THE FEASIBILITY OF AN UPPER EXTREMITY POSTSTROKE NEUROPROSTHESIS

by

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Towards Determining The Feasibility Of An Upper Extremity Poststroke Neuroprosthesis

Abstract

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Hemiparesis after stroke makes bimanual tasks difficult. Functional Electrical Stimulation (FES) for reach and hand opening, coupled with residual voluntary movement, may be able to provide functional arm and hand movement. This type of approach has been attempted in the past, focused on the hand, but effort to reach and open the hand produces involuntary flexor co-activation that hand opening stimulation cannot overpower. Limiting effort and augmenting it with stimulation may limit the expression of co-activation patterns and produce useful movements. Low levels of effort that limit the expression of co-activation patterns may provide an effective command signal as well. The end goal is an FES system that uses limited effort as a command signal and produces movement through a combination of residual movement and FES.

We tested key aspects of the system by evaluating 1) the feasibility of using stimulation to produce functional movement in the presence of voluntary effort and 2) how well stroke patients can control assistive forces analogous to those produced by FES. To assess these aspects of this approach we evaluated the following aims: 1) Determine if FES in the presence of limited effort produces useful reach and hand opening. 2) Evaluate the interaction of FES with voluntary
effort in stroke to understand if stimulation of a volitionally activated muscle produces additional force/movement and how those forces combine. 3) Evaluate poststroke control of assistive forces to assist in reach.

Results indicate that FES coupled with limited effort can produce useful reach and hand opening in some stroke patients. The force increment provided by this stimulation decreases as the voluntary effort exerted increases, but it still adds to the assistive force. Additionally, stroke patients can control assistive forces from a robot using residual arm movements on their affected side, indicating that EMG from the affected upper extremity could be used as an effective command signal to control FES and might be used in concert with FES to produce useful arm and hand movements. The positive results from these studies are a step towards an assistive technology to help people move their arm and hand after stroke.
Chapter 1 – Introduction

This chapter outlines the background for this project, including the motivation, resulting approach, and the specific aims of this project. The chapter includes an overview of the impact of poststroke impairment, a possible mechanistic cause of impairment, and current therapeutic approaches. Next, the proposed approach for using an FES system during everyday tasks is outlined. Lastly, it includes the resulting aims and experiments to assess the feasibility of such an approach.

Stroke Impairment

Stroke is a leading cause of disability in the United States, with 6.8 million people currently suffering from stroke, and about 795,000 additional strokes occurring every year [1]. In 2003 it was reported that half of chronic ischemic stroke survivors over the age of 64 have a degree of upper limb hemiparesis [2], which limits arm and hand function on one side of the body. These impairments make bimanual tasks difficult, if not impossible. There is some recovery of motor function after stroke, but it varies widely between patients. Stroke victims’ motor recovery traditionally occurs within the first six months after stroke, with functional gains plateauing thereafter [3, 4] leaving half of the stroke victims with arm and hand impairments after their initial recovery.

Typical stroke impairments are characterized by weakness, an increased sense of effort even for small movements [5], slow and segmented movements [6-8], difficulty making isolated movements [9-12], and spasticity [13, 14]. These
movements are typically characterized as synergy patterns which result in undesired involuntary muscle activation [9-12, 15] during voluntary movement. The upper extremity synergy patterns are clinically characterized in two forms. First, the flexion synergy pattern is characterized by co-activation of shoulder abductors, elbow flexors, wrist flexors, and finger flexors. For example, an attempt to abduct the shoulder causes involuntary elbow flexor and finger flexor co-activation, thereby limiting reach and hand opening. Second, the extension synergy pattern is characterized by co-activation of shoulder adductors, elbow extensors, wrist extensors, and finger extensors. In addition to co-activating muscles across different joints, the synergy patterns produce excessive co-contraction. This co-activation can completely prevent some movements, like hand opening [16]. Co-activation patterns are seen to some extent in able-bodied individuals [11, 17], but the magnitude and amount of flexibility change considerably when evaluating the stroke population.

**Potential Mechanism of Impairment**

The causes of the different impairments may be linked. A change in the descending drive from upper motor neurons to lower motor neurons is a current theory. Specifically, it is thought that damage in the motor cortex or connected pathways limits use of the corticospinal tract for transmission of commands to lower motor neurons. Inability to transmit motor commands then results in increased cortical overlap due to increased activation of pre-motor and supplementary motor areas [18] and brainstem pathways to compensate for the
lack of muscle activation provided by the primary motor cortex [19]. A potential explanation is that in response to decreased activity from the primary motor cortex, the cerebellum attempts to account for inconsistencies between the expected model of the motor system and the sensory response. As a result, the motor system relies heavily on brainstem pathways as a primary means of motor control. Brainstem pathways tend to activate groups of muscles, which would explain the resultant synergy patterns.

The synergy patterns have been studied isometrically [11] and during movement [9, 12]. Isometric studies have investigated the contributions of several joints, but kinematic studies have tended to group only two joint movements at a time. For example, the interactions between shoulder abduction and elbow extension [9, 20] and shoulder abduction and finger flexion [12] have been studied. Evaluating two joints at a time provides valuable insight into the synergy patterns, but it would also be useful to determine the extent of involuntary responses at each joint in response to both individual and multi-joint movements. A significant finding of the kinematic studies is proportional co-activation in other muscles in response to voluntary effort. Despite these findings, the co-activation patterns and stroke patients ability to independently control the activation are still not fully understood.

Understanding how long after a stroke the synergy patterns express themselves provides insight into if the resultant motor changes are produced by motor relearning after the initial insult or are an inherent response to the brain
damage. Preliminary research indicates that the synergy patterns happen directly after the stroke and remain consistent [21], implying that there is an initial fundamental change in the pathways used for muscle recruitment, rather than an adaptation. For example, more severely impaired patients can generate movement in different directions but have difficulty grading movement direction [8]. These stereotyped movements would fit the theory that resultant synergy patterns are an inherent response to brain damage, because the increased damage in more severe patients would require greater reliance on alternative pathways that are not primarily used for fine control. The primary reliance on motor pathways used for gross movements could explain why severely impaired patients exhibit stereotyped reaching patterns and limited ability to grade the direction of movement.

Both cortical drive and the sensory response are impaired after stroke. Poststroke stretch reflexes [22-27] are more sensitive to perturbations [26] and elicit a stronger response when compared to able-bodied control subjects. However, voluntary activation produces a smaller change in reflex strength in stroke patients than in able-bodied controls. For stroke patients, fewer muscles are required to account for the EMG variability in the reflex response, suggesting that the increased response to afferent input is not equally distributed across motor pathways [22]. Arm orientation changes sensitivity to perturbations, further suggesting a proprioceptive component to the impairment [26]. Unlike in controls, poststroke reflex response is coupled across joints, which also points to an
increased contribution from brainstem pathways [28]. Increased low level
excitation to motor neurons [29] provides a physiological basis for more sensitive
reflexes.

In addition to changes in reflex responses and the extent that muscle
groups are activated, there are also changes to how the individual motor units
are recruited after stroke. Alterations in poststroke motor unit recruitment appear
to be characterized by earlier recruitment and reduced firing frequency [30] as
well as a potential reduction in the total number of motor units recruited [31].

**Therapeutic Approaches**

Poststroke functional improvements and the resulting brain adaptions in
human and animal models imply that movements can be relearned and the brain
can be retrained, thereby restoring lost function. This relearning potential
motivates current and researched therapies. The current recommendation from
the Department of Veterans Affairs [32] is that poststroke therapy should be goal-
oriented, with a focus on patients practicing functional activities. As patients
improve, the difficulty of tasks attempted should progressively increase.
Additional recommendations include cardiovascular exercise and strengthening
and stretching the affected limb.

Motor control after stroke can be improved by therapies that emphasize
practice, performance feedback, task-engagement, and complex problem solving
[33, 34]. However, a large percentage of stroke survivors, particularly those with
the most significant impairments, gain little functional improvement with current
treatments. Several research efforts are underway to improve the outcomes of upper limb therapy. These incorporate robotic training [35-41], constraint induced movement therapy [42], functional electrical stimulation [43-46], virtual reality [47], and cortical stimulation [48, 49] into the therapy approaches. Therapeutic approaches have shown promise but often produce limited gains. Participants who receive the most benefit from these approaches tend to be more mildly impaired, while more severely impaired patients receive less benefit, suggesting that additional assistance during everyday tasks may be beneficial for these patients.

**Assistive Device for Activities of Daily Living**

While therapeutic approaches show promise, a significant portion of the stroke population retains motor deficits after completing therapy [32]. Thus, another approach has been to restore function by means of an assistive device during everyday tasks. Devices used typically include a powered exoskeleton [50] or an FES system [51-53] to augment movement. With either type of system, the user traditionally controls the assistive device with a physiological command signal (e.g. EMG, EEG, limb mechanics). The system then assists in movements by either stimulating relevant muscles or moving the arm with a robotic attachment. The general concept is shown in Figure 1.1. This type of system can be used in conjunction with residual voluntary effort to complete the desired task.
Popovich et al. [54] outlined a few considerations for a poststroke FES system. The system can either be implanted or use surface electrodes. Popovich made the point that implanted systems should be targeted towards spinal cord injury (SCI) patients and surface systems for therapy should be targeted towards stroke patients; however, there may be a limit to the benefit that can be regained through therapy. Stroke patients still require arm and hand assistance even after therapy, which could be provided by an implanted system. An implanted system is preferable because it would simplify the necessary donning and doffing, provide consistency in recording and stimulation, and allow greater stimulation.
with less discomfort. Additionally, the system must be easy to control and require little cognitive load to operate, which has implications for how the command signal is derived. It is important that the system provide effective movement and have a natural control scheme because use of the assistive system cannot detract significantly from the use of the unaffected arm and hand.

Despite challenges faced in limited clinical evaluations of upper extremity poststroke FES systems, an increased understanding of the synergy patterns suggests that FES could still be an attractive solution for providing functional bimanual capabilities. Currently, upper limb applications have focused almost exclusively on providing hand function. The perspective is that a population of stroke survivors exist who retain the ability to reach and grasp, but lack the ability to open the hand. As a result, the focus has centered on developing a device that stimulates finger extensors to restore hand function. The approach focusing on hand opening has been tested in multiple laboratories [52, 55-59] and has been tested with stimulation of some finger flexion and elbow extension muscles [53]. A principal finding of Chae’s attempt to implement a hand-only FES system in the stroke population showed that enhanced proximal arm control should be incorporated into the total system because of the deleterious interactions between reach and grasp. When the user was relaxed, stimulation was able to produce effective hand opening, but effort to reach with the arm or try to open the hand produced involuntary co-activation and co-contraction that overpowered the stimulation, preventing hand opening.
Limited Effort Poststroke Neuroprosthetic Approach

Our proposed approach uses the outlined framework for a poststroke assistive device, while taking into consideration what has already been learned about involuntary co-activation. The device described here is a neuroprosthesis, which is an assistive device that interacts with the nervous system. In this case the neuroprosthesis is an FES system that drives movements by stimulating muscles.

Considering the undesired co-activation effects produced by voluntary effort, our approach is to incorporate shoulder muscle stimulation into the system and use low levels of voluntary effort as a command signal for the neuroprosthesis. The resulting stimulation then produces a significant amount of the movement. Graded effort is detected by recording electromyograms (EMG) from shoulder, arm, and hand muscles on the paretic side of the body. Based on the detected effort, the system applies stimulation to shoulder, arm, and hand muscles on the same side to assist in producing the movement in conjunction with the voluntary effort. The flow chart for the general concept is shown in Figure 1.2. Using residual effort on the paretic side allows the unaffected side to be the primary actor during bimanual tasks. The goal of the neuroprosthesis is to restore sufficient function in the affected arm for assistance in bimanual tasks, such as writing, cutting food, and dressing (e.g. buttons, zippers, socks, pants).
An important aspect in the neuroprosthetic approach is reducing effort on the affected side. Using low levels of effort as the command signal reduces the expression of the co-activation patterns. Reduced voluntary effort allows stimulation to have a greater effect but also requires producing more of the movement by stimulation. Reduced voluntary activation may also require
producing some of the shoulder movement through stimulation in individuals who can abduct their shoulder, but the resultant co-activation limits too much of the stimulation response at the elbow/hand.

**Command Signal**

EMG may provide an effective command signal for the assistive system, considering that lower motor neurons still receive inputs from motor pathways. Features of EMG signals have been extracted for use as a command signal [60] and have been used for myoelectric control of amputees’ prostheses [61]. If poststroke EMG provides sufficient information for a command signal, muscles can be chosen for recording based on an understanding of the muscle’s action and contribution to the different synergy patterns. Despite the limited movements that result from the synergy patterns, there appears to be the same number of independent combinations of muscle activity between able-bodied and stroke patients [62]. The fact that both able-bodied and stroke force generation can be decomposed into the same number of independent synergistic patterns suggests that despite limited limb movement, affected limb movement may contain sufficient information to provide a command signal. Independent control of these synergistic subsets has not been assessed, but would provide further insight into the effectiveness of residual effort as a command signal. Further support is provided by participants’ ability to generate independent movements, like reaching in a circle, during involuntary co-activation [9]. Similarly, a muscle’s
ability to perform a specific action does not guarantee that the muscle is used during all movements and postures where that action is performed [63, 64].

*Mapping Estimates to Stimulation*

After extracting the appropriate features from the EMG, the features need to be mapped to appropriate stimulation patterns to drive the muscles. Time Delayed Artificial Neural Networks (TDANN) can map continuous inputs and outputs of non-linear systems such as EMG to continuous position [65], joint angle [61, 66], force [67-69], stimulation levels [70, 71], trajectories [72], and fatigue effects [73]. Discrete classifiers have also been used to classify poststroke movements or attempted movements using EMG, EEG, and accelerometers [74-78]. Most estimation approaches map the input to what the individual can do [61, 65, 66, 72], but some map activity to what the user is attempting to do [70, 71, 74].

The constrained poststroke movement patterns could produce a potential challenge if directionally independent active forces cannot be generated. Constrained movements suggest that the ability to generate independent forces may be limited, which would make it difficult to extract independent command signals. However, these constrained movements do not preclude the ability to detect intent. Roh et al. demonstrated that EMG contributions to the synergy patterns differ after stroke, but can be separated into distinct patterns. Hand intent has been classified with high accuracy (96%), even though participants
had difficulty producing the movement [76]. Despite redundant limb kinematics, affected limb EMG can be modulated to provide an effective command signal.

There are two primary approaches for training the network relating input EMG and output stimulation levels. One method is to assume that the user is attempting the instructed movement and to map EMG to the instructed movement instead of the resultant limb kinematics. This assumption requires that unique EMG patterns are produced during redundant limb kinematics. Another option is to train the network based on the individual’s residual movement, and then create a second map from residual active movement to the desired movement. Mapping residual activity to desired movement would be most effective with joint torques or endpoint forces because these changes can be produced without actual movement.

A non-linear mapping between residual activity and desired activity allows for the creation of the most effectively controlled workspace. There are three fundamental components of the map: 1) Offset – This is a translational shift of the limb’s equilibrium point. The offset biases the endpoint force, position, or individual joint torques, 2) Gain – Amplifies volitional activation, allowing small amounts of effort to produce larger movements or forces, and 3) Rotation – A rotational shift of the direction of action for volitional activity to shift useful levels of control into a more functionally relevant workspace. Linear examples of the offset, gain, and rotation maps are shown in Figure 1.3. When implemented in an assistive device, the complete map would incorporate a combination of the three
fundamental transformations or a nonlinear combination that is dependent on joint kinematics.

Each mapping component has a particular benefit. The offset helps to shift the resting position in a direction that is not easily accessible; for example, constant force can be used to support the arm if the person has difficulty lifting the arm. A constant force could also push the arm forward if they have difficulty actively extending their elbow. The gain condition amplifies effort to produce the movement, thereby reducing both effort exerted and as a result, expression of the synergy patterns. For example, a gain of 2 could enable reaching while exerting half effort in some muscles. The action of rotation is similar to that of the translational offset, except that it shifts direction instead of the resting forces. If someone has difficulty pushing in a particular direction, their workspace can be rotated so that force generation that is well controlled in a certain direction is shifted to control forces in a more functionally relevant direction.
Figure 1.3: Overhead view of examples of the three basic mapping methods (offset, gain, and rotation). The black line and blue ellipse represent the active range of motion (arom) without assistance. The red line and ellipse represent the arom with the indicated mapping method.

When creating the composite map as a function of the transformation components, it is important to understand how well people can control both their residual movements and the assistive movements that are created as a result of the transformation maps. Motor control research has used force fields to evaluate how well people adapt to position dependent forces that affect their movements [79]. These force fields are primarily an obstruction that needs to be compensated for in order to complete a reaching task; however, our goal is to provide an assistive field that is controlled by the user to facilitate task completion. Prior work focusing on how people learn and adapt to the fields demonstrates that people can internalize the adaptations into their internal motor control model and can learn consistent mappings that are applied to their movements [79].
Similar to understanding adaptation to assistive forces, it is useful to know the magnitude of the changes, such as the force fields, that can be applied to residual function without causing the user to become aware of the assistance. Brewer et al. observed that the Just Noticeable Difference (JND) in joint position is larger in the elderly and appears to be even greater in neurologically impaired subjects [80, 81]. The larger JND indicates that proprioceptive responses are not so sensitive that proprioception prevents the body from adapting to or incorporating large changes to the internal motor model without automatically creating a large cognitive load. The force fields are not limited to being produced by a haptic interface. Forces and torques produced by the FES system would effectively produce a comparable force field that changes in response to voluntary effort instead of position.

In an effort to further understand poststroke impairments, research groups have evaluated the use of the mappings described here. Antigravity forces to support the weight of the arm have enabled stroke subjects to reach a greater distance [9, 10] and have decreased the involuntary finger and wrist flexion [12]. Similarly, lateral forces have also enlarged the active range of motion, presumably because stroke patients can use the extension synergy to increase reach [82]. Able-bodied individuals are also able to control gain assistance that amplifies volitional activity. Studies have evaluated effort amplification along a single degree of freedom using a robotic [83-85] or FES [86] system to assist able-bodied participants with their movements. Similar to the rotation, groups
have studied how well able-bodied people can learn to control intuitive and non-intuitive mappings between EMG and cursor movement on a screen [87]. Wright and Rymer used a similar approach to help train stroke patients to unlearn synergy patterns [88]. Participants learned to control the mappings, indicating that stroke patients can learn non-intuitive maps from their residual function to a mapped assistive workspace.

Poststroke impairment may be so severe for some patients that a direct mapping from residual function to assistive patterns may be too difficult to control. While it could enlarge the workspace, the enhanced movements may not be controllable. Instead of a static mapping, the map could dynamically respond to effort. A dynamic map could be implemented with a dynamically changing bias that responds to effort over time. By restricting bias shifts to slower changes, an integrator could add intent over time. This would allow slow changes to the offset without requiring large gains the user cannot control. Instead of creating a single map that enables reach to the entire workspace, this enables mapping to a small workspace that can be dynamically shifted. This would enable people to move their arm to the table or near their face and complete their current activity while maintaining a stable posture.

