in Figure 2.

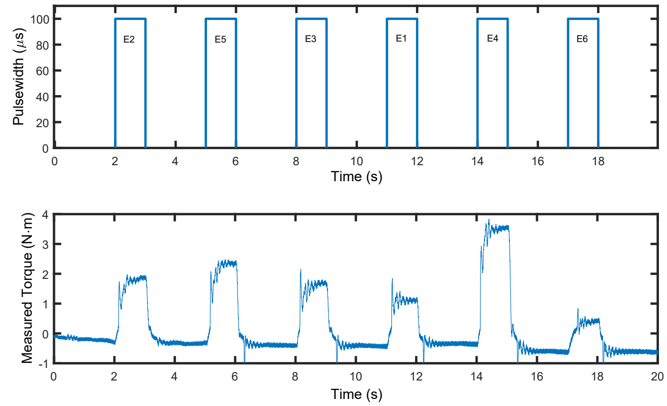

Fig. 2. Example stimulation pattern sent to the electrode array (top) and the resulting torque caused by biceps contraction (bottom) obtained from a single participant's arm fixed at a flexion angle of 90 degrees. E1 - E6 refer to Electrode 1 - Electrode 6, respectively. Note that the low frequency drift in baseline torque is accounted for when calculating mean contraction torque.

F. Data Analysis

For each flexion angle and electrode combination, the mean contraction torque was calculated. Mean contraction torque (subsequently referred to simply as contraction torque) was defined as the difference between the mean torque measured during contraction (averaged from data taken in the central 0.5 seconds of the 1 second of stimulation) and the mean baseline torque measured in the 0.5 seconds prior to stimulation, during which the arm was at rest. Defining contraction torque in this way compensated for the effects of gravity and sensor drift on the measured torque.

For each flexion angle in the trial an optimal electrode was determined, defined as the electrode within the array which produced the maximum contraction torque at that flexion angle. Similarly, for each electrode within the array, the flexion angle that yielded the maximum contraction torque was also calculated (henceforth termed the peak-torque flexion angle). Repeated measures ANOVA was used to examine the influence of flexion angle on the optimal stimulation site and similarly to examine the influence of stimulation site on the peak-torque flexion angle. Statistical analysis was performed in R.

III. Results

Contraction torque about the elbow was recorded at 11 flexion angles between 0 and 100 degrees for 6 stimulation sites on the biceps, corresponding to the 6 electrodes within the biceps electrode array (previously depicted in Figure 1). The mean normalized contraction torque for each electrode within the array (averaged across all 24 tested arms) is plotted as a function of flexion angle in Figure 3. Electrode 3 and Electrode 4 (positioned at 32% and 38% of the distance between the elbow crease and acromion, respectively) produced the largest contraction torques throughout the entire range of motion with reasonable consistency; in particular it was observed that, on

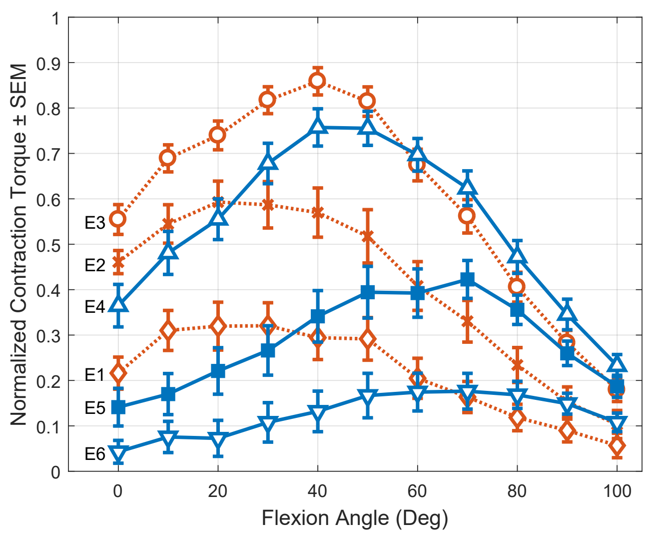

Fig. 3. Contraction torque produced by stimulation at each electrode as a function of flexion angle. To account for intersubject variability in stimulation response (i.e., contraction strength), the torque was normalized for each arm by the maximum torque elicited over the entire trial (i.e., over all 66 electrode/flexion angle combinations in the single trial). Depicted is the mean normalized contraction torque (across all 24 arms tested) ± the standard of the mean (SEM) for each data point. E1 - E6 refer to Electrode 1 - Electrode 6, respectively.