*Producing Movement Poststroke*

After creating a TDANN relating EMG to voluntary force generation and creating a map between voluntary effort and assistance, the required assistance can be used to generate the necessary stimulation patterns to produce the
desired force or movement as shown in Figure 1.4. EMG estimates provide an estimate of voluntary force while the map determines the total transformed force. The difference between the voluntary estimate and the desired force created by the mapped forces determines the additional force the FES system needs to generate. An inverse musculoskeletal model [89] can calculate the activation necessary to produce the assistive forces as shown in Figure 1.2.

![Diagram of the controller concept](image)

**Figure 1.4:** Flow of information for the controller concept; including EMG processing, mapping the voluntary force estimate to the desired force, and the inverse model to produce stimulation parameters.

The inverse musculoskeletal model produces the normalized muscle activation required to produce the desired forces, but it also requires understanding the relationship between stimulus parameters and muscle activation. This relationship is characterized by the muscle’s force response to different stimulation parameters. Force response curves have been measured in steady state conditions [90-92] and modeled dynamically [93]. The model used to create this relationship between stimulus parameters and force must account for
mechanical muscle properties impact on the stimulation response, considering the effect of passive movement on stimulation response [94]. In addition to muscle’s mechanical muscle properties impacting the stimulation response, the stimulation response changes during simultaneous volitional muscle activation [95]. Stimulating a volitionally activated muscle produces a smaller increment than stimulating a relaxed muscle. Considering the previously described changes to poststroke motor recruitment, it is important to examine these non-linear summations in stroke.

The input commands and output stimuli in the neuroprosthesis are not independent. Stimulated muscle forces or externally applied forces can elicit a stretch reflex in antagonist muscles. If that antagonist contributes to the command signal, the resulting reflex could be interpreted as a command to produce force or movement in the opposite direction. It is therefore important to test the estimates provided by such a system in consideration of the assistance producing EMG in reflex responses. The model that generates the stimulus parameters may need to account for reflex behaviors.

In addition to detecting a command signal from paretic muscles and determining the map from command signals to stimulus parameters, we must choose muscles to stimulate and appropriate stimulation parameters. Most FES systems in SCI utilize command signals above the level of injury or significantly below the level of injury while stimulating muscles below the level of injury [96-100]. Few involve stimulating muscles that are already active [101].
Deciding which muscles to stimulate has additional complications in the stroke population. Unlike spinal cord injuries in which the user has control over specific muscles and drastically reduced contributions from other muscles, in stroke all of the muscles on the hemiparetic side are affected and weakness is expressed to different extents. When choosing the muscles to stimulate it is useful to know how much force or torque needs to be generated in a particular direction and how much force the user’s residual activity can generate. An upper extremity poststroke system would need to be able to produce effective reach and hand opening even in the presence of the effort required to produce the command signal. This requires limiting exerted effort so the co-activation patterns do not overpower the stimulation response.

Project Specific Aims

*FES for reach and hand opening during limited effort*

There is evidence that a poststroke neuroprosthesis could be effective, but the approach needs to consider how the impairments impact use of the FES system. Keller *et al.* showed significant variation in how well stimulation can produce elbow extension torques and that with increasing shoulder abduction effort, the elbow extension stimulation response decreases [20]. Stimulation has been shown to produce effective hand opening when the participant is relaxed, but not when they are exerting effort for reach or hand opening [52]. These approaches tended to rely on the user exerting maximal effort while the stimulation provided assistance to produce the desired movement. Miller *et al.*
showed proportional co-activation at the hand in response to voluntary shoulder abduction and reach [12], which is likely the cause of the decreased stimulation response. While stimulation produces additional muscle force, stimulation cannot always overpower the co-activation [20, 52, 53].

Our approach reduces the user's effort, thereby limiting the expression of undesired synergy patterns. For this approach to be effective, it requires that FES applied for reach provides enough reaching torque to sufficiently reduce reaching effort so the co-activation patterns do not overpower stimulation at the hand. While studies have shown that stimulation produces useful hand opening while the participant is relaxed and that reach and hand opening effort can overpower this stimulated hand opening in some patients, the effect of this effort has not been well quantified or documented. Measuring reach and hand opening during different levels of stimulation and voluntary effort can provide insight into how much of an impact effort has on stimulated hand opening. Additionally, measuring the stimulation response with and without effort would help determine if stimulation can effectively produce reach and hand opening in concert with residual voluntary effort.

**Specific Aim 1a:** Determine if submaximal reaching effort combined with stimulation of reaching and hand opening muscles will result in larger hand opening and greater reaching distance than during maximal voluntary reaching.

**Specific Aim 1b:** Quantify the effect of effort to open the hand on stimulated hand opening.
The interaction of voluntary effort and FES

While limiting effort can limit the expression of the synergy patterns, some effort is still necessary to generate a command signal and contribute to the desired movement. Muscles that are either involuntarily co-activated or volitionally activated might also be stimulated to increase force. Therefore, it is important to know how voluntary and stimulated activation interact to generate a net force. Perumal et al. showed reduced stimulation response in able-bodied individuals [95] in response to voluntary effort. As a result, more stimulation may be required to elicit the desired torque [102], or the stimulation response could potentially be reduced if concurrent effort is no longer beneficial.

Potential mechanisms contributing to the reduced stimulation response include occlusion of muscle force generation due to activation of the same motor units [103]. Considering poststroke changes to muscle properties and motor recruitment [30, 31, 104], it is important to investigate this interaction in the stroke population as well. This is also important considering that a higher level of muscle activation may be required to produce a particular net torque at a joint as a result of the co-contraction patterns.

**Specific Aim 2a:** Determine if FES of elbow extensors increases forward force during voluntary effort.

**Specific Aim 2b:** Evaluate the effect that increasing effort has on the stimulation response.
Controlling assistance with residual poststroke movement

For a poststroke neuroprosthesis to be effective in increasing function, it is insufficient to produce arm and hand movement. Ease of use requires that the user control it in an intuitive manner related to their residual activity. One method to control the stimulation is using the mappings described previously. It has been demonstrated that able-bodied individuals can control single degree of freedom amplification using EMG signals, reducing the effort exerted [84, 85, 105]. Stroke patients are able reach further when lateral and antigravity forces are applied to the arm [9, 82], and have also adapted to amplification transformations to their movements and errors [106-108] as well as changes to how EMG signals control cursor movement [109].

Control of assistive forces with residual effort for 3-dimensional reach has not been shown before in the stroke population. While assistive forces have been shown to increase the accessible workspace, the quality of reaching to and holding a target position with assistance has not been evaluated. Quantifying how well people control assistive forces from a powered exoskeleton provides insight into how well people could control FES for reach and hand opening assistance.

**Specific Aim 3a:** Determine if stroke patients can control assistive forces to increase reach while maintaining controllability.

**Specific Aim 3b:** Determine if the movements can be performed while exerting less effort.
The current aims address the core issues of whether useful stimulation responses can be achieved in the presence of voluntary effort and whether people could control these assistive forces with their residual voluntary effort. The answers to these questions should determine if a poststroke neuroprosthesis is feasible.
Chapter 2 – Functional Electrical Stimulation to augment poststroke reach and hand opening in the presence of voluntary effort.

This chapter has been submitted for publication: Makowski, N.S., Knutson, J.S., Chae, J. and Crago, P.E. (2013), Functional Electrical Stimulation to augment poststroke reach and hand opening in the presence of voluntary effort.

Abstract: Hemiparesis after stroke can severely limit an individual’s ability to perform activities of daily living. Functional Electrical Stimulation (FES) has the potential to generate functional arm and hand movements. We have observed that FES can produce a functional amount of hand opening when a stroke patient is relaxed, but the FES-produced hand opening is often overpowered by finger flexor co-activation that occurs when the patient reaches and attempts to open the hand. Therefore, simultaneous reaching and FES hand opening is difficult to achieve by applying FES only to muscles that open the hand. However, stimulating both reaching muscles and hand opening muscles may make it possible to achieve useful amounts of simultaneous reach and hand opening even in the presence of submaximal reaching effort. Therefore, we measured reach and hand opening during a reach-then-open the hand task under different combinations of voluntary effort and FES for both reach and hand opening. As effort was reduced and stimulation generated more of the movement, a greater amount of reach and hand opening was achieved. For the first time, this study quantified the effect of voluntary effort for reach and hand opening on stimulated hand opening. It also showed variability in the interaction of voluntary effort and
stimulation between participants. Additionally, when participants were instructed to reach with partial effort during simultaneous FES, they achieved greater amounts of reach and hand opening. In the future, an upper extremity neuroprosthesis that utilizes a combination of voluntary effort and FES assistance may enable users to perform activities of daily living.

**Introduction**

Poststroke hemiparesis limits arm and hand function, making bimanual tasks difficult if not impossible[110]. Hemiparesis is worsened by disuse and involuntary co-activation patterns across multiple joints (i.e. synergy patterns)[11]. For example, when a stroke patient abducts and/or flexes their shoulder (as when reaching), the biceps and forearm flexors may involuntarily activate (flexor synergy) at the same time, preventing lifting the arm and reaching forward simultaneously. These synergy patterns have been well quantified[11, 12, 111] and appear to be expressed in proportion to shoulder abduction and reaching effort[9, 10, 12].

Functional Electrical Stimulation (FES) of paretic muscles has the potential to generate functional reach and hand opening and has been used successfully to improve function after spinal cord injury[71, 97, 112]. Similarly, electrical stimulation of finger extensors in stroke[51, 52, 54, 94, 113, 114] produces hand opening while the participant is relaxed. However, when the user exerts effort to reach or to open the hand during stimulation, the hand does not open as much as when the person remains relaxed[57, 115], presumably
because effort to open the hand produces involuntary finger flexor contractions[16, 52, 57]. Voluntary hand opening effort reduces FES extension of the metacarpal phalangeal joint[57]; therefore, maximal stimulated hand opening is achieved when the user remains relaxed. Similarly, voluntary shoulder abduction and reaching effort produces a net wrist and finger flexion force and forearm co-contraction[12]. Elbow extension due to triceps stimulation is limited by both simultaneous voluntary shoulder abduction[20] and increased voluntary effort in stimulated muscles[95, 116]. These results strongly infer that voluntary effort for reach and hand opening will interfere with FES hand opening, but do not directly quantify the effect of voluntary effort on stimulated hand opening.

Our long-term objective is to develop a neuroprosthesis that restores arm and hand function in moderate to severely impaired stroke survivors. Ideally, residual voluntary movement of the same limb will control stimulation to produce desired arm and hand movements. Considering the proportional gradation of forces at different joints produced by synergy patterns, partial effort may not completely limit the effects of stimulation, allowing useful levels of hand opening. Therefore, our aim is to achieve adequate FES-generated movement even in the presence of some simultaneous effort.

The goal of this pilot study is to provide estimates of the potential hand opening that could be achieved with an FES system if simultaneous voluntary reaching and hand opening effort are reduced. There are two hypotheses: 1) submaximal reaching effort combined with stimulation of reaching and hand opening
opening muscles will result in larger hand opening and greater reaching distance than maximal voluntary reaching effort with stimulation of hand opening muscles, 2) during stimulation of hand opening muscles and stimulation for reach, zero simultaneous hand opening effort will result in greater hand opening than maximal hand opening effort. The resulting preliminary data will provide a basis for designing experiments to evaluate the interaction of voluntary effort and FES in the general stroke population and provide an initial assessment of the hypotheses. If these hypotheses are true, partial voluntary effort may provide a command signal for an FES system that produces functionally relevant reach and hand opening.

Methods

Participants

Participants were recruited from an outpatient stroke clinic. Primary inclusion criteria included: 1) at least 6 months post-stroke, 2) able to follow 3-stage commands, 3) able to reach forward at least 10 cm while the investigator supported the elbow and wrist, 4) unable to fully reach and open the hand while the arm is unsupported, 5) functional upper extremity passive range of motion, and 6) functional stimulated hand opening while the hand and arm are relaxed. All participants provided written informed consent prior to participation in this study, which was approved by an Institutional Review Board. An upper limb Fugl-Meyer Motor Assessment (FMA) and modified Ashworth test (mAsh) characterized each participant’s upper limb motor impairments.
Seven participants enrolled in the study, and five completed the protocol. Participants’ demographic information is in Table 2.1. Subjects 6 and 7 did not complete the protocol for the following reasons: 1) when we stimulated the arm, subject 6 could not avoid simultaneous volitional effort to lift the arm, and 2) we could not identify electrode positions for subject 7 that produced consistent hand opening.

Table 2.1: Participant Demographics and Settings.

<table>
<thead>
<tr>
<th>Participant</th>
<th>Age</th>
<th>Affected/Dominant Side</th>
<th>Time since stroke</th>
<th>FMA (arm)</th>
<th>mAsh elbow flexors</th>
<th>mAsh finger flexors</th>
<th>Average support provided by mobile arm support in all trials (N)</th>
<th>Average arm weight at target positions (N)</th>
<th>Distance between the shoulder and wrist at far target (cm)</th>
<th>Vertical wrist distance below shoulder at far target (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>S1</td>
<td>60</td>
<td>R/L</td>
<td>9 yrs</td>
<td>13</td>
<td>1+</td>
<td>1</td>
<td>26</td>
<td>40</td>
<td>54/22</td>
<td></td>
</tr>
<tr>
<td>S2</td>
<td>55</td>
<td>R/R</td>
<td>5 yrs</td>
<td>22</td>
<td>3</td>
<td>2</td>
<td>24</td>
<td>43.5</td>
<td>54/14</td>
<td></td>
</tr>
<tr>
<td>S3</td>
<td>65</td>
<td>L/R</td>
<td>1 yr</td>
<td>13</td>
<td>1+</td>
<td>1</td>
<td>30</td>
<td>45</td>
<td>60/12</td>
<td></td>
</tr>
<tr>
<td>S4</td>
<td>58</td>
<td>R/R</td>
<td>3 yrs</td>
<td>29</td>
<td>*</td>
<td>*</td>
<td>24</td>
<td>54</td>
<td>64/16</td>
<td></td>
</tr>
<tr>
<td>S5</td>
<td>67</td>
<td>R/R</td>
<td>3 yrs</td>
<td>35</td>
<td>1+</td>
<td>1</td>
<td>2.5</td>
<td>40</td>
<td>*Near Position 50/14</td>
<td></td>
</tr>
</tbody>
</table>

Abbreviations – Side: R-Right, L-Left; FMA: upper limb portion of Fugl-Meyer Motor Assessment, 66 point maximum, shoulder/elbow/forearm in parentheses, 34 point maximum; mAsh: modified Ashworth test, 5 point maximum

Setup

Participants performed a series of reach and open the hand trials with their paretic arm (described below). Starting with the wrist positioned at the base of the sternum and the elbow flexed 90°, participants were cued to reach towards a target and attempt to open their hand under a variety of stimulation and effort conditions. A participant is shown performing the task in Figure 2.1a. We recorded arm position with an optical tracking system (Optotrak), and recorded
surface electromyograms (EMG) from flexor digitorum superficialis and extensor digitorum communis. A custom device recorded hand opening (measured as the distance between the fingers and the tip of the thumb) and grasp force[117].

Two computer controlled stimulators activated muscles for reach and hand opening. Surface stimulation electrode sizes ranged from 1.25” round to 2”x5” rectangular. We targeted the anterior and middle deltoids for shoulder abduction and flexion, and triceps for elbow extension. We also stimulated subject 4’s biceps to prevent overshooting the near target. The goal for stimulated hand opening was to produce finger extension, thumb abduction, and thumb extension by stimulating extensor digitorum communis, extensor pollicis longus, and abductor pollicis brevis.

Surface stimulation produced hand opening and elbow extension without pain, but it was difficult to recruit adequate shoulder abduction and flexion to achieve full arm elevation. Therefore, an elevating mobile arm support (Jaeco) provided an upward force at the forearm in all trials, reducing the shoulder force.
that FES needed to generate. The goal of the arm support was to provide participants with enough support to enable them to volitionally reach approximately halfway between their resting position and their maximum forward passive reach. Subject 4 was able to lift his arm on his own, but had difficulty independently controlling the elbow at the same time; therefore, for subject 4 the purpose of the support was to allow the stimulation to produce maximal reach.

Experimental Procedures

During the first experimental session we determined electrode positions and stimulation levels that produced reach and hand opening without eliciting pain while the participant relaxed. Then, over several sessions, participants learned the reach and open tasks under 12 combinations of 4 reaching conditions and 3 hand opening conditions, which are described in Table 2.2. Participants were cued to reach toward a target in front of them under one of the reaching conditions (RE, RM, RS, RES). Once the arm came to a steady position, participants were instructed to maintain their reaching effort while attempting to open their hand under one of the hand opening conditions (HE, HS, HES). Trial timing is illustrated in Figure 2.1b.
Table 2.2. The four Reaching and three Hand Opening conditions that comprise the 12 task conditions

<table>
<thead>
<tr>
<th>Four Reaching Conditions</th>
<th>Three Hand Opening Conditions</th>
</tr>
</thead>
<tbody>
<tr>
<td>RE = voluntary effort alone</td>
<td>HE = maximum voluntary effort alone</td>
</tr>
<tr>
<td>RS = stimulation of shoulder and elbow</td>
<td>HS = stimulation of hand opening muscles</td>
</tr>
<tr>
<td>muscles to bring the wrist to the target</td>
<td>while the participant remained relaxed</td>
</tr>
<tr>
<td>while the participant remained relaxed</td>
<td></td>
</tr>
<tr>
<td>RES = partial voluntary reaching effort</td>
<td>HES = maximum voluntary effort to open</td>
</tr>
<tr>
<td>and the same stimulation as in RS.</td>
<td>the hand and the same stimulation as</td>
</tr>
<tr>
<td>Participant was asked to limit reaching</td>
<td>condition HS</td>
</tr>
<tr>
<td>effort to half of their maximum</td>
<td></td>
</tr>
<tr>
<td>RM = mechanical support holding the arm</td>
<td></td>
</tr>
<tr>
<td>at the target while the participant</td>
<td></td>
</tr>
<tr>
<td>remained relaxed</td>
<td></td>
</tr>
</tbody>
</table>

Once the participant could perform the reach and open task consistently under the 12 conditions, data (i.e., amount of hand opening, reaching distance, EMGs) were collected. Two separate sessions were used to record data for the near and far target respectively. The near target position was half of the participant's maximum active reach distance. The far target position was the furthest position that shoulder and elbow stimulation could achieve while the subject remained relaxed. The distance between the shoulder and the wrist and the vertical distance of the wrist below the shoulder during manual support (RM) to the far target are included in Table 2.2. Each participant performed at least three trials under each combination of conditions.
Data Analysis

The last second of each trial was used to calculate hand opening and the distance from the target to the wrist. If all of a participant’s trials of a single reach and hand opening condition had hand opening increase less than half a centimeter, the result was treated as no hand opening. This prevented passive hand movement during reach from appearing like active hand opening.

Analysis of variance (ANOVA) models were used to compare hand opening and the distance from the target. The models included subject as a random factor while reaching condition, hand opening condition, and target position were fixed factors. Interaction effects were included for the fixed effects. If factors were statistically significant, the Tukey-Kramer comparison of means was used to determine which conditions were statistically different. The distribution of residuals was statistically different from a normal distribution (p<0.05, Kolmogorov-Smirnov test). Contributing factors were residual outliers and residual skewness of -0.22. A visual inspection of residuals’ normality suggests the difference from a normal distribution does not alter p-values sufficiently to disbelieve the results considering that analysis repeated with outliers removed produced the same conclusions.

To assess the primary hypotheses, we compared hand opening and the distance from target in subsets of the data: Hypothesis 1) the combination of partial reaching effort with stimulation plus relaxed hand with stimulation (RES+HS) would produce greater hand opening and further reach than the
combination of full reaching effort plus relaxed hand with stimulation (RE+HS), and Hypothesis 2) the combination of stimulation for reach with a relaxed arm plus relaxed hand with stimulation (RS+HS) would produce greater hand opening than the combination of stimulation for reach with a relaxed arm plus maximum effort to open the hand with hand stimulation (RS+HES).

To assess the net effect of reach on the hand we compared hand closing forces and forearm EMGs. We compared hand closing forces during voluntary reaching effort (RE) under the three hand opening conditions (HE, HS, and HES). We compared forearm EMG generated during the combination of reaching effort plus hand opening effort (RE+HE) and during the combination of mechanical support plus hand opening effort (RM+HE). These provide insight into forearm co-activation in response to reaching effort. The three analysis time windows were one second long: 1) at the beginning of the trial while the participant was relaxed, 2) during reach prior to hand opening, and 3) during the final second of hand opening.

**Results**

Reducing voluntary reaching and hand opening effort and augmenting it with stimulation increased hand opening at reaching distances that were both within and beyond participants’ voluntary reaching distance. Hand opening increased with stimulation and reduced levels of reach and hand opening effort. These findings support both hypotheses: Hypothesis 1 – hand opening during stimulated reach with partial reaching effort and stimulated hand opening (RES+HS) was greater than during voluntary reaching effort and stimulated hand opening (RE+HS) (p<0.05); and Hypothesis 2 – hand opening was greater during stimulated reach and stimulated hand opening (RS+HS) than during stimulated reach and stimulated hand opening with maximal hand opening effort (RS+HES) (p<0.05). Figure 2.2 shows the average hand opening across all participants and
target positions for the twelve combinations of reach and hand opening.

![Hand Opening Chart](image)

**Figure 2.2:** Amount of hand opening (mean±SD) averaged across all participants for the 12 task combinations. The passive closing limit (3cm) is the smallest amount of hand opening that the sensor could measure.

*Hand Opening*

The reaching condition × hand opening condition interaction was significant (p<0.001) while the other interactions were not significant: reaching condition × target position (p=0.709) and hand opening condition × target position (p=0.692). Significant main effects were reaching condition (p<0.001), hand opening condition (p<0.001), and participant (p<0.001) but not target position (p=0.196).

The model was simplified to calculate effect means and variances shown in Table 2.3. Target position was removed from the model because it was not statistically significant in the interaction or main effects. Since there was minimal voluntary hand opening without stimulation (HE), that data was also removed.
from the model. The random effect of participants accounted for 2.1 cm of the variance while model error variance was 1.5 cm.

Table 2.3: Mean effects for hand opening response during the different reach and hand opening conditions.

<table>
<thead>
<tr>
<th>Hand Opening Conditions</th>
<th>Mean Effects (cm)</th>
<th>Reaching Conditions</th>
<th>Voluntary Effort (RE)</th>
<th>Partial Effort and Stimulation (RES)</th>
<th>Stimulation Alone (RS)</th>
<th>Mechanical Support (RM)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stimulation Alone (HS)</td>
<td>5.8</td>
<td>7.9</td>
<td>8.7</td>
<td>9.2</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hand Opening Effort and Stimulation (HES)</td>
<td>5.5</td>
<td>6.5</td>
<td>7.2</td>
<td>7.4</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

We also looked at the effects of reaching and hand opening effort in individual participants. The effects of reducing reaching effort while stimulating reach and hand opening are shown in Figure 2.3a. The effects of voluntary effort to open the hand during hand stimulation and stimulated reach are shown in Figure 2.3b. Variation between subjects is large, ranging from virtually no effect for either type of effort (S2) to large effects of voluntary reach (S1, S4, S5) and a large reduction of hand opening with voluntary hand opening effort (S1).
Figure 2.3 (a,b): a) Amount of stimulated hand opening (mean ± SD) during RE+HS and during RES+HS for each participant. b) Amount of hand opening (mean ± SD) during RS+HS and during RS+HES for each participant.

Reaching Distance

Reaching distance was measured as the distance from the target to the wrist. To simplify comparisons across participants and sessions, we subtracted the shortest average distance from the target in any of the 12 conditions from all of the trials for that participant. The reaching condition × target position had a significant interaction (p<0.001). Non-significant interactions included target...
position × hand opening condition (p=0.652) and reaching condition × hand opening condition (p=0.625). The primary effects had the following p values: reaching condition (p<0.001), target position (p<0.001), participant (p<0.001), and hand opening condition (p=0.477).

The primary factor that affected distance from the target was reaching condition. Reaching distance from target is shown in Figure 2.4 for each participant during the different reaching conditions at the far target position. For Subjects 1, 3, and 5, voluntary reaching (RE) was difficult and still left them at least 15 cm further away from the far target than with stimulation (RS or RES). With stimulation (RS and RES), all of the participants were able to reach both targets. Partial reaching effort did not prevent the participants from reaching the targets.
Figure 2.4: Reaching distance from the far target (mean±SD) during different reaching conditions.

Hand Closing Force

The combinations of reach and hand opening where the hand produces a compression force instead of opening provide additional insight into stimulation’s effects. Figure 2.5 shows the effect of reaching effort (RE) and subsequent hand opening conditions (HE, HS, HES) on hand closing force. Subjects 1,3,4, and 5 appear to increase hand closing force during reach before adding the hand opening condition. For Subjects 4 and 5, there was no further increase in hand closing force during maximum effort to open the hand (HE). When stimulation was added at the hand (HS and HES) during voluntary reaching effort alone.
(RE), stimulation overpowered the flexion forces and produced hand opening as shown in Figures 2.2 and 2.3a. However, for Subjects 1 and 3, hand closing force increased during HE. Hand opening stimulation (HES) overpowered these forces for Subject 3, but not for Subject 1.

Figure 2.5: Grasp force during voluntary reaching and different hand opening conditions. The horizontal axis shows the progression from being relaxed, to reaching, to reaching while opening the hand. The different lines represent the three hand opening conditions. The top plot shows averaged data for Subjects 4 and 5. The middle plot shows data for Subject 3, and the bottom plot shows data for Subject 1.

Forearm EMG

Forearm flexor and extensor EMGs showed a trend of increasing with either reaching or hand opening effort. EMG magnitudes were averaged across
all of the available participants (Subjects 2-5). During voluntary reach and hand opening effort trials (RE+HE), average flexor/extensor EMGs were 10µV/8µV while relaxed (before attempting reach), 15µV/21µV during voluntary reach (RE before attempted HE), and 35µV/34µV during reach and hand opening (RE+HE). During mechanical support and effort to open the hand (RM+HE), flexor/extensor EMGs were 10µV/9µV while relaxed (RM before attempted HE) and 22µV/27µV during hand opening (RM+HE). This suggests an increase in both flexor and extensor activation during volitional attempts to open the hand or reach.

Discussion

This study illustrates the impact of maximal effort to reach or open the hand on limiting hand opening achieved with FES. These results also suggest that functional arm movement and hand opening can be produced with FES in the presence of partial reaching effort. Combining FES with partial effort allowed some participants to reach further and achieve greater hand opening for both the near and far target locations, as hypothesized. In these experiments, we observed a general trend that decreasing voluntary effort for reach and hand opening enabled FES to have a greater effect on hand opening, although the effect size varied between participants. The results support the hypotheses for a portion of the stroke population, and have implications for the design of neuroprostheses to augment arm and hand function in stroke survivors.

The residuals are statistically different from a normal distribution, violating one of the assumptions of the ANOVA test. As a result, p-values could be biased
to lower values. Considering that $p$-values for statistically significant hand opening are less than 0.001 and that removing outliers produced the same conclusions, it is unlikely that the change in $p$-values would be large enough to change the interpretation.

**Implications for Neuroprosthetic Restoration of Arm and Hand Function**

The differences in the interaction between stimulation and voluntary effort across different individuals suggests multiple approaches for implementing an arm and hand neuroprosthesis for stroke survivors. For individuals in whom increased voluntary reach or hand opening effort decreases stimulated hand opening[16, 52, 57], it is important to limit effort exerted during stimulation. It is important to note that partial voluntary reaching effort and even maximum attempted hand opening effort in some individuals did not completely overpower the stimulated hand opening response, and allowed functionally relevant levels of hand opening.

In a neuroprosthetic application, partial effort would produce arm and hand EMGs that could be useful as command signals to control stimulation without completely limiting the beneficial effects of reduced effort. Individuals that do not show a strong impact of effort on the stimulated response will require less reduction of voluntary effort, generating a larger EMG command signal. Thus, the results suggest a feasible approach to restoring functional arm and hand function through integrated voluntary and stimulated muscle activation. The combination of proximal and distal stimulation, with partial voluntary effort, will potentially
restore function to a much broader group of individuals than previously described approaches that only stimulated hand opening[52, 57, 58]. Limiting voluntary effort may have an additional benefit of decreasing the perceived level of effort, encouraging arm use, and potentially reversing learned disuse.

In addition to increasing function, a neuroprosthesis could enable people to participate in therapies that require a prescribed level of voluntary ability[36, 109, 118, 119]. If sufficient motor relearning is obtained from combined therapy and FES and persists after the FES is turned off, it may be possible to discontinue use of the neuroprosthesis. However, if insufficient motor relearning is achieved, continuing use of a neuroprosthesis could provide substantial benefit indefinitely. FES during and following therapy is most likely of value in cases of moderate to severe chronic impairment, where therapeutic interventions to date have been less effective.

Mechanisms Underlying Reduced Hand Opening

Co-activation and co-contraction of finger flexors are likely contributors to the reduction of stimulated hand opening achieved during effort to reach or to open the hand. Results shown here (Fig. 4) and by others[12, 20] show that voluntary shoulder abduction and reach effort increases wrist and hand flexion force, requiring additional extension force to open the hand. Finger flexor and extensor co-contraction in response to effort could reduce the effect of hand stimulation even if it does not change the net grasp force. Voluntarily contracting a stimulated muscle reduces the incremental force added by stimulation in both
able-bodied[95] and stroke patients[116]. If electrically stimulated and volitionally activated motor units overlap, then for stimulation to increase muscle force, either inactive motor units must be recruited by stimulation or volitionally activated motor units must be stimulated at a higher frequency[103].

Reflexes might also limit hand opening. Reflex responses increase after stroke[22, 120]. Finger extensor stimulation can elicit finger flexor stretch reflexes[57] in individuals with stroke. In addition, electrical stimulation at a proximal joint can increase these stretch reflexes[121], but the effects vary by muscle. Stimulation pulse width and stimulus location, nerve trunk versus muscle belly, can vary the neural contributions to force from central and peripheral pathways[122]. Our stimulation parameters are similar to parameters that do not primarily recruit motor units through reflex pathways[120], suggesting that peripheral pathways are primarily responsible. In addition to stimulation producing and modulating reflexes, voluntary effort modulates reflexes as well. Poststroke effort generally decreases the stretch reflex threshold[26] and increases the size of the reflex response[28]. Stroke reflexes are less sensitive than controls to change in voluntary effort[22]. The overall reflex contribution to the change in hand opening is unclear because of modulation from both voluntary effort and stimulation.

Over time, FES can modulate sensory and motor responses in the sensorimotor cortex[123]. The duration of stimulation in our experiments was less than that used to produce a lasting effect in other studies[123], but the sensory
response to FES could affect volitional cortical activation in addition to producing a reflex response. The limitations with hand FES systems for stroke[52, 53] and experiments evaluating simultaneous voluntary effort and stimulation[20, 57, 95, 116] suggest that the variability in the response to FES is not simply a result of altered cortical activation.

Study Limitations and Future Work

The results provide estimates for mean effect sizes and variability due to effort for reach and hand opening. The effect sizes will help determine sample sizes for future studies that examine these interactions in further depth. While the sample size was small, the relative variability in participants’ stimulation responses during effort suggests that patient impairment and lesion location should be incorporated into the model in a larger study. Accounting for these differences may help determine potential sources of variability in this heterogeneous population. There is additional value in evaluating the effect of both sequential and simultaneous effort during static conditions so that force can be used as an estimate for voluntary effort. Isometric force measurements would also allow better understanding about effort’s effect on the stimulation response since the force-movement relationship is highly nonlinear.

Surface stimulation of shoulder muscles for abduction has significant limitations. The goal was to stimulate the axillary and radial nerves to activate the deltoids and triceps without producing discomfort or activating distal muscles. Stimulation lifted the arm above the lap without mechanical support in three
participants. In the future, it is important to stimulate additional muscles to confirm that functional reach can be generated without mechanical assistance in a large population of participants. Percutaneous or nerve cuff stimulation could stimulate deeper shoulder muscles and increase selective muscle activation while reducing discomfort. Since activating and measuring hand opening was more critical to the goals of this study than recording EMGs, the placement of stimulation electrodes took precedence over EMG electrode placement. Therefore, the EMG recordings should not be viewed as highly selective, but as containing information from multiple wrist and finger extensors.

The amount of mechanical arm support varied between the participants (Table 2.2) and was set on the basis of functional reach distance, rather than as a percentage of arm weight, as employed by Miller [12]. This variation in support level would impact the voluntary shoulder abduction activation needed, and contribute to between participant arm/forearm co-activation variability. It may be possible to reduce this variability by standardizing support to achieve a set reaching distance, since the support required to achieve a specified range of movement varies across subjects[9]. However, it is unlikely that the difference in stimulated hand opening is totally accounted for by differences in arm support considering that S1-S4 had similar moderately high levels of support with variable reductions in hand opening, while S5 had minimal support during reach but only a moderate reduction in stimulated hand opening.
The hand sensor used in these experiments provides information about average hand opening and finger grasp force, but there could be variations in individual fingers that aren’t measured. These experiments should be repeated while independently measuring the digits to confirm that functional tasks can be completed using combined voluntary effort and FES. A functionally useful average amount of hand opening may not be useful when the hand posture is unconstrained.

**Conclusions**

These data support our hypotheses that reducing voluntary effort during reach and hand opening and augmenting the effort with stimulation for reach and hand opening can elicit greater hand opening. It is important to note that the effect of voluntary effort during reach and hand opening on the stimulation response varies between participants, even with similar levels of impairment. To produce effective stimulated hand opening, one must be aware of the effect that voluntary effort has on the stimulation response. While partial voluntary reaching effort can have a limiting effect on the stimulation response, partial effort does not completely overpower it. Some reach and hand opening effort could be used in conjunction with hand opening stimulation, and still produce functional hand opening.
Acknowledgements

The authors thank Peggy Maloney for arranging participants’ visits; Steve Sidik, PhD, for statistical consultation; and Mary Harley, Terri Hisel, Amy Friedl, and Kristine Hansen for performing the assessments.

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Chapter 3 – Interaction of post-stroke voluntary effort and functional neuromuscular electrical stimulation

Abstract: Functional Electrical Stimulation (FES) may be able to augment functional arm and hand movement after stroke. Post-stroke neuroprostheses that incorporate voluntary effort and FES to produce the desired movement need to consider how the forces generated by voluntary effort and FES combine together, even in the same muscle, in order to provide an appropriate level of stimulation to elicit the desired assistive force. The goal of this study was to determine if the force produced by voluntary effort and FES add together independently of effort, or if the increment in force is dependent on the level of voluntary effort. Isometric force matching tasks were performed under different combinations of voluntary effort and electrical stimulation. Participants reached a steady level of force and while attempting to maintain a constant effort level, FES was applied to augment the force. Results indicate that the increment in force produced by FES decreases as the level of initial voluntary effort increases. Potential mechanisms causing the change in force output are proposed, but the relative contribution of each mechanism is unknown.
Introduction

Stroke is a leading cause of disability in the US. Six months after their stroke, 50% of ischemic stroke survivors over the age of 64 still have a degree of upper limb hemiparesis [110] that limits arm and hand function, making bimanual tasks difficult if not impossible. Hemiparesis is worsened by disuse and co-contraction patterns across multiple joints (i.e. synergy patterns) [124]. These synergy patterns have been well quantified [11, 111] and appear to be expressed in proportion to effort [9, 10]. Greater effort to abduct the arm increases the involuntary flexor contractions that oppose the desired movement.

Functional Electrical Stimulation (FES) of paretic muscles has the potential to elicit functional limb movements [112], such as reaching and hand opening [20, 71, 112]. For example, electrical stimulation of finger extensors in stroke [51, 52, 54, 113, 114] can produce hand opening while the participant is relaxed. However, when the user exerts effort to open the hand during stimulation, the hand does not open as much as when the person remains relaxed, presumably because their effort to open the hand produces involuntary finger flexor contractions [16, 52, 125]. Therefore in order to receive maximum hand opening from the stimulation, the user must remain relaxed, which is unnatural and runs counter to motor rehabilitation principles, which encourage active attempts to produce functional movement. Our long-term objective is to develop an upper limb FES system that the stroke survivor controls with residual EMG signals recorded from their affected upper limb. Thus, the control strategy
will require the user to exert *some* effort to produce desired arm and hand movements.

In order to develop an effective neuroprosthesis that integrates stimulation and voluntary effort in an efficient and natural way, we need an understanding of their interaction. Previous studies have reported that exerting effort reduces the effectiveness of stimulation both across joints [20], and within the same joint [95]. As a result of changes to central commands and reflexes after stroke, there could be differences between how stroke and able bodied participants' volitional and electrically stimulated forces interact. The quantitative relationship between the level of voluntary effort and the effect of electrical stimulation in the same direction in stroke has not been quantified. It may be possible that small or moderate amounts of effort do not completely limit the effects of stimulation.

The goal of this study was to determine how voluntary effort and electrical stimulation of elbow extensors add together to produce force after stroke (i.e. do the two forces add linearly?). To quantify the relationship between voluntary effort and the effect of electrical stimulation, we stimulated the elbow extensor muscle (triceps) of stroke patients and measured the isometric force that was produced while the participants were asked to exert various levels of simultaneous voluntary effort to push forward, extending the elbow. There are two hypotheses for this study. The first hypothesis is that electrical stimulation of elbow extensors produces greater forward force than voluntary effort alone even in the presence of simultaneous voluntary effort. The second hypothesis is that
as the level of voluntary effort increases, the amount of force added by stimulation decreases.