average, Electrode 3 was optimal for flexion angles below 60 degrees while Electrode 4 was optimal for flexion angles of 60 degrees and above. Repeated measures ANOVA indicated that flexion angle significantly affects the optimal (i.e., torque maximizing) stimulation site ($F_{10,230}$ = 13.42, P-value < 2×10^{-16}). Similarly, electrode position (i.e., stimulation site) was a significant predictor of the flexion angle at which peak torque occurred, i.e., the peak-torque flexion angle ($F_{5,115}$ = 21.72, P-value = 2.9×10^{-15}).

IV. Discussion

The results of this study indicate that 1) flexion angle significantly influences the location of the optimal electrode within an electrode array and 2) stimulation site significantly influences the flexion angle at which contraction torque is maximal (i.e., peak-torque flexion angle). From Figure 3 it is clear that, on average, Electrode 3 (located at 32% of the distance between the elbow crease and acromion) was optimal for flexion angles below 60 degrees and Electrode 4 (38%) was optimal for flexion angles 60 degrees and above. Stimulating at these optimal locations results in the maximizing of muscle torque output per unit stimulation. Since increasing contraction intensity has been shown to improve muscle strengthening [20] and reducing stimulation is known to improve patient comfort [17], it may be beneficial to utilize an electrode array to switch between optimal stimulation sites as a function of muscle geometry (i.e., flexion angle) during NMES/FES. It should be noted, however, that while the optimal electrode was shown to depend on flexion angle, the results of this study indicate that variation in the mean optimal stimulation site was limited between two of the six electrodes within the current array (i.e., Electrodes 3 and 4, occupying a region roughly 6% of the total elbow crease to acromion upper-arm length). Thus, future efforts may focus on using an array of smaller electrodes more finely distributed in this region to improve selectivity and better target the optimal stimulation site as it shifts with the underlying muscle geometry.

The size of this optimal region – typically about 2 cm based on the range of elbow crease-acromion distances tested (28.5-36.8 cm) – is comparable to the results of Crochetiere et al. which found that the triceps brachii motor point linearly shifts approximately 2 cm during arm flexion [13]. This indicates that a fine electrode array may be also a useful tool in the tracking of optimal stimulation sites of other muscles which exhibit changes in geometry during FES-induced limb flexion/extension. Furthermore, the observed shift of the biceps optimal stimulation site mirrors that of the biceps innervation zone – the region of the biceps brachii corresponding to a high concentration of neuromuscular junctions – during arm flexion; one previous study reported a 1.5-2 cm shift in location of the innervation zone on the biceps brachii over an 80 degree flexion range [38]. While the locations of the innervation zone and motor point on the biceps have been shown to differ by up to 1.1 cm [39], their close proximity and similar response to limb flexion indicate the two landmarks are closely related, although the specifics of this relationship remain unclear. The innervation zone has also been shown to shift proportional to the intensity of voluntary muscle contraction [40]. Thus, it is reasonable to suggest that stimulation intensity during NMES may influence the location of the motor point (and thus, the optimal stimulation site); in particular, higher intensity stimulation may cause the muscle to shorten more, amplifying the change in muscle geometry induced by limb flexion. Future efforts may include analyzing the effect of increased stimulation intensity on the tracking of optimal stimulation sites during arm flexion.

Additionally, while it is known that the elbow torque-joint angle relationship exhibits a peak [41], previous studies have not examined the effect of electrode location on the magnitude or location of this peak during NMES. The results of the present study show that electrode placement significantly influences the flexion angle at which peak torque occurs. This further emphasizes the impacts of electrode placement (i.e., stimulation site) during NMES/FES, particularly with regard to functional performance.

V. Conclusion

The results of the present work indicate opportunities for improving functional performance during upper-arm FES applications that induce significant changes in muscle geometry (e.g., lifting or arm cycling). In particular, flexion angle of the elbow was found to influence the optimal stimulation site within an electrode array, and stimulation site was found to influence the flexion-angle which maximizes contraction intensity for a given stimulation level. Based on these findings, future efforts could combine closed-loop control of FES with flexion angle-based electrode switching to maximize contraction intensity throughout a desired trajectory while minimizing commanded stimulation. This may yield improved tracking and reduced fatigue compared to traditional NMES/FES.

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