**Methods**

*Participants*

Six people who suffered a single stroke were enrolled in this study (Table 3.1). Participants were recruited from an outpatient stroke clinic. The primary inclusion criteria included 1) being at least 6 months post-stroke, 2) the ability to reach at least 20% of their full passive reach starting from the closest their hand can passively sit in front of the shoulder while an investigator manually supported the elbow and wrist at 90 degrees of shoulder abduction, and 3) the ability to follow 3-stage commands. Exclusion criteria included 1) shoulder pain, 2) uncompensated spatial neglect, 3) apraxia, 4) insensate chest, arm, or forearm, and 5) impaired cognition or communication. One participant was excluded from data analysis due to great difficulty completing the tasks involved in this study. All participants provided informed consent in accordance with the Declaration of Helsinki prior to participation in this study, which was approved by an Institutional Review Board.
### Table 3.1: Participant Demographics.

<table>
<thead>
<tr>
<th></th>
<th>Age (yr)</th>
<th>Gender</th>
<th>Affected Side</th>
<th>Dominant Side</th>
<th>Time Since Stroke (mo)</th>
<th>FMA Score (arm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>S1</td>
<td>61</td>
<td>M</td>
<td>R</td>
<td>R</td>
<td>22</td>
<td>17 (16)</td>
</tr>
<tr>
<td>S2</td>
<td>53</td>
<td>M</td>
<td>L</td>
<td>R</td>
<td>11</td>
<td>48 (26)</td>
</tr>
<tr>
<td>S3</td>
<td>43</td>
<td>F</td>
<td>R</td>
<td>R</td>
<td>10</td>
<td>26 (21)</td>
</tr>
<tr>
<td>S4</td>
<td>57</td>
<td>M</td>
<td>R</td>
<td>R</td>
<td>12</td>
<td>37 (23)</td>
</tr>
<tr>
<td>S5</td>
<td>79</td>
<td>M</td>
<td>R</td>
<td>R</td>
<td>25</td>
<td>25 (20)</td>
</tr>
<tr>
<td>S6*</td>
<td>48</td>
<td>M</td>
<td>R</td>
<td>R</td>
<td>56</td>
<td>18 (16)</td>
</tr>
</tbody>
</table>

* Shown as total FMA score (66 points maximum), with shoulder/elbow/forearm score in parentheses (34 points maximum)

* Excluded from data analysis

F = female, FMA = Fugl-Meyer Motor Assessment, L = left, M = male, R = right
Setup

Participants performed a series of isometric upper extremity forward force generation tasks (described below) while seated with their arm in two different standardized positions (near or far). Each participant sat in front of a computer monitor with his or her trunk restrained as shown in Figure 3.1. The reference frame used for force directions is shown in Figure 3.1a.

Figure 3.1(a,b): 1a – General setup of participant’s arm orientation: x, y, and z axes show coordinate frame for measurements and θy and θz indicate directions of rotations for hypothetical target placement, as indicated by darker arrows. 1b - Setup showing near arm location and relative equipment locations.
The arm was positioned at 90° shoulder abduction with the shoulder flexed and elbow extended to bring the wrist directly anterior to the shoulder at a distance from the shoulder of either 50% (near) or 75% (far) of the maximum passive reach. Different positions were compared to determine if arm posture affected the interaction. A fiberglass cast over the wrist and forearm connected the wrist to a 2-degree-of-freedom (DOF) gimbal that was attached to a 6-DOF force transducer (JR3 Inc. Model 30E15A-U560A). The forearm was supported by an elevating mobile arm support (Jaeco JME) in addition to the gimbal in order to reduce pressure at the proximal edge of the cast due to the weight of the arm. Less than 0.059 N of horizontal force was required to move the arm support in the horizontal plane. For the participant with the smallest stimulated force, the force required to move the arm support was less than 0.9% of their stimulated force. The elevating mobile arm support used rubber bands to apply a vertical passive force. The rubber bands providing the support are highly compliant, and the stiffness of the device is less than 9 N/m. We measured the maximum vertical elbow movement during all of the trials to determine the maximum force transmitted through the support. The participant whose movement could have the largest effect on their normalized force had a maximum vertical elbow translation of 0.024 m during a single trial. The maximum active vertical force that could have been transmitted through the support during an entire trial was 3.1% of the stimulated force magnitude. The force transmitted through the support would have been less in the rest of the trials. While less than 3.1% of the stimulated force magnitude was transmitted through the support, the majority of the active
vertical forces were recorded by the transducer. Isometric endpoint forces
generated by pushing forces were low pass filtered at 5 Hz, sampled at 60 Hz,
and then down sampled to 12 Hz.

Surface stimulation electrodes (0.038m x 0.089m) were placed over the
triceps to generate elbow extension torques. The stimulation electrodes were
approximately placed over the long and lateral heads of the triceps. It is difficult
to estimate the proportional contributions of the different heads of the triceps to
the stimulated force because the primary target was simply the radial nerve.
Current pulses were delivered through a custom computer controlled stimulator
with a pulse frequency of 35 Hz, pulse amplitude of 40-60mA, and pulse width
that could be modulated from 0 µs to 255 µs. The pulse amplitude and pulse
width were set for each participant to levels that produced the desired magnitude
(see below) of isometric force without pain when the participant was relaxed with
his or her upper arm fully supported. The stimulator ramped the pulse width from
0 µs to the preset level in 0.5 seconds corresponding to cues during the tasks,
and remained on for 3 seconds.

Position tracking markers (LED clusters) were placed on the trunk, upper
arm, and forearm to record limb movement relative to the trunk with an optical
tracking system (Northern Digital, Inc.) in order to confirm that the participant did
not move during the trials.
Experimental Procedures

Prior to any isometric force task sessions, an occupational therapist performed a Fugl-Meyer Motor Assessment [126] to characterize the degree of upper limb motor impairment. Participants then returned to the lab for 2 to 4 sessions to learn the force generation tasks and become accustomed to the sensation of surface electrical stimulation. During two final sessions, force and kinematic data were collected for analysis.

Participants performed upper extremity isometric forward force generation tasks that included various combinations of two factors: voluntary effort and electrical stimulation. Before beginning the force generation tasks, participants were shown how to interpret force magnitude and direction feedback on a computer monitor; this was repeated at the beginning of every session. For each force generation task, the participants were presented a target force direction and magnitude. The target direction was the direction closest to a line extending directly anterior from the shoulder in which the participant could consistently generate voluntary force. The target magnitude was a percentage of the maximum voluntary force that participants could maintain in the target direction. Participants were given verbal encouragement to maintain the target force magnitude in a direction as close to anterior as possible. During these tasks, the percentage of the maximum voluntary endpoint force served as an estimate of effort level. Zero effort was treated as the participant being relaxed and not actively exerting any force. Maximum effort (100%) was considered to be when
the maximum force was exerted. Measuring endpoint force provided a quantifiable estimate of the participant’s effort during the tasks.

Due to difficulty generating a completely anterior force, a target direction for voluntary force was chosen for each participant in a direction they could consistently generate force. We measured the maximum endpoint force that each participant could generate in the target direction. The participant was instructed to reach a target force and try to maintain that force level. Both the force generated and target force were displayed on the monitor. The magnitude of the maximum force was determined by starting with a level that was initially well within the participants’ achievable range. The magnitude of the target was then incrementally raised in successive trials until participants could no longer generate sufficient force to reach the target. The investigator verbally encouraged participants to generate as much force as possible while determining their maximum force. The maximum magnitude force that participants could maintain in the target direction for 1.5 seconds was defined as their maximum endpoint force, which was used to determine the target force magnitude for each participant.

During the tasks, tolerances on the target force allowed participants to stay within the target without maintaining an exact force. The vertical force tolerance in either direction was 5% of the force being generated in the target direction. The horizontal force tolerance from the target was 25% of the magnitude of the target force. Participants who could generate force more
consistently had smaller tolerances on their targets. The reason for a 3-dimensional force matching task was to elicit consistent voluntary force levels by allowing proportional co-contraction levels. While it has been shown that during maximum voluntary contractions stroke victims generate secondary forces in constrained patterns [127]; at sub-maximal effort levels people can generate movement/force outside of these patterns [9, 35].

The following force matching tasks were repeated during both the practice and data collection sessions. Each trial type was repeated four to eight times per session depending on the participant. Table 3.2 shows each participant's average rotations for the isometric force target placement around the y and z axes shown in Figure 3.1a, range of repetitions of the number of trials, and the total average error that participants had during the 1.5 seconds prior to removing feedback and adding stimulation.

Table 3.2: Participants’ average preferred directions for voluntary force generation, range of repetitions for various tasks and each average error during the hold prior to removing feedback and adding stimulation

<table>
<thead>
<tr>
<th>Preferred direction rotations (qz/qy)</th>
<th>Repetitions</th>
<th>Error</th>
</tr>
</thead>
<tbody>
<tr>
<td>S1 3°/17°</td>
<td>4-5</td>
<td>17%</td>
</tr>
<tr>
<td>S2 14°/25°</td>
<td>5-7</td>
<td>19%</td>
</tr>
<tr>
<td>S3 48°/4°</td>
<td>6-7</td>
<td>16%</td>
</tr>
<tr>
<td>S4 8°/14°</td>
<td>4-7</td>
<td>19%</td>
</tr>
<tr>
<td>S5 24°/12°</td>
<td>5-8</td>
<td>16%</td>
</tr>
</tbody>
</table>
Voluntary Effort Maintenance Task: Participants exerted effort to match a force magnitude that was 20% or 50% of their maximum voluntary force (vFmax) in the target force direction. Once they reached the target magnitude and maintained a steady force for 1.5s, visual force feedback was removed while the participant was instructed to continue exerting the same level of effort.

Stimulation Force Task: Triceps was stimulated to generate a force magnitude equal to 20% of the anterior component of the participant’s vFmax in the target direction. The participant was instructed to remain relaxed and visual force feedback was not provided. A 0.5s linear ramp increase in pulse width began when the stimulation was turned on. Stimulation stayed on for at least three seconds.

Combined Voluntary Effort and Stimulation Force Task: These tasks started in a similar manner to the voluntary effort maintenance task, with the participant generating either 20% or 50% of vFmax. When the participant had maintained a steady force for 1.5s, visual force feedback was removed and they were instructed to “keep pushing in the same way you’re pushing, even when the stimulation comes on” in order to have them maintain the same level of effort during stimulation, similar to the ‘do not intervene’ instructions in [22, 128, 129]. The same stimulation parameters used in the Stimulation Force Task were applied. An example of this combined effort and stimulation force trial is shown in Figure 3.2.
Figure 3.2: Example of combined low voluntary effort and stimulation trial performed by subject 1.

Data Analysis

For each participant in each position, the change in force in the stimulated direction after feedback removal and/or stimulation onset was calculated. Separately, for each participant and position, the stimulation alone trials where the participant remained relaxed were all ensemble averaged from the time when the stimulation ramp was started to two seconds after that point, i.e. 1.5 s after the stimulation reached its plateau in order to determine the change in force in the direction of the stimulation. The maximum force magnitude generated during that two second window was considered the maximum stimulated force. Then, using that stimulation alone ensemble averaged dataset, two time points were found. The first time point was when the stimulation alone force reached 5% and the second was when the stimulation alone force reached 90% of the maximum stimulated force. Using these two time points, the change in force in the direction of the triceps stimulation alone was calculated with respect to when feedback was removed and stimulation was turned on. The change in force was calculated for all three conditions as the difference between the forces at the points where
the stimulated force was at 5% and 90% of its maximum stimulated force. To reduce variance in each trial from changes in voluntary effort while feedback was removed and the arm was being stimulated, the force at 90% was averaged together with the forces from 85%-95%. To compare force changes caused by stimulation across participants and arm positions, the changes in force were normalized by the maximum voluntary force that each participant volitionally generated in the direction of stimulation at each arm position.

The normalized force increments generated by the stimulation were compared across different levels of initial voluntary effort to assess whether the dependence on initial voluntary effort was statistically significant (p<0.05). An ANOVA was used to compare the change in force using initial effort level (none, low, and high) and position (near and far) as factors while blocking for participant. If values were statistically significantly different, the analysis was repeated while including participants as a fixed factor. Then the Tukey-Kramer comparison of means was used to determine statistical difference between force increments for separate factors. To evaluate the effects of the change in force during the voluntary force maintenance task’s effect on the combined task, the average change in force during the voluntary force maintenance task using the time points described above was subtracted from each of the combined voluntary effort and stimulation force tasks. The analysis described above was repeated comparing the stimulation alone values to the combined effort and stimulation values minus the average of the change during the voluntary force maintenance task.
Results

At all three levels of voluntary effort, the addition of stimulation increased the force, as illustrated by the superimposed individual trials from one participant shown in Figure 3.3a. However, the force increment caused by stimulation decreased as the voluntary force (effort) increased, as shown by the superimposed incremental force responses in Figure 3.3b.

![Figure 3.3(a,b): Examples from participant 5 of stimulation response while he maintains different levels of voluntary force (effort). 3a Total force in stimulated direction before and after onset of stimulation. 3b Change in force from time point when feedback was removed and stimulation ramp was started. Solid lines are trials where participant is relaxed and exerting no effort. Dotted lines represent low (20%) voluntary effort during the onset of stimulation. Dashed lines represent moderate (50%) voluntary effort during the same stimulation.](image)

To compare force changes caused by stimulation across participants and arm positions, the changes in force from stimulation were normalized by the component of the maximum voluntary force that is in the direction of stimulation for each participant and arm position. For all five subjects at both arm positions, the combined stimulation and voluntary force increased with increasing voluntary
force (effort), as shown in Figures 3.4a and 3.4c. However, the size of the force increment after the onset of stimulation decreases with increasing voluntary force (effort) in all cases, except for one: subject 3 at the near position, as shown in Figures 3.4b and 3.4d.

Using the previously described statistical model that blocks for participant and includes initial effort level and position as factors, the response to stimulation was evaluated. Main effects were effort level, position, and participant while...
interaction effects included two way combinations of all three main effects. All of the interaction effects were statistically significant (participants and effort $p<0.001$, participant and position $p<0.001$, effort and position $p=0.029$). The main effect of effort was significant ($p=0.002$) while participant and position were not statistically significant ($p=0.186$ and $p=0.094$). The sum of squares from the different initial effort levels is 0.41 while the sum of squares for the interaction of participant and initial effort is 0.11. R squared was 87.3%.

Data averaged across participants is shown in Figure 3.5. Despite changes in $p$ values, the same effects were significant when the change in force during the voluntary effort maintenance task was subtracted from the combined force increment. Post-hoc analyses provided more insight into which positions/effort levels were statistically significant for individual participants ($p<0.05$). In the near position, participant 3 was not statistically significantly different from stimulation alone at low or high effort and participant 4 was not significantly different at low effort as shown in Figure 3.4b. Subjects 2, and 5 were statistically different in all positions and at all effort levels. Participant 1 was not statistically significantly different in the far position at low or high effort, though the data exhibited a similar downward trend and an ANOVA applied to that set of data revealed significant differences at the two effort levels. Table 3.3 shows that in seven of the ten instances, the increment in force during stimulation alone was significantly greater than during the combination of low effort and stimulation ($p<0.05$) with a mean difference between the force...
increments for the two tasks of -18.0%. The force increment in six of those seven cases was still significantly greater when the change in force during the voluntary effort maintenance task was subtracted. At moderate (50%) effort, eight of the stimulation alone force increments were significantly greater than combined effort and stimulation force increments (p<0.05) with a mean difference of -44.9%. All eight of the force increments were still significantly greater when the change in force from the voluntary effort maintenance was subtracted from the combination trial.

![Figure 3.5: Averaged stimulation response and standard deviations across participants.](image)

The dependence of the normalized force increment on the voluntary component of the force (effort) is described well by the equation \((F_t-F_v)/F_s=(1+a^*F_{vn})\) where \(F_t\) represents the total force after the onset of stimulation, \(F_v\) represents the voluntary force prior to the onset of stimulation, \(F_s\) is the force due to stimulation alone, \(F_{vn}\) is the normalized voluntary force.
(magnitude of the voluntary force normalized by the maximum voluntary force $(F_{vn}=F_v/vF_{max}$)) and $a$ is the slope parameter characterizing how the force increment changes with voluntary effort. This equation is similar to the model used by Perumal et al. [95], except with the assumption that the change in force is linear over the range measured. The equation was fit to each participant’s data separately, forcing the line through 1 on the vertical axis. The average value for the slope $a$ is -0.23 (s.d. = 0.13). The mean difference in slope ($a$) between the near and far positions was 0.04, but a paired t-test did not indicate statistical significance ($p=0.63$).

Table 3.3: Summarized results showing number of instances where change in force in combined voluntary effort and stimulation task is different than change in force during stimulation alone, as well as number of instances when difference between those changes is different than change in maintained voluntary force during voluntary effort alone without either stimulation or feedback. *(p<0.05)*

<table>
<thead>
<tr>
<th>Effort Combination</th>
<th>Change significantly different from stimulation alone*</th>
<th>Combination change as proportion of stimulation alone (%)</th>
<th>Mean combination change difference (%)</th>
<th>Mean voluntary change (no stimulation) (%)</th>
<th>Combination increment minus voluntary maintenance significantly different from stimulation alone*</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low</td>
<td>7/10</td>
<td>82.0</td>
<td>-18.0</td>
<td>-2.8</td>
<td>6/10</td>
</tr>
<tr>
<td>Moderate</td>
<td>8/10</td>
<td>55.1</td>
<td>-44.9</td>
<td>-9.0</td>
<td>8/10</td>
</tr>
</tbody>
</table>

While Figures 3.4b, 3.4d, 3.5 and Table 3.3 show a decrease in the response to stimulation as voluntary effort increases, in all instances the stimulation response was statistically greater than zero ($p<0.05$), indicating that stimulation increased the total force in all cases.

The target force directions among the set of participants covered a range of directions, as shown in Table 3.2. Across the participants, the target force directions followed a trend, with the more medial preferred directions (+z
rotations) having smaller downward components (+y rotations). The directional rotations are with respect to the coordinate frame shown in Figure 3.1a. None of the participants had a target force direction in the same direction as the stimulated triceps force.

While the data set is incomplete for calculating the full set of inverse torque calculations, we can make some general observations about the directions of forces being generated by the elbow and shoulder based on the directions of endpoint forces. The average angle between forces generated by the shoulder and elbow in the near and far positions were 62° and 94° respectively.

Discussion

The above results support the first hypothesis that electrical stimulation can produce greater forward force even in the presence of some levels of voluntary effort. This increase in force should lead to increased movement produced with a neuroprosthesis. However, the results also support the second hypothesis that the force response depends on the effort level of the participant. The results also indicate that there may be individual variations in the relative dependence on voluntary force. On average, the force increment caused by stimulation was reduced by 15% in the presence of 20% voluntary effort and by 36% in the presence of 50% voluntary effort. While the results indicate that the total force is not a simple summation of voluntary and stimulated forces, the differences between individuals implies that a more targeted set of experiments
to specifically determine the mechanisms that result in these changes and what is causing the variations across individuals would be useful.

These results are similar to those reported in able bodied participants. Langzam et al. [102] studied the tibialis anterior, and observed that higher stimulation levels were necessary to produce the same total torque levels when superimposed on higher volitional torques. Their experimental and analysis methods were based on an explicit assumption that the voluntary and stimulated motor neurons did not overlap, as well as the quantitative estimation of voluntary contraction levels from the EMG of the stimulated tibialis anterior, whereas our experimental and analysis methods did not. Perumal et al. [95] observed that stimulation superimposed on voluntary effort decreased the stimulation response with increasing voluntary effort in the quadriceps. They assumed that motor unit recruitment overlap was the only mechanism contributing to the change in force. We studied a different muscle group in an impaired group of participants, and also observed significant variations between participants and similar decreases in force increment. Both the current study and the earlier studies observed less than linear summation of the stimulated and voluntary force responses that was force dependent. The similarity of the conclusions in the perspective of the differences in approach gives increased confidence in the robustness of the effect.

These experiments did not examine the mechanism creating these changes, but there are a few that could be causing this difference. One possible
mechanism contributing to the reduction in incremental force is that some motor units are activated both by voluntary effort and by electrical stimulation (overlap). If the motor units being recruited by the stimulation overlap with the volitionally activated motor units in the same muscle, the increment in force from stimulation will be limited to the additional recruitment of inactive motor units, and the possible increase in excitation frequency of volitionally activated motor units. Even though the brain recruits in order of smallest to largest motor units [130] and it is generally accepted that neuromuscular electrical stimulation has been shown to recruit motor units in the opposite order using direct nerve stimulation [131], the response to surface stimulation has also been observed to be more mixed and unselective [131, 132]. Despite the differences in recruitment order, rate modulation continues after full recruitment with the result that full voluntary recruitment can occur well before maximal force is achieved [133-135]. This interpretation is consistent with the observation (Figures 3.4b and 3.4d) that the reduction in the incremental force is larger at higher effort levels, since we would expect both greater overlap of recruitment, and higher motor unit firing rates at the higher force level. It has been demonstrated during fatigue tests employing maximal voluntary activation that short trains of supramaximal stimulation do not produce a force increment [136]. Thus, we would not expect an increase in force if either stimulation or voluntary effort alone was fully activating the muscle. If the primary mechanism for the decrease in force increment is motor unit activation overlap, variations in the extent of overlap between the directions of the volitional force and stimulated force, including a lack of overlap in the near position for
participant three, could partially explain the variations in force dependence across participants. In addition, participant three’s preferred voluntary force direction in the near position was the most different from the stimulated direction of all of the combinations of participants and positions. It is possible that overlap was low because the shoulder was contributing significantly to the force output and the elbow was not contributing much extension.

Another possible mechanism is that stimulation elicits reflexes that are modulated by effort. Surface stimulation would be expected to excite both cutaneous and proprioceptive afferents, and there are widespread reflexes in the arm following stroke that are augmented by effort [22, 26, 28]. Reflexes could either enhance or reduce the total force. Since the changes in force increment are not significantly different (p<0.05) between the near and far arm positions based on the slope of the model, this indicates that a position dependent reflex effect is not significantly contributing to the difference in force increments. It should be noted though that this study is not powered to account for ß error, so while we cannot say that the force increments for the two positions are significantly different we also cannot confidently say that they are the same.

Another explanation for the force increment caused by stimulation decreasing with increasing effort is a failure to maintain voluntary effort in the presence of both stimulation and a loss of feedback. However, our analysis above indicates that failure to maintain effort due to loss of feedback is unlikely to be the primary explanation. While there was a decrease in the voluntary force
generated when feedback was removed, the difference between the changes during stimulation alone and combined voluntary effort and stimulation was significantly greater than the change in force when feedback was removed during voluntary effort alone. This supports the hypothesis that the changes are a result of more than just the feedback removal during the task. There could also be a change in voluntary effort as a result of participants feeling and responding to the sensation of electrical stimulation. Participants participate in multiple training sessions before the test sessions to allow them to become comfortable with the sensation of electrical stimulation, however the current study cannot statistically verify that sensation is not a mechanism for the decrease in force increment.

The variance in combined force reduction across subjects could partially be a reflection of the variance in synergy pattern expression. The maximum net force is partially limited by antagonist muscle co-contraction in the synergy patterns and participants potentially being unable to generate maximal contractions post-stroke. Both of these would prevent knowing the maximum forces produced by individual muscles. The target forces and stimulated forces are scaled to the maximum net force, not the actual maximum individual muscle force. Thus, variation of the intensity of synergies across subjects could lead to variation of effects. Similarly, if motor unit overlap is the mechanism responsible for the reduction, variation in recruitment and rate modulation patterns across subjects [137-139] could lead to variability in the amount of voluntary effort that is replaced by electrical stimulation.
The decrease in stimulation response with increasing voluntary effort is significant statistically for effort and the interaction between participants and effort, with a weaker dependence on the interaction. The sum of the squares of the effort is 0.41 as compared to 0.11 for the interaction between effort and participant indicating that effort is generally the stronger contributor. In future studies, percutaneous stimulation, measurement of EMG levels across muscles and an experimental design that simplifies force analysis could provide insight into the mechanisms responsible.

The experimental design could be improved by standardizing the arm position/orientation by joint angle rather than end point position. This would allow choosing a consistent arm orientation across subjects making it easier to back calculate joint torques. We cannot rule out contributions of shoulder torques to the endpoint force. Calculating the joint torques would enable assessments of the force contributions of different joints, similar to the method used by Keller et al. [20]. The current experimental design could not distinguish between forces generated by shoulder and elbow torques as previously mentioned in the discussion. Inverse joint torque calculations are sensitive to small changes in limb configuration, and we did not measure the horizontal arm orientation with adequate precision. As described in the setup, the small forces transmitted through the arm support would also affect the calculated joint torques, but the impact of these forces in our experimental setup is less than the impact of errors in the shoulder/elbow positions.
Similarly, it would be ideal to have the same voluntary force target direction for each participant. Targeting joints that have fewer muscles crossing them would further simplify the interpretation of the data. Lastly, able bodied trials incorporating these design changes would help establish a baseline before evaluating these summations in groups who have altered reflexes and central inputs. These changes would provide greater understanding of the mechanisms, which is important in the future design of neuroprosthesis to exploit voluntary force augmentation.

The relationship between the response to stimulation and the underlying level of voluntary effort has potential ramifications for the design of stroke neuroprostheses that integrate a user's voluntary effort with electrical stimulation. While there is a decrease in the force increment as voluntary effort increases, the force increment is still substantial (64%) at moderate effort for a stimulation level that produces 20% vFmax when exerting no effort. Partial effort does not completely block the stimulation response indicating that voluntary effort can be used as a command signal and FES can still augment movements as long as effort level is considered while designing post-stroke neuroprosthesis control schemes. Limiting effort to limit the expression of synergy patterns will allow FES to have a greater effect. This type of neuroprosthesis would use EMG recorded at low effort levels as the command signal for large levels of stimulation that would be the primary movement generators. The user's focus would be to generate suitable EMG levels for a command signal rather than attempting to
generate maximum effort. As observed in the results from Keller et al. [20], shoulder abduction generates elbow flexion that can be difficult to overcome by elbow extensor stimulation. Our preliminary studies support the hypothesis that reducing voluntary effort during reach and hand opening and supplementing that effort with stimulation can generate hand opening even during reach [140]. By generating most of the shoulder abduction with stimulation instead of voluntary effort, elbow and hand stimulation may be able to have a greater net effect.

One approach in designing neuroprostheses for stroke is to rely as much as possible on the user’s residual voluntary ability to move and only add stimulation to supplement the voluntary effort as needed. This approach minimizes the extent of intervention (number of channels, intensity of stimulation) and maximizes the potential therapeutic benefit of requiring greater voluntary control. However, there are potential benefits to reducing voluntary effort and increasing the use of stimulation in a neural prosthesis. First, lower voluntary effort reduces the intensity of synergistic contraction patterns [10]. Synergy patterns scale somewhat proportionally to effort, increasing the forces in multiple muscles [9]. In order to decrease the undesired synergy response and maximize the stimulation response, it is beneficial to reduce the effort exerted on the part of the user. Thus, less stimulation force would be required to overcome undesired contractions of antagonists. For example, relaxation of the arm allows FES to open the hand by reducing synergistic finger flexor activation. While FES has been hypothesized to reduce antagonist contractions by reciprocal inhibition,
experimental evidence does not support that hypothesis [125]. Secondly, there is no loss of maximal force potential by limiting voluntary effort, since lower voluntary effort increases the forces that can be elicited by FES in the same muscle. Maximal stimulation and maximal voluntary activation produce the same force [136]. Third, lower effort reduces the intensity of stretch reflexes, which slow down and limit the extent of voluntary movement in stroke [22, 28].

If the mechanism of reduction in the stimulation response as a result of increased effort is a result of overlap between activated motor units, this has another implication for the design of neuroprostheses for stroke. For systems which use EMG from stimulated muscles as part of the control signal [86], one must be aware that some of the asynchronous action potentials that were activated by the central nervous system will be replaced by synchronous action potentials activated by stimulation. Even if the synchronous action potentials (M-waves) are removed from the signal [86, 141-143], the residual EMG is not entirely indicative of the user’s effort in that muscle.

FES, as a functional neuroprosthesis, has the potential to increase function by improving reach and hand opening [140] augmenting the gains in function that are achieved by physical rehabilitation and robotic therapies. Neuroprosthetic and therapeutic approaches are not mutually exclusive approaches to increasing either reach [36] or hand function [144] after stroke, and combined approaches may provide the best results. Assistive forces from an FES system may enable stroke survivors to utilize some of these therapies that
they were previously unable to participate in. Similarly, participating in these therapies may increase volitional movement and disconnect synergy patterns [36], thereby allowing more assistance from volitional effort, providing a more robust command signal, allowing for finer movements in response to stimulation, and progressively decreasing a reliance on the stimulation.

Future stroke neuroprostheses should be designed with an understanding of the relationship between effort, stimulation level, and motor output. Doing so can allow the user to derive optimal benefit from the device. These results indicate that even in the presence of moderate voluntary effort, FES can increase post-stroke force production, decreasing the impairment on the affected side.

Conclusions

Electrical stimulation is capable of increasing endpoint force even in the presence of voluntary effort post-stroke. The stimulation response is dependent on the level of voluntary effort and to a lesser extent on the individual participant, though the contributions of different mechanisms are unknown. This change in force response should be taken into consideration in the design of future post-stroke neuroprostheses.

Acknowledgements

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Chapter 4 – Control of Force Assistance for Poststroke Reach

This chapter will be submitted for publication:
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Introduction

Hemiparesis after stroke severely limits arm and hand function. Retained movements are commonly characterized by involuntary muscle co-activation, which is described as synergy patterns, because patients have difficulty independently controlling joint movements [124]. For example, in response to voluntary shoulder abduction, patients experience involuntary elbow, wrist, and finger flexion and have difficulty producing elbow extension [9, 11, 12]. The synergy patterns have been well documented and appear to be expressed in proportion to effort. While patients can make some independent movements within these patterns, more severely impaired patients have difficulty grading reaching direction [8].

Therapeutic approaches have shown some promise to improve arm function [36, 118, 145], but moderately and severely impaired patients may benefit from additional assistance during everyday tasks. Currently, several research groups are developing powered exoskeletons that could have potential stroke applications as assistive devices for everyday use or therapy [50, 84, 146-149]. In most cases, the assistance is either independent of the user’s attempted movements, or the robot provides assistance to compensate for the user’s inability to fully complete a predefined movement. In some cases, the user triggers the movement with their electromyograms (EMG) [150]. If the assistive
system produces movement through a combination of the user’s voluntary effort and assistance, the system must be low impedance and provide assistive forces. These assistive forces could be controlled in different ways, but there are three basic ways to relate the assistance to the user’s effort. From the perspective of the endpoint (i.e., the hand), the exoskeleton could apply forces in the form of 1) an offset bias, a constant force that is independent of the user's efforts, 2) a gain, an amplification of the user’s volitional reaching force, and/or 3) a rotation, a rotational shift in the direction of the applied force based on the user's activity.

Constant antigravity assistance and lateral forces can increase the horizontal workspace after stroke [9, 82] even when the assistive forces are perpendicular to the direction of movement; antigravity assistance has even been incorporated into therapeutic approaches [36]. In addition to constant forces, assistance can be modulated in different directions based on the user's intentions. EMG has been used to control single degree of freedom assistive torques from both exoskeletons and Functional Electrical Stimulation (FES) systems [84, 86, 105]. Participants were able to control the assistance in the same degree of freedom and used less effort while moving with assistance. While the EMG controlled assistance in these studies was performed about a single degree of freedom, EMG has been used to accurately estimate 3-dimensional position [65, 151], joint angles [61, 66], force [67, 69, 71, 152], and task [75], suggesting that EMG has the potential to control assistance for multiple degrees of freedom arm and hand movements.
An FES system or powered exoskeleton controlled by physiological signals related to residual movement (e.g. EMG) on the affected side may provide functional assistance during activities of daily living as well as therapeutic benefit. Upper extremity FES systems for stroke have primarily focused on restoring hand function [52, 53], but patients may benefit from additional FES or robotic assistance at the shoulder and elbow. In addition to direct assistance, controlled arm support may have an indirect effect of reducing the expression of synergy patterns, since less shoulder abduction effort is required to support the limb. This could enable participants to complete tasks using a combination of their own effort and the system's assistance, thereby encouraging active use of the upper limb as the user gains independence.

While robotic assistance has been shown to increase poststroke range of motion, the effect on quality of reaching movements has not been evaluated. Similarly, it has been shown that able-bodied users can control assistance based on voluntary muscle activity, but these same methods have not been evaluated in the stroke population. The goal of this study was to determine how force assistance impacts the range of motion and controllability of arm motions performed by individuals with stroke. Participants performed a 3-dimensional reaching task with different types of sagittal plane robotic assistance. There were three hypotheses: 1) assistive forces will increase the active workspace, 2) participants can control assisted reaching movements as well as or better than unassisted movements, and 3) participants will exert less effort (EMG) during
assisted reaching movements. This study expands upon previous studies by transitioning from single degree of freedom assistance to 2-dimensional assistance during 3-dimensional movements and evaluating how the type of assistance impacts stroke participant’s movement control.

**Methods**

**Participants**

Participants were recruited from an outpatient stroke clinic. Inclusion criteria were: 1) at least 6 months poststroke, 2) active forward reach of at least 10 centimeters with the wrist and elbow fully supported, 3) follows 3-stage commands, 4) unable to fully reach forward and open the hand simultaneously, 5) sufficient upper extremity passive range of motion to reach to a table and open the hand. All participants provided written informed consent prior to participation in this study, which was approved by an Institutional Review Board. Eight participants were enrolled and completed the study. Participant demographics are in Table 4.1.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Age (yr)</th>
<th>Sex</th>
<th>Side (Affected/Dominant)</th>
<th>Time since stroke (mo)</th>
<th>FMA Score (arm)</th>
<th>Elbow Ashworth (flexors/extensors)</th>
</tr>
</thead>
<tbody>
<tr>
<td>S1</td>
<td>58</td>
<td>M</td>
<td>L/R</td>
<td>22</td>
<td>35 (24)</td>
<td>(1+/1)</td>
</tr>
<tr>
<td>S2</td>
<td>57</td>
<td>F</td>
<td>R/R</td>
<td>9</td>
<td>23 (14)</td>
<td>(2/0)</td>
</tr>
<tr>
<td>S3</td>
<td>61</td>
<td>M</td>
<td>L/R</td>
<td>36</td>
<td>23 (12)</td>
<td>(1/1+)</td>
</tr>
<tr>
<td>S4</td>
<td>79</td>
<td>M</td>
<td>L/R</td>
<td>35</td>
<td>35 (22)</td>
<td>(1+/0)</td>
</tr>
<tr>
<td>S5</td>
<td>66</td>
<td>M</td>
<td>R/R</td>
<td>22</td>
<td>14 (11)</td>
<td>(2/1)</td>
</tr>
<tr>
<td>S6</td>
<td>79</td>
<td>M</td>
<td>R/R</td>
<td>16</td>
<td>30 (16)</td>
<td>(0/0)</td>
</tr>
<tr>
<td>S7</td>
<td>57</td>
<td>M</td>
<td>R/R</td>
<td>48</td>
<td>15 (14)</td>
<td>(2/0)</td>
</tr>
<tr>
<td>S8</td>
<td>66</td>
<td>M</td>
<td>L/R</td>
<td>22</td>
<td>22 (13)</td>
<td>(2/0)</td>
</tr>
</tbody>
</table>

FMA: upper limb portion of Fugl-Meyer Motor Assessment, 66 point maximum, shoulder/elbow/forearm in parentheses, 34 point maximum. Ashworth test: 5 point scale
Setup

Participants completed a series of force generation and reaching tasks while seated in a chair with a seatbelt and shoulder straps that prevented trunk movements from contributing to their overall movement and force generation. The affected hand and forearm were placed in a wrist brace (AirCast) that was attached to a HapticMaster (MOOG), which recorded force and position of the wrist, as shown in Figure 4.1a. In some trials, the HapticMaster applied assistive forces at the wrist. Position and force were recorded from the HapticMaster at 40 Hz. A computer monitor displayed feedback about the endpoint force and position.

Surface electromyograms (EMG) were recorded from the following shoulder and arm muscles: upper trapezius; pectoralis major; serratus anterior; anterior, middle, and posterior heads of the deltoid; biceps; and triceps. EMGs were lowpass filtered at 1000Hz and amplified with a gain of 9900, and subsequently sampled at 2500 Hz. Recorded signals were then bandpass filtered between 40 and 500 Hz, filtered with a real-time ECG removal filter [153], rectified, and lowpass filtered at 3 Hz with a 2\textsuperscript{nd} order Butterworth filter, similar to previously used approaches [85].
Experimental Procedures

Participants spent one to three sessions learning the force generation and reaching tasks and becoming accustomed to the display of endpoint force and position feedback. After the practice sessions, participants completed a testing session. During the testing session, participants performed the reaching task with standardized sets of assistance, described below. One participant (S2) completed a second testing session, in which she performed the reaching tasks with a unique set of assistive forces in an attempt to enable her to reach as much of the workspace as possible while reducing voluntary effort.

Reach and Hold Task

The participants' task was to reach to a target position and maintain the position for 1.5 seconds. All prescribed movements were in the sagittal plane, and positions were measured with respect to a reference position at the height of the xyphoid process and at an anterior position halfway between the xyphoid process and the furthest passive position their wrist could reach at that height.
Participants started with their hand positioned above their lap at the base of the HapticMaster's workspace, directly below the reference point. They were then cued to reach to one of three target positions. An example time course for the task is shown in Figure 4.2(a, b).

Figure 4.2 (a,b,c): (a, b) Example data for S8 reaching to the forward target showing (a) a 3-dimensional path and (b) the time course of position and force assistance with the outcome metrics highlighted. This trial utilized EMG controlled assistance. Abbreviations: S/I – Superior/Inferior, A/P - Anterior/Posterior, M/L – Medial/Lateral. (c) Example range of motion for S2 while reaching with the four reaching conditions. The bottom center circle represents the reference (start) position. The other dots represent the reach targets.
The targets were 4 cm diameter spheres at the following three positions:

1) Forward: 10 cm anterior to the reference position, 2) Near: 10 cm posterior to the reference position, and 3) Up: 10 cm above the reference position. These target positions were within the passive range of reach, and are functionally relevant endpoint positions. The near position is useful for self-care, like buttoning a jacket. The forward position is useful for movements such as placing the hand on a table. The up position extends toward the head and would be useful for eating. During the test session, the reaches were performed using four standardized assistance conditions: 1) No assistance, 2) Offset: constant vertical assistive force equal to half the weight of the arm at the reference position, 3) Gain: dynamic assistance equal to the participant’s estimated effort in the sagittal plane, based on EMG measurements (below), and 4) the sum of constant vertical assistance (same as offset, condition 2) plus effort-based assistance (same as gain, condition 3). EMG measurements were used to assess effort that participants exert.

During the gain condition, participants’ voluntary force was estimated from EMG signals. An additional assistive force equal to the estimated force was applied to the wrist by the HapticMaster to help produce the desired movement. The flow diagram for the assistive force calculation and application to the arm is shown in Figure 4.1b. The reaching task was repeated with the three target locations and four assistance types for a total of twelve combinations. Each combination of target direction and assistance comprised a block of repetitions.
Participants performed one practice trial and then did five repetitions for analysis in each block. The order of the twelve blocks was randomized for every session.

**Voluntary Force Estimation**

In order to estimate voluntary force exerted during the reaching task, we trained a time-delayed artificial neural network (TDANN) that related EMG from shoulder and arm muscles to active forces along each axis in the sagittal plane. Training data for the TDANN included forces measured during isometric force generation and dynamic movements with compliant loads. Force targets increased and decreased in a sinusoidal manner and were displayed on a computer monitor. The target force period depended on how well participants could modulate force and ranged from 6 to 15 seconds. Participants tracked these target forces by pushing and pulling with their arm. The TDANN training set consisted of a total of 15 minutes of data, separated into 30-second force matching blocks. Training data was collected at the beginning of each test session.

During the 30-second blocks, participants attempted to match target forces along the sagittal plane parallel to the anterior/posterior (AP) and superior/inferior (SI) axes while minimizing perpendicular forces along the medial/lateral (ML) axis. Thus, although all the targets were in the sagittal plane, movement was a 3D control task. The task required matching the moving target force within a radius of 2N. The target forces along the AP axis also required
generating a constant positive force in the superior direction to balance the gravitational load.

During the gain condition, the HapticMaster adds the participants’ estimated voluntary force, enabling them to reduce their effort to reach the target. Considering this expected decrease in effort during reaches with the gain condition, the majority of target forces were less than the passive force required to hold the arm at the target locations. Across participants, the average maximum force targets along the AP axis were 61% of the opposing passive force required to hold the arm at the forward location. The maximum target force in the SI direction was equal to the weight of the arm in the reference position, but the majority of target forces in the SI direction were closer to half of the arm weight in the reference position.

During tracking tasks with a compliant load, participants tried to reach along the AP and SI axes to the target positions (near, up, and forward). The stiffness of the compliant load during reach along the AP directions required participants to generate a net force equal to the isometric target forces in the AP direction. Stiffness during reach towards the up target required generating a net force at the up target equal to half the weight of the arm. We expected effort exerted during assistance conditions to have been within the training region since the HapticMaster is supplying force to assist with the movement, but effort during the no assistance condition may have been beyond the majority of the training force targets.
Because stroke patients are not completely limited to the stereotyped synergy patterns [8, 9], and different tasks can be estimated from residual movement [75, 154], we expected participants to generate some force in independent directions despite difficulty generating arm movement. Some participants had difficulty matching the target forces during the collection of the TDANN training data. If participants could not match a target force, they were instructed to prioritize force target tracking in the correct direction, even if they generated smaller or off axis forces.

To evaluate participants’ ability to generate independent forces, we used principal component analysis to separate the direction of the voluntary forces generated while the arm was in the reference position. Across subjects, the mean variance accounted for (VAF) by a single principal component was 71% while the range across subjects was 61%-90%. The average VAF for the first and second components was 97%. These data suggest that participants had some ability to modulate voluntary force direction, but some subjects had difficulty generating independent forces as indicated by the subject with 90% VAF by the first component. If participants could not generate independent forces, a single component would account for 100% of the VAF. Since force generation was a 3-dimensional task, a single component’s minimum possible VAF was 33%. If participants perfectly generated forces within the sagittal plane, a single component’s minimum possible VAF was 50%. The VAF was partially biased by the ranges of target forces since the target force range in the AP and SI
directions were not equal. On average, participants generated force along 94% of the AP target axis (17N range centered around 0N) while maintaining a force in the superior direction.

The direction of the primary and secondary components indicates how well participants generated force within the sagittal plane without generating perpendicular forces in the ML direction. If participants perfectly exerted force along the sagittal plane, the out of plane components would have been zero. If participants only generated forces along the ML direction, the component would have been 1. The average magnitude and range of the ML portion of the primary component direction was 0.09 (0.05-0.15) and for the second component direction was 0.26 (0.03-0.74), suggesting that subjects could partially limit force generation to the sagittal plane, but had some difficulty maintaining force within the sagittal plane. The average magnitude of force in the ML directions across trials and subjects was 1.4N, which was within the 2N target force size. Despite movement impairments, the ability to modulate force direction along multiple axes suggests that residual movement provided sufficient information to generate unique command signals within the sagittal plane, without generating significant movements in the medial-lateral direction.

After recording force data, a TDANN was trained using the 8 EMG channels as inputs and the sagittal components of the force matching task as outputs. The TDANN architecture consisted of a linear input layer, one hidden layer comprised of 4 nodes with tangential-sigmoid activation functions, and a
linear output layer. The TDANN was trained for 600 iterations. Inputs included the rectified and lowpass filtered EMG sampled at the current time point and 64ms prior. The network sampled the processed EMG every 32ms. The initial choice for TDANN structure was based on results from other studies that incorporated upper extremity kinematics and force generation in both able-bodied and impaired populations [61, 66, 68, 154, 155]. Using data from one of S1’s practice sessions, we tested a range of network architectures including the size of the hidden layer and the input time delay to determine a small network size that did not impair force estimation. The same network architecture was used with each of the participants.

It is difficult to evaluate the accuracy of the TDANN during the reach and hold trials, so TDANN accuracy was evaluated offline. Three TDANNs were trained and tested for each participant. 90% of the data were used to train the TDANN while the other 10% of the data were used to evaluate TDANN accuracy. $R^2$ values, calculated to compare the TDANN estimates for the test data and the actual forces generated, were (mean ± SD) 0.52 ± 0.16 and 0.75 ± 0.05 in the AP and SI directions respectively.

Range of Motion

To determine the maximum active workspace, participants performed a maximum range of motion (ROM) task. Participants were instructed to slowly reach in the biggest circle that they could in the sagittal plane. They could see their arm but were not given any other feedback about how far they had reached.
Participants reached in a circle twice with each of the four assistance conditions (none, offset, gain, offset + gain). The average time for a single rotation around the plane was 9s ± 3s (mean ± SD). Using a composite of the ROM task and the reach task, we calculated the outer area of all the reaches. An example of the area encompassed is shown in Figure 4.2c. Reach was partially constrained by the available workspace of the HapticMaster, which had a range of 40cm in the SI directions and 36cm in the AP direction for a cross section of 1440cm².

Participants were seated with the base of the HapticMaster directly above the lap and positioned so the wrist could touch the base of the sternum while it was attached to the HapticMaster. This workspace allowed full range of forward reach and allowed reaching to approximately the top of the head for some participants.

Data Analysis

To assess how assistance changed the reachable workspace, we compared the effects of the assistance condition on the sagittal ROM. Since some of the participants could reach most of the workspace without assistance, we separated the participants into two groups: 1) four participants could reach more than half of the area in the HapticMaster’s sagittal workspace (720cm²) without assistance and 2) four participants could reach less than half of the HapticMaster’s sagittal workspace without assistance. These groups were representative of the participants who were able to reach and hold all of the target positions without assistance and those who could only reach and hold some of the target positions without assistance. We then calculated the
difference in ROM with and without assistance to determine if the assistance conditions increased the ROM. We used an analysis of variance (ANOVA) with assistance condition as a fixed factor to evaluate if assistance increased the active ROM. If the change in area was significant, the individual factors were tested for statistical significance using a Tukey test.

To determine if participants who could not already reach the targets had improved reach, we separated out targets that participants could not already reach without assistance. This prevents a floor effect from participants who could already reach the target. We calculated distance from target during the reach and hold task for those targets that participants could not reach without assistance in the majority of the trials. We compared the distance from the target under the different assistance conditions using an ANOVA, with subject as a random factor and direction and assistance as fixed factors.

In addition to reaching over a larger workspace, it was important that participants could control movements within that larger workspace. To test the hypothesis that performance with assistance was better than performance without assistance, we assessed the three reach and three hold outcome metrics shown in Table 4.2. Some participants could not reach or hold particular targets. So despite attempting all trials, they did not have outcome metrics for some combinations of target and assistance conditions. In order to compare the effect of assistance on movement control required that participants could reach and hold the target position without assistance. Datasets were constructed using the
trials when participants could reach and hold a target location during at least half of the no assistance trials for that target. If a participant could reach and hold a target without assistance, their reach and hold outcome metrics for all of the reaching conditions to that target were included in the dataset used for statistical analysis. If a participant could not reach and hold a particular target location without assistance, none of that data was included in this dataset, even if they were enabled to reach the targets with assistance. Separating the data in this manner avoided confounding the data with reach characteristics for trials when participants could only reach the targets with assistance.

Table 4.2: Reach and Hold outcome metrics

<table>
<thead>
<tr>
<th>Reach Characteristics</th>
<th>Hold Characteristics</th>
</tr>
</thead>
<tbody>
<tr>
<td>Distance from target (cm) – closest distance to the target center during the reach</td>
<td>Time to stabilize (s) – time between when the wrist is first in the target range (2cm radius) and stays within the target range for 1.5s</td>
</tr>
<tr>
<td>Time to target (s) – time between the initial cue and the first time the wrist moves inside the target range (2cm radius)</td>
<td>Average distance (cm) from the target during the ‘Time to stabilize’</td>
</tr>
<tr>
<td>Path length – sum of the path traveled between moving within 95% of the target distance and first reaching the target divided by the straight line distance between those two points</td>
<td>Stability during hold (variance) – the sum of the position variance during the 1.5s hold inside the target</td>
</tr>
</tbody>
</table>

Separate ANOVAs were used to compare each of the metrics using target location and assistance as fixed factors and subject as a random factor. We tested for the interaction between the fixed factors. Additionally, we used a two-sample t-test to compare the reach and hold characteristics between target
locations that participants could only reach or hold with assistance and target locations that participants could reach and hold without assistance.

The effort exerted while holding the target position was assessed from the individual muscle EMGs. EMG magnitudes were averaged over the middle 1s window during the 1.5s hold period. The EMGs for each muscle and each target were normalized to the average magnitude measured for the corresponding muscle and target during voluntary hold without assistance. EMG data were compared with an ANOVA to determine if less effort was necessary to hold the position when assistance was provided. The model incorporated subject as a random factor; and target location and assistance as fixed factors; including the interaction effect for the fixed factors.

The outputs of the TDANN provided estimates of the participants’ voluntary forces during the reach phase of each trial, and thus allow comparisons of the voluntary effort across the different assistance conditions and target locations. Participants varied in how successfully they reached the different targets, so force estimates were calculated differently depending on how well participants reached. Participants were separated into three groups when calculating the estimated voluntary force: 1) able to reach the target, 2) starts moving towards the target, but is unable to reach target, and 3) does not move target when attempting to reach. For trials where participants could reach the target (1), the average forces were calculated over the same time frame as the path length calculation. In trials when participants moved toward the target but
could not reach the target position (2), the estimated force was calculated over a 1.5s time window beginning when the participant reached within 95% of the distance from the target or the closest distance to the target they were able to achieve. Some participants moved away from the target when they attempted to reach (3). In this case, the 1.5s window started when the force estimate in at least one of the directions changed by 4N from the initial force estimate. The force estimates from all of the trials were compared with an ANOVA using the same factors as the other outcome measures.

*Individualized Assistance*

In addition to the standardized set of assistance conditions (none, offset equal to half the arm weight, gain of 2, and offset with a gain) used for all eight participants, we tested a unique assistance condition with S2, who was one of the participants unable to reach targets without assistance. The goal was to increase the range reached while decreasing effort. Feedback from S2 was incorporated into choosing the parameters for the offset and gain assistance.

*Results*

Participants who could not reach all of the targets were enabled to reach a larger portion of the workspace with assistance (p<0.05). In addition, participants could reach either the same number or more targets while exerting less effort (p<0.05) in at least six of the eight muscles measured. However, in a few instances, the assistance reduced the quality of the reach and hold movements (p<0.05).
Range of Motion

The unassisted ROM and the average change in ROM with assistance are provided in Table 4.3. For participants who could reach more than half of the sagittal workspace (720 cm$^2$) without assistance, the assistance conditions did not increase ROM significantly (p=0.25). For the more impaired participants who could not reach more than half the sagittal workspace without assistance, the increase in ROM was significant when using assistance (p=0.04). Post-hoc analyses indicate that only the combined assistance of gain plus offset provided statistically larger ROM than reaching without assistance (p < 0.05).

Table 4.3: Summary of average reaching area with and without reaching assistance for the two groups of participants, those who could reach more than half of the sagittal workspace without assistance and those who could not.

<table>
<thead>
<tr>
<th>Amount of sagittal workspace reached without assistance</th>
<th>Area reached with no assistance Mean±SD (cm$^2$)</th>
<th>Change with Offset Mean±SD (cm$^2$)</th>
<th>Change with Gain Mean±SD (cm$^2$)</th>
<th>Change with Offset and Gain Mean±SD (cm$^2$)</th>
</tr>
</thead>
<tbody>
<tr>
<td>More than 50% (S1, S3, S6) (n=3)*</td>
<td>1177 ± 163</td>
<td>+115 ± 102</td>
<td>-103 ± 196</td>
<td>+82 ± 137</td>
</tr>
<tr>
<td>Less than 50% (S2, S5, S7, S8) (n=4)</td>
<td>276 ± 218</td>
<td>+106 ± 85</td>
<td>+177 ± 154</td>
<td>+258 ± 138</td>
</tr>
</tbody>
</table>

*ROM data not recorded for S4.

The ROM describes the total area encompassed during the range of motion trial, but some discrete target locations were more difficult to reach than others. Table 4.4 shows the percentage of trials in which participants could reach and hold the target position. Without assistance, participants consistently reached the near target, but they had more difficulty reaching the forward and the up targets. Reaching with assistance in the forward and up directions increased
the number of times participants reached and held the target position. The greatest increase was for the combined gain plus offset.

Table 4.4: Percentage of trials that all participants reached and held the target position with and without assistance.

<table>
<thead>
<tr>
<th>40 Trials per direction and assistance combination</th>
<th>No assistance</th>
<th>Gain</th>
<th>Offset</th>
<th>Offset and Gain</th>
</tr>
</thead>
<tbody>
<tr>
<td>Forward</td>
<td>52.5%</td>
<td>62.5%</td>
<td>62.5%</td>
<td>70%</td>
</tr>
<tr>
<td>Up</td>
<td>60%</td>
<td>77.5%</td>
<td>72.5%</td>
<td>87.5%</td>
</tr>
<tr>
<td>Near</td>
<td>95%</td>
<td>92.5%</td>
<td>100%</td>
<td>97.5%</td>
</tr>
</tbody>
</table>

Reach and Hold Outcome Measures

The outcome metrics described in Table 4.2 for the reach and hold components of the task are plotted in Figure 4.3. Table 4.5 contains the $p$-values for the main effects and interaction effect for all the reach and hold metrics.

Assistance was the primary factor of interest because it represents a change in support and how it is controlled. The condition of offset assistance was significantly different from unassisted reach only for reducing the distance from target, while the condition of gain assistance was significantly lower (better) for distance from target and higher (worse) for path length, time to stabilize, and stability during hold than unassisted reach. The combined gain + offset condition reduced distance from target but increased path length and settling time. Distance from target decreased with all of the assistance conditions and had an interaction between the assistance and target direction. Time to target was the only reach and hold outcome that had a significant interaction effect between
target and assistance. The other reach and hold metrics did not have an interaction effect between target and assistance.

Figure 4.3 (a,b,c,d): Reach (a,b) and Hold (c,d) Characteristics (mean ± SD) for trials where participants could reach the target at least half of the trials without assistance (a,c) and could not reach the target without assistance (b,d). *p<0.05. na – there is no data for the metric at that target location because participants were able to reach the target (a) or participants were still not able to reach the target (b,d) or all participants were able to reach the target without assistance (b,d – near target).
Table 4.5: Statistical results for reach and hold metrics, EMG characteristics, and estimated forces based on target direction, assistance conditions. Effects with p<0.05 are in bold and italics.

<table>
<thead>
<tr>
<th>Outcome measures</th>
<th>Main Effects</th>
<th>Interaction Effect</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Subject</td>
<td>Target</td>
</tr>
<tr>
<td>Distance Metrics (N=120)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Distance from target</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Reach Metrics (N=357)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Time to target</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Path length</td>
<td>&lt;0.001</td>
<td>0.001</td>
</tr>
<tr>
<td>Hold Metrics (N=331)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Time to stabilize</td>
<td>&lt;0.001</td>
<td>0.165</td>
</tr>
<tr>
<td>Mean distance during stabilization</td>
<td>&lt;0.001</td>
<td>0.901</td>
</tr>
<tr>
<td>Stability during hold (variance)</td>
<td>0.445</td>
<td>0.223</td>
</tr>
<tr>
<td>EMG Magnitudes (N=331)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Upper Trapezius</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Pectoralis Major</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Serratus Anterior</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Anterior Deltoid</td>
<td>&lt;0.001</td>
<td>0.535</td>
</tr>
<tr>
<td>Middle Deltoid</td>
<td>&lt;0.001</td>
<td><strong>0.001</strong></td>
</tr>
<tr>
<td>Posterior Deltoid</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Biceps</td>
<td>&lt;0.001</td>
<td>0.309</td>
</tr>
<tr>
<td>Triceps</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Force Estimates (N=480)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>AP force</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>SI force</td>
<td>&lt;0.001</td>
<td>&lt;0.001</td>
</tr>
</tbody>
</table>

While reach and hold metrics were different from unassisted reach during the gain assistance, Figures 4.3a,b shows the size of the difference and that these metrics were only different for some combinations of the assistance condition and target direction. For each figure, the outcome measures are plotted
separately, based on the whether or not a participant could reach a particular target without assistance during at least half of the trials (Figures 4.3a,c), or if assistance enabled them to reach a target that they could not reach without assistance (Figures 4.3b,d). Since all of the participants could reach the near target, no characteristics are provided in Figures 4.3b,d for that direction. For some metrics (time to target, path length, and settling time for the up target in Figure 4.3a,c) gain assistance significantly increased reach and hold metrics i.e., poorer performance. The reduced performance cases all occurred in subjects who could reach the targets without assistance during at least half of the trials.

Effort to maintain the limb at the target position was assessed from the individual muscle EMG during the hold period. $P$-values for the main effects and interaction effect are provided in Table 4.5. Effort required to maintain the target positions decreased in six of the eight muscles with assistance ($p<0.05$) while two muscles did not decrease or instead increased in some directions ($p<0.05$) (Hypothesis 3). Six muscles had an interaction between the assistance condition and target direction ($p<0.05$).

EMG values for each of the muscles, with mean and standard deviation across all of the participants, are provided in Figure 4.4. There was a general decrease in EMG for most muscles when assistance was provided. In some cases, there was a further decrease when the gain and offset were used at the same time. While most of the statistically significant differences indicated a decrease in effort, the posterior deltoid and triceps had higher EMG levels in the
combined gain and offset condition when compared to no assistance or offset for the near target. There were no significant differences for posterior deltoid or triceps for the other targets or assistance conditions.

The two sample t-tests revealed that most of the reach and hold characteristics were not significantly different for participants who were enabled to reach the targets with assistance compared to those who could already reach the targets without assistance. Time to target (p=0.476), path length (0.803), variance (p=0.860), and settling distance (p=0.860) were not significantly different with assistance. Settling time (p=0.048) was lower with assistance, i.e. better.
**Force Estimates**

The passive forces that must be overcome to produce movements were measured while the arm was relaxed; average passive forces across subjects were -14N, -4N, and 2N in the anterior direction for the forward, reference, and near positions respectively and 22N in the superior direction for the reference position.

The voluntary components of the total active forces were estimated from the TDANN outputs. The total active force was the sum of the estimated voluntary force plus the force produced by the robot. Table 4.5 contains the $p$-values for the main effects and interaction effect for the voluntary force components during the reach phase of movements. In the AP directions, the interaction effect was not significant and the only significant main effects were direction and subject. In the SI directions, the interaction effect was not significant and all of the main effects were significant.

The participant voluntary force estimates (mean ± SD) provided insight into the accuracy of the TDANN estimated voluntary force, the assistance provided during trials, and whether assistance changed the effort needed for the task. The estimated forces averaged across the dynamic portions of the reach trials are given in Figure 4.5. Post-hoc analyses showed that AP force estimates were different for each target, but were not different during the different assistance conditions. The SI estimates were also different by target and were significantly smaller during conditions when more assistance was supplied.
Figure 4.5: Force estimates during the dynamic portion of the reach and hold trials (mean ± SD) while reaching to all three target locations (forward, up, and near). * p<0.05

**Individualized Assistance: Case Study**

A unique form of assistance was tested in an additional session with S2 to expand the range of assistance conditions. The individualized assistance amplified active AP forces by 3 and SI forces by 5. A 15N constant offset was also applied in the anterior direction. Without assistance S2 could only reach the near target, as shown by the ROMs in Figures 4.2c and 4.6. With either the standard combination of gain and offset or the individualized assistance, the ROM encompassed all three targets, but the ROM with individualized assistance was much larger. With the individualized assistance, S2 reached all three targets during the 15 trials and held the target position in 14 of the 15 assistance trials. The maximum assistive forces from the trials (including ROM) were 19N in the
anterior direction, 0N in the posterior direction, 31N in the superior direction, and -12N in the inferior direction.

Figure 4.6: S2’s range of motion, reach and hold metrics, and EMG while holding the near target direction with the modified assistance parameters. In addition to a 15N anterior bias, AP and SI force estimates were multiplied by 3 and 5 respectively. The offset and gain range of motion from the standard assistance set is shown for comparison. Lines extending from the perimeter show the direction of the assistive force (N) applied to the arm along the ROM. The force magnitude is divided by five to maintain scale. * - p<0.05

Discussion

The combination of offset plus gain force assistance enabled participants to reach a larger portion of the workspace and all of the different types of force assistance enabled participants to reach closer to the target positions
(Hypothesis 1). Offset assistance did not degrade movement control, but the gain assistance partially degraded performance through increased *time to target, path length*, and *settling time* for some target locations (Hypothesis 2). Participants exerted less effort with all types of assistance while holding the target positions (Hypothesis 3). While movement control was worse with assistance in some cases, maintaining movement control with less effort indicates that EMG from residual movement on the affected side can be an effective command signal for controlling assistive forces, and that residual movement coupled with assistive forces could enable functional movements.

**Force Estimates**

Trials incorporating gain assistance assumed that the voluntary force estimated by the TDANN was accurate. If the estimate was too low, the subject would not receive the intended assistance, and if the estimate was too high, the subject may have experienced difficulty controlling the movement. The $R^2$ reported here during offline testing are similar to the results for joint angle estimation reported by Pulliam *et al.* [61], and less than the results for force estimation reported by Ameri *et al.* [67] but the subject populations, training sets, and estimated outputs are different for these studies, so it is difficult to make a direct comparison.

The force estimates shown in Figure 4.5 refer to the average voluntary forces estimated while the participants were reaching towards the target, whether or not they actually reached it. The EMG results that are used for estimating the
forces indicate an interaction effect between the assistance conditions and targets, but the forces estimated by the TDANN from the same EMG do not, which suggests some errors in the force estimates. The AP force estimates were statistically different for each of the target directions but not statistically different for the different assistance types. This difference could indicate that the TDANN was able to detect differences in the direction of movement but was not sensitive enough to detect different levels of graded effort that were present in the EMG.

When the arm was relaxed, the average passive forces measured across all subjects were -14N, -4N, and 2N in the anterior direction in the forward, reference, and near positions respectively. Passive forces provided approximate expected forces for the voluntary force estimates, however this cannot be viewed as a direct comparison because the estimated forces were dynamic and the comparison forces were static. Dynamic components would be higher than their passive components due to inertial components of movements and viscoelastic musculotendon properties. Averaging estimated voluntary forces over a movement range that included lower passive stiffness could have decreased the estimated force.

It was expected that the estimated horizontal forces while reaching to those positions would be close to the passive force required to hold the target location during the no assistance and offset condition but closer to half as large in the gain conditions. Although the estimated forces were lower than the passive forces required to hold the target location, the estimated forces were not as
different as expected. Averages in the near target directions trended in the
direction of having higher estimates during the non-gain trials, but the standard
deviations were too high to detect a change of 1N-2N.

The SI force estimates appeared to be more sensitive to changes in effort
than the AP force estimates. In addition to differences based on target location,
the vertical estimates were also different based on reaching condition. Estimated
force was highest in the no assistance condition and lowest in the combined gain
and offset condition. The gain and offset were not individually statistically
different from each other, but were different from the other two conditions. These
results suggest that less effort was exerted to lift the arm as additional assistance
was provided. Furthermore, the effort exerted in the vertical direction during the
constant vertical offset condition and EMG controlled gain condition were similar,
suggesting that the gain condition could be controlled sufficiently well to reduce
effort when assistance is provided. The differences in force estimates between
conditions suggest that the TDANN successfully estimated graded levels of effort
in the SI direction. The average weight of the arm across subjects was 22N in the
reference position, but the estimated forces were below 22N in the no assistance
condition and below half (11N) in the other conditions. Despite detecting
significantly different graded effort, these estimates were still lower than
expected.

Two factors may have contributed to unreliable force estimates at high
effort levels. As described in the methods, most of the data used to train the
TDANN were less than the maximum passive force required to hold the target locations because we expected the robot to supply equal assistance during the gain conditions. As a result, the TDANN may have been less accurate when effort was exerted beyond that used during the force matching tasks as was the case in the no assistance and offset conditions. Additionally, the dynamic training trials did not account for passive forces changing as a function of endpoint position, which could contribute to the TDANN underestimating the actual force values. These errors could contribute to inaccuracies in TDANN estimates during the reaching task, particularly trials that did not include assistance in the direction of the force estimate. Furthermore, we expected participants who could not reach a target location, with or without assistance, to exert their maximal effort, and thus have approximately equal force estimates during all reaching conditions that were less than the passive force for that target direction. Since participants were instructed to lift their arm while reaching to the forward target rather than move along the bottom of the HapticMaster's workspace, they might have limited their horizontal forces in order to maintain arm elevation.

The ability of the force estimates to indicate force direction and magnitude was demonstrated by both the different estimates while reaching towards the different targets, and the differences based on the assistance condition. There were differences between the measured and expected estimates, but the magnitude and direction could be controlled to increase reach and were statistically different.
These experiments utilized a relatively simple EMG processing method and control scheme. While the training data and TDANN architecture should be further optimized to improve force estimates, these experiments demonstrated that 15 minutes of training data with a quickly trained neural network could provide force estimation that provided useful control. The amount of time spent training and completing the session is reasonable for clinical testing, but the need for repeated training is an open question and will depend on the EMG recording methods incorporated into a non-research assistive device.

**Effort for Reach**

Participants were able to reach to the targets with lower EMG levels in at least six of the eight muscles that were recorded, and with little change in the other two muscles except for the near target. Furthermore, there was no evidence that participants increased co-contraction levels to achieve stability with gain assistance, since the EMG during the gain and constant assistance were not statistically different in 92% of the combinations of muscles and targets. Therefore, in addition to less effort being required to reach targets while reaching with constant assistance, participants were able to control the amplified forces without co-contraction.

With force assistance, the triceps EMG increased for the near target direction and did not decrease for the other directions. This EMG increase was unexpected, since less effort should be required to abduct the arm, decreasing...
the resulting flexion co-activation. The increase in the near target direction could partially be a stabilizing response as a result of the assistance in the posterior direction. In addition, a forward force was exerted during the gain condition for the up and forward target directions, which implies that less triceps activation should have been necessary. Biceps EMG decreased, suggesting that elbow co-contraction was not the primary cause for triceps EMG not decreasing. There are additional muscles that produce elbow and shoulder flexion and extension that were not recorded, but it is unlikely that a decrease in EMG in these muscles would be seen if there was an increase in the recorded muscles. One potential explanation is a change in the arm orientation, as part of a stabilization strategy, changed the direction of the endpoint force produced by the triceps. Shoulder abduction and flexion, coupled with elbow extension, could produce the same endpoint position as shoulder flexion and external rotation with elbow extension. In this second case, the triceps produces a downward force, which could partially be acting to stabilize against the upward assistive forces. Thus, a change in arm orientation would account for increased triceps activation during the combination of gain and offset. There may be other muscles that have an increase in activity that were not recorded as well. Even with an increase in triceps activation, the EMG results support the idea that people can generate effective commands for an assistive device outside of the constrained movement patterns.
Control of an Assistive Device

The results shown here support using EMG from residual movements as a command signal for an assistive device. While participants were able to reach and hold the target position a few more times during the gain and offset conditions, only the combination of gain and offset provided a statistically larger workspace. As a result, it is not clear if the increased reach was a result of increased vertical support or the addition of the amplified horizontal force. A comparison of the ROM for the unique assistance set and the combined offset and gain assistance shown in Figure 4.6 shows that other combinations of the assistance parameters can provide a significantly larger workspace that is still well controlled.

The results using both the standard and the unique amplification assistance support the potential use of residual voluntary EMG as an effective command signal for an assistive device during everyday tasks. For most of the reach and hold metrics, performance with the EMG controlled assistance was similar to performance with the constant offset, including effort exerted to hold the target position. The only metric that was consistently worse with amplification was settling time. This is likely the combined result of variability in the force estimate and the user's response to the assistance during the cooperative movement. Voluntary or involuntary responses to the assistance would alter the estimated force, introducing a disturbance to the assistive force. Continuous changes in response to the dynamic assistive forces could contribute to the
increased settling time. During these experiments the robot applied pure forces and did not incorporate damping, which might have reduced the dynamic responses under the gain conditions.

Participants were given very limited time to practice reaching with the assistive forces. Further training could improve their control and improve the reach and hold metrics. Able-bodied subjects have been shown to adapt to movements through force fields [79], and stroke patients can adapt to visuomotor transformations based on their movements [156]. Adaptations in both stroke and able-bodied participants indicate that additional time spent practicing with the assistive forces may improve reaching while controlling assistance. Participants’ ability to control the assistance coupled with the increased workspace and reduced effort support the approach of using low effort as a command signal while the assistive device produces the majority of the movement.

Participant controlled assistance could be provided through either mechanical assistance like a powered exoskeleton or through FES. The forces the robot applies to the arm are similar to those produced through FES but do not account for dynamic muscle properties. Although studies have shown a reduced stimulation response during simultaneous voluntary contractions [20], the reduction in effort enabled by the assistive forces implies that greater benefit could be achieved from stimulation [116, 157]. Furthermore, the approach used here could be adapted into a screening tool for upper extremity FES systems. Forces and torques produced by an exoskeleton could be used to simulate the
joint torques that FES produces. A robot that simulates both the control interface and forces produced by FES would provide tools to evaluate the quality of command signals, as well as the effectiveness of stimulating certain muscles prior to implanting a system for use in everyday life. Such a screening system would require incorporating the mechanical properties of muscles, i.e. force-velocity and length-tension properties, as well as reflex properties, into the model that determines the forces to apply to the arm. Incorporating the mechanical properties of muscle and reflexes would help simulate the force response during FES and its effectiveness for assisting movement. Adding the stabilizing properties of muscle may also increase arm stability during the gain condition.

Participants’ Reactions

We did not provide a standardized questionnaire to the participants, but we did ask the participants’ opinions about reaching with and without assistance from the robot. The more impaired participants liked reaching with the robot assistance because it enabled them to reach further. Some perceived it as very easy to control while others still found it difficult to control. The participants who could reach the targets without assistance had mixed reactions to the assistance. One liked the assistance from the robot. Two liked the assistance, except when it added perturbations to the movement as seen in the increased settling times. The remaining participant preferred reaching without any assistance.
Study Limitations and Future Work

While this study provides evidence that it is feasible to control assistive forces with residual voluntary effort, there are additional questions to be explored. The reaching task used in this study was 3-dimensional, but all of the targets were in a single plane and the assistance was only provided in two dimensions. Because the reaching task was 3-dimensional, there is little reason to expect that reaching to targets in the perpendicular direction would be worse with the current assistance or addition of assistance in the perpendicular direction. The reduction in effort needed to lift the arm implies that the current assistance would enable additional movement in the medial/lateral directions as well. This increased movement provides evidence that a system that abducts and possibly flexes the shoulder and extends the elbow could provide a very effective increase in workspace. It would be useful to evaluate this approach by providing assistance and evaluating reach in all three directions to determine how adding a third dimension to the assistance changes controllability. The target forces used to train the model that converted EMG to force estimates provided an initial estimate of user intent, but the dynamic tracking movements did not account for position dependent changes in passive joint stiffness, which could increase the accuracy of the force estimates.

Additionally, it would be useful to record from a more complete set of shoulder and hand muscles to determine the most effective subset of muscles needed to provide the command signal and to confirm that the expression of co-
activation at the hand is low enough that stimulation can still open the hand. This study did not incorporate muscles that only adduct the shoulder, which might be useful to incorporate into the command signal.

This study demonstrated that people could control assistive forces with EMG on the affected side after stroke; however, the tested range of assistance was limited. This evaluation should be expanded to determine the limits to assistance that can be controlled poststroke. Dynamic tracking tasks, multiple point movements, and regulated hand posture would provide further insight into both the quality and limits of this control. People may improve their ability to control assistive forces after additional training. This training could be used to evaluate how people incorporate assistive forces into their own internal model of movement. This approach could potentially be incorporated into therapies, enabling people to complete tasks that they could not complete previously while requiring them to exert their own effort. The level of assistance could be decreased or eliminated if participants regain sufficient volitional ability. These experiments only evaluated muscular effort and did not examine cognitive effort. It would be useful for future studies to evaluate the cognitive load required to perform tasks with and without assistance. Reducing the cognitive effort required to control the assistance for the affected side is important for increasing the participants’ use of the affected side, potentially reducing learned non-use.
Conclusions

Participants were able to reach more of their workspace with assistance after stroke, although some aspects of movement control were worse when assistance was controlled by residual voluntary ability. Additionally, assisted movements required less effort and participants had sufficient control to hold target positions within the increased workspace. These control schemes could be used as a basis for controlling assistance from an FES system or powered orthosis to increase poststroke arm and hand function.
Chapter 5 – Conclusions, Discussion, and Future Work

Summary

The results from these three studies demonstrate the feasibility of an upper extremity FES system for reach and hand opening to improve poststroke arm and hand function sufficiently to enable bimanual tasks. Aim 1 (Chapter 2) evaluated the effect of voluntary effort to reach and open the hand on the amount of stimulated hand opening. Participants reached to a target and opened the hand with and without voluntary effort and with and without FES. The results of Aim 1 demonstrate that reducing voluntary arm and hand effort during stimulation allows stimulation to produce a greater hand opening response (Specific Aim 1a) and provide a basis for estimating the effect of voluntary effort on stimulated hand opening (Specific Aim 1b). We demonstrated that surface FES in conjunction with partial voluntary effort increased reach (Specific Aim 1a) and was able to support the arm in a way that reduced effort enough to reduce the level of co-activation at the hand. The study also demonstrated variability in subjects’ stimulation response. This variability highlights the value of understanding additional contributing factors and evaluating candidates’ stimulation response during simultaneous effort in order to determine if the stimulation response is sufficient during voluntary effort to produce useful movement.

Surface FES applied to the elbow and hand consistently produced elbow extension and hand opening during reduced effort. Surface stimulation could
support some participants’ arms (shoulder abduction and flexion) to produce movement, but others required additional mechanical assistance. In the future, stimulating the branch of the axillary nerve that innervates the anterior and middle deltoids with an implanted nerve cuff electrode may be sufficient to produce shoulder flexion/abduction. If deltoid stimulation is insufficient, it may be necessary to stimulate additional muscles involved in glenohumeral shoulder abduction/flexion and scapular rotation.

Aim 2 (Chapter 3) evaluated the summation of endpoint forces generated by voluntary effort and surface FES. Stimulation was applied while participants pushed to generate a consistent force. The change in the stimulation response was measured for different levels of effort. The results of Aim 2 demonstrate that while the interaction between voluntary effort and FES reduces the effect of FES (Specific Aim 2b), it still produces an additive endpoint force (Specific Aim 2a). These results further indicate that FES has the greatest impact when the user exerts less effort, but that effort does not completely confound the effects of stimulation. For example, the average triceps stimulation response across participants was 12N. Exerting 50% of the maximum effort produced a 45% decrease in the stimulation response, and a 36% decrease when accounting for changes in voluntary effort due to difficulty maintaining effort without visual feedback, resulting in an additional force of 6.6N or 7.7N respectively. Ideally, multiple muscles will be stimulated to produce the desired torques and forces, so the summation of stimulated forces across multiple muscles could produce a
significant amount of force to contribute to arm movement. While participants were asked to exert half effort in Aim 1 (Chapter 2) and then exerted forces to reach half of their maximum force in Aim 2 (Chapter 3), it is difficult to compare the effort exerted during these tasks because one was isometric while the other was unloaded movement.

In Aim 3 (Chapter 4), participants reached to targets both with and without assistance to determine if assistance improved reach. The assistance was either independent of or regulated by participants’ residual effort (EMG) to evaluate how well people can control assistance with their residual movement. Results in Aim 3 demonstrate that stroke patients can control assistive forces with their residual EMG (Specific Aim 3a). The assistive forces are similar to those that would be generated by an FES system. Participants who were not already able to reach half of the workspace had a significantly larger range of motion with assistance. Participants exerted less effort with assistance (Specific Aim 3b), as evidenced by lower magnitude EMG signals, in six of eight muscles during the reach and hold task. Some of the reach and hold characteristics were statistically worse with the amplification of their own effort, but participants were still able to reach to and hold the target position. Due to variations in the tasks, participants, and stimulation artifacts in the previous aims, we cannot make comparisons between the effort exerted during Aim 3 (Chapter 4) for the reach and hold task and the effort exerted during Aims 1 and 2 (Chapters 2 and 3).
Discussion

The results for Aim 3 (Chapter 5) indicate that people can control assistive forces to augment their residual movements in a manner that uses less effort and enables greater reach compared to when they reach without assistance. The ability to control movements while exerting less effort indicates that poststroke residual movement provides an effective command signal and that residual effort in conjunction with assistance improves reach. While stroke patients’ movement patterns are stereotyped, there is evidence that they can generate some movements outside of those patterns [8, 9]. Our results in Aim 3 suggest that residual movement is sufficient to control at least two degrees of freedom for reach. The ability to control assistance with submaximal effort implies that the approach of using limited effort as a command signal for an FES system has merit. Similarly, the results in Chapter 5 and the appendix imply that different effort augmentation control schemes can increase the range that patients can maintain stable movements. The assistive forces necessary to improve reach provide insight into the endpoint forces that must be produced by FES to provide effective movement. Others have demonstrated the feasibility of using EMG from muscles involved in the movement to control multidimensional visualizations [158] and single degree of freedom control of powered exoskeletons [84, 105] and FES [71, 86]. This study demonstrated that stroke participants can control 2-dimensional assistance to improve 3-dimensional reaching movements using EMG from residual movement.
The reduced effort required for reaching in Aim 3 in conjunction with the results from Aims 1 and 2 indicates that an FES system that uses low levels of effort for a command signal will have a greater effect during reaching and hand opening. Reducing effort reduces the effect of co-activation at the hand, allowing stimulation to produce greater hand opening. Reduced effort also limits the interaction of FES and voluntary effort. FES coupled with partial effort can produce greater reach. Partial effort during stimulation and reach allows stimulation to produce a greater amount of muscle force. The results in Aim 2 show that despite a decrease in the stimulation response during volitional effort, FES produces an additive force. When determining appropriate stimulation parameters, models can be adjusted to account for changes in the force output in response to the interaction between voluntary effort and FES. These results show the value of applying FES or exoskeleton support to shoulder and elbow muscles into future stroke upper extremity assistive devices. The additional support, which can be controlled with residual effort, reduces the effort that causes co-activation patterns. As a result, the FES system should be able to produce effective hand opening.

While these results are relevant in designing poststroke neuroprostheses, they also provide further insight into the mechanism causing the decrease in the stimulation response seen during simultaneous effort in stroke [20, 52, 53, 57]. The results in Aim 1 (Chapter 2) support the idea that involuntary finger flexors are activated in response to reach [12] and attempted hand opening [16],
producing undesired force that FES needs to overpower. The results from Aim 2 (Chapter 3) suggest that additional mechanisms may decrease the stimulated force response. In this aim, stimulation was applied after the participant had maintained a steady state force, at which point the change in force at the hand was measured (instead of the net output, as in Aim 1). Assuming participants were able to maintain effort, the reduction in stimulated force would not be accounted for by the co-activation from the synergy patterns, suggesting that multiple mechanisms ultimately contribute to the reduced output of poststroke FES systems.

Insight into the mechanism causing the decreased stimulation response may inform the stimulus parameters used for a poststroke FES system as well. Chae and Hart stimulated at 16 Hz [52, 159] while our studies stimulated at 35 Hz. If motor neuron resetting and occlusion of orthodromic action potentials [160] are the primary cause for the reduced stimulation response in Chapter 3, then there may have been a larger decrease in force response while testing percutaneous stimulation for poststroke hand opening. The reduced rate for stimulation requires recruiting more motor units, which would have increased the occlusion of action potentials and decreased the stimulation response. While increased firing rates fatigue muscles more quickly, a higher stimulus frequency may be necessary in a poststroke FES system.

These results show promise for the general application of an upper extremity stroke neuroprosthesis, but they also demonstrate that the system
needs to be individualized in order to meet the users’ particular needs. Results from Aim 1 show significant inter-subject variability in the co-activation response to both effort for reach and hand opening. Some subjects experienced such significant co-activation in response to full effort that stimulation could not open the hand, while in others, hand opening was less impacted by effort. As a result, the effort exerted as part of a command signal varies between subjects. Another potential method to reduce the effort dependent co-activation could use stimulated nerve block [161]. In addition to individualized ranges of voluntary effort that can be exerted as part of the command signal, different individuals will be able to control the assistance in different ways and may need different amounts of assistance. The results in Aim 3 and the appendix show that there are different ways that EMG can be mapped to assistance and that different individuals will require different levels of assistance and be able to control different types of assistance.

A poststroke FES system may only benefit a portion of the stroke population, depending on the assistance that can be provided by stimulation and how well the assistance can be controlled. As a result, it would be useful to have a screening tool to determine which stroke subjects are eligible, what muscles would be most effective for recording and stimulation, and what general control scheme would be most effective. This would provide potential neuroprosthesis candidates with a general understanding of how the system could change their functional ability.
Future work

These studies used surface stimulation and recording to answer important questions related to the feasibility of a poststroke FES system, but there are several questions that should be answered to more completely assess the probability of success of an implanted upper extremity poststroke FES system. A primary concern is the stimulation response at the shoulder. While stimulation was able to produce a functional amount of arm movement in some participants without mechanical assistance, the level of movement was not consistent between participants and it is desirable to attain greater arm elevation. Additionally, the ability to estimate and subsequently control arm movements and simultaneous hand opening movements needs to be demonstrated in all three dimensions. It should also be confirmed that an effective command signal can be detected during simultaneous stimulation. We have presented initial estimates of the interaction of voluntary effort and FES, but it would be useful to understand more completely the effect of modulating both voluntary effort and stimulation intensity on their interaction.

FES for Shoulder Abduction and Flexion

One way to approach the limited arm movement produced by FES would be to determine the additional torque that FES would need to generate at the shoulder using a protocol similar to that used in Aim 2 (Chapter 3). Passive shoulder torques, including those to gravity, could be measured at different levels of shoulder abduction and elbow extension to determine the necessary shoulder
torques required to support the arm in those postures. After recording maximum voluntary contractions and the maximum torque that can be produced by stimulation in those locations, the assistive torques produced by FES of the deltoids could be determined under different levels of voluntary effort. Compared to the initial passive torques, the stimulation response would provide the necessary shoulder torque that must be provided by other shoulder muscles. If strength data for those other muscles indicates that they would be suitable candidates to provide the missing torque, then stimulation may be able to produce sufficient torques at the shoulder. If FES is not sufficient to support the arm, an exoskeleton or other mechanical system may be able to provide the necessary support to reduce the expression of the synergy patterns, allowing FES to provide functional assistance at the hand.

_Detailed Interaction Between Voluntary Effort and FES_

Results from Chapter 2 and 3 suggest that a more detailed study of the interaction between voluntary effort and FES would provide greater insight into the level of effort that can be generated while simultaneously producing a useful FES response. The data presented here for 5 subjects show an interaction between voluntary effort for reach and hand opening. Studies using isometric force generation as an indicator of effort would provide more exact parameter estimates for the effect of effort. Force measurements should be performed across a range of effort at multiple joints to determine the interaction more fully.
Based on variability in the patient responses, a broader set of participants should be included. Patient information including impairment measures; spasticity, residual movement, and residual tone; and other patient information about stroke type and lesion location should be incorporated into the model to determine indicators that will determine which patients are most likely to have an effective response to FES during effort.

Furthermore, it would be useful to have a model that estimates the change in assistive force as a function of both voluntary effort and stimulation level, instead of only voluntary effort. It would also be useful to examine the changes in joint torques in addition to endpoint forces. While there is definite value in determining the effect of effort on hand opening, surface stimulation has poor selectivity and most joints are controlled by multiple muscles, making it difficult to determine the effect on a particular muscle. Evaluating the interaction between FES and voluntary effort at a joint with fewer muscles would provide more accurate parameter estimates for specific mechanisms of interaction.

For example, change in the stimulation response during simultaneous effort could be measured during voluntary finger abduction and stimulation of the first dorsal interosseus to provide insight into the response in a single muscle. This experiment should be done in both able-bodied and stroke participants to confirm that the observed changes in force/EMG response are due to the hypothesized reasons [103]. Simultaneous force generation could be used to evaluate M-wave removal techniques [86, 141, 143], since a poststroke
neuroprosthesis could benefit from stimulating and recording from the same muscle. The same experimental setup could be used to evaluate cooperative control of movement and force production with FES at a single joint.

*Simultaneous Arm and Hand Intent Estimation and Stimulation*

Aim 3 evaluated control of 2-dimensional assistance and demonstrates some success in estimating force intention, but it would be useful to examine simultaneous intent estimation for the shoulder elbow and hand within the desired workspace. Others have shown the ability to detect intent in the paretic hand [76-78], even during simultaneous partial shoulder abduction [78]. The poststroke synergy patterns may provide a mechanism for determining what independent movements patients can control by decomposing EMG signals into their base sets [62]. Additionally, intent estimation needs to be evaluated while controlling stimulation. EMG has been used to control stimulation within a single degree of freedom [86], but EMG control would need to be expanded to confirm that intent estimation from multiple muscles during stimulation is adequate for producing functional movements.

*Functional Evaluation and Patient Training*

The effectiveness of the FES system would primarily be measured by how well the user can perform bimanual and unilateral tasks with the affected arm and by whether bilateral movements affect control of the affected arm. It is important to evaluate system use in the lab and when the user is at home, where the user may not consider using their affected arm because of learned disuse and
compensation techniques. As a result, users may require significant training to learn to use the system on their own and to relearn using their affected arm. The control system should incorporate natural control, but training will help the user adapt to using the system and performing tasks bimanually. Such training should involve a therapist and possibly an engineer or an individual thoroughly familiar with the system. The training regimen may in fact incorporate therapy that a stroke patient is already undergoing.

This training period would be useful for understanding how people learn to use an assistive device and the impact on cognitive load. The cognitive effort required to use the system should decrease as the user integrates the assistance into their natural control strategies, but it is not currently clear how long the training process needs to be. Users may benefit from more than the direct functional assistance, because of the system over time would encourage active use of the affected arm to perform tasks, which is a key component of therapy. The increased function may further enable therapeutic gain; moreover, arm use facilitates active stretching, which would help prevent contractures, reduce swelling, and increase blood flow.

Neuroprosthesis Screening Tool

In addition to demonstrating the feasibility of a poststroke neuroprosthesis, results related to control of assistive movements and the interaction between stimulation and simultaneous voluntary effort provide a framework for a screening tool to determine which patients would benefit from a neuroprosthesis.
and the potential effect the system could have during activities of daily living. Chapter 2 showed high variability in participant’s stimulation response during voluntary effort, indicating that different participants may require FES to provide varying levels of support and that different people will be able to generate varying levels of effort for a command signal. A screening tool would provide confidence that someone would be able to functionally control the neuroprosthesis as well as give the user some sense of the function they might gain. The purpose of the screening is to determine 1) deficiencies in reach and grasp that need to be improved, 2) if the candidates impairments indicate that stimulation during simultaneous effort will be effective 3) the assistance needed to provide functional reach and grasp, 4) effective control schemes, 5) muscles to target for recording and stimulation, and 6) potential benefit to the patient.

**Conclusions Summary**

Through the studies described above, we have evaluated the effect of reduced effort on stimulated hand opening, as well as the interaction between FES and voluntary effort in the same muscle. Additionally, we have developed an approach to provide assistance that can be controlled intuitively by the affected side. In conclusion, these results demonstrate that an upper extremity poststroke assistive device could be controlled and improve movement, potentially providing significant benefit during activities of daily living. The results also provide insight into future considerations about the design and implementation of poststroke assistive devices.
Appendix 1: Unique assistance patterns to increase poststroke movement patterns

Introduction

The assistance conditions outlined in Chapter 4 indicate that assistance can increase range of motion (ROM) and be well controlled. While there is value in assessing how well multiple people can control the same types of assistance, different people may be able to control different levels of assistance after stroke and may need different amounts of assistance to reach a functional work area. Participants who had difficulty reaching to the targets with the standard set of assistance conditions described in Chapter 4 also participated in one additional session with a unique set of assistance parameters (Table 6.1). These individual case studies consisted of combinations of an offset and gain in both the Anterior/Posterior (AP) and Superior/Inferior (SI) directions. The primary goal in choosing these parameters was to enable participants to reach all of the target positions. The secondary goal was to decrease the muscular effort required to do so by increasing the gain. The chosen parameters represent only a portion of the parameter space that should be considered.

Voluntary force estimates are in the AP and SI directions described by the terms $x_{est}$ and $z_{est}$ respectively. Force applied in the AP and SI directions are denoted by $\hat{i}$ and $\hat{j}$ respectively.
Table 6.1: Summary of the case study parameters and results for each participant.

- \( x_{est} \) – Estimated force in the anterior/posterior directions
- \( i \) – Force applied in the anterior/posterior directions
- \( z_{est} \) – Estimated force in the superior/inferior directions
- \( j \) – Force applied in the superior/inferior directions

*The ROM measure was not performed during these sessions. This number represents the area enclosed during all of the reach trials for a given condition.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Assistance</th>
<th>Voluntary ROM (cm²)/Assistance ROM (cm²)</th>
<th>Reach/Hold success without assistance [%]</th>
<th>Reach/Hold success with assistance [%]</th>
<th>Max/Min forces [N]</th>
</tr>
</thead>
<tbody>
<tr>
<td>S2</td>
<td>((3<em>x_{est})+15)i ( ((5</em>z_{est})+0)j )</td>
<td>194/823</td>
<td>F - 0/0</td>
<td>U - 0/0</td>
<td>B - 100/100</td>
</tr>
<tr>
<td>S5</td>
<td>PI Control With damping</td>
<td>124/307*</td>
<td>F - 0/0</td>
<td>U - 0/0</td>
<td>B - 100/100</td>
</tr>
<tr>
<td>S7</td>
<td>((3<em>x_{est})+12)i ( (2.5</em>z_{est})+4)j ) With and without damping</td>
<td>196/439/316*</td>
<td>F - 0/0</td>
<td>U - 0/0</td>
<td>B - 100/80</td>
</tr>
<tr>
<td>S7</td>
<td>Integrator</td>
<td>126*/432*</td>
<td>F - 0/0</td>
<td>U - 0/0</td>
<td>B - 100/100</td>
</tr>
<tr>
<td>S8</td>
<td>PI Control</td>
<td>648/746</td>
<td>F - 100/100</td>
<td>U - 100/100</td>
<td>B - 100/100</td>
</tr>
</tbody>
</table>

Changes to assistance

While voluntary force can be amplified, there is an upper limit that can be controlled without producing unstable movements. Errors in the force estimates, delays in the system response, and poorly controlled movements by the participant all contribute to undesired changes in the assistive force. Amplifying this variability in the desired assistance can create a system that participants cannot control. To overcome this variability, we also investigated an additional method of controlling the assistive force as well as a potential damping field. Damping was added to smooth some of the undesired dynamics, reduce the likelihood that stretch reflexes are evoked, and mimic the increase in active...
muscle dynamic stiffness produced by FES. Damping force was generated based on the damping response of activated muscle spindles [162]. Gielen and Houk evaluated the effect of perturbations to a wrist that is already generating a load. Their results found a non-linear velocity response of $\Delta F = 0.24v^{0.17}\Delta x$ where $\Delta F$ is the force response (N); $\Delta x$ is the muscle length displacement (mm); $v$ is velocity (mm/s). Since multiple muscles contribute to the movement, a possible muscle length change was estimated for the triceps and the following damping field was used $F_{damp} = 2.5v^{0.17}$, where $v$ is measured in cm/s. The author made a conversion error and the units for velocity should have been mm/s. The resulting damping force was less than intended, but still within a few Newtons of the desired damping force.

Some of the unique sets of assistance have a relatively high level of constant assistance. While constant assistance could be practical with an exoskeleton, an FES system that continuously stimulates to produce an endpoint force of 15N is unrealistic and will cause rapid and unnecessary fatigue. Another way to provide a large offset force when needed is by having a slowly responding integrator that integrates the estimated force, allowing the user to shift the resting position over time. A fast responding proportional gain in parallel with this slowly modulated force (Figure 6.1) provides both quickly and slowly reacting force assistance. This combination of assistance allows for a quick response, but the assistance is not so dynamic that it cannot be controlled. An additional slow decay could be added to the force integrator so a small level of effort is required.
to maintain that assistance. The slow decay would not require the user to bring the system near to the rest position. This additional decay was not tested in these experiments, but could be useful for functional implementation.

From the perspective of the system, it is still a feed-forward controller. Having the user in the loop provides feedback, thereby creating a feedback system that incorporates the user as part of the plant in the overall system.

Figure 6.1: Diagram of PI control method.
Results for Individual Controllers

S2 Assistance – Direct Assistance Parameters

The force assistance for participant S2 can be summarized as $F_{\text{assist}} = ((3x_{\text{est.}})+15)\hat{i} + ((5z_{\text{est.}})+0)\hat{j}$ where $\hat{i}$ represents force applied in the AP directions and $\hat{j}$ represents force applied in the SI directions. $x_{\text{est.}}$ and $z_{\text{est.}}$ are the force estimates in the AP and SI directions respectively. No damping was applied and no integration was performed. A strong forward bias was applied and the participant used their own amplified effort to pull back towards themselves. Vertical assistance consisted of only a gain. The gains in the two directions were not equal, i.e. a non-uniform gain field.

This unique assistance enabled S2 to reach all of the targets. In addition to the reaching task, S2 tried to reach in as large a circle in the sagittal plane (ROM) as possible. Figure 6.2 shows an increase of about four times the size of the area that could be reached without assistance. There was an increase in the Time to Target and Path Length while reaching to the near target, but these can be partially explained by the experimental setup. Participants are relaxed when they are given the cue to start reaching and the robot starts to apply forces to the arm. With a strong forward bias, there was a 15N forward force applied to the arm pulling forward until the participant started to pull back towards the target. Despite the 15N forward bias force, the participant still exerted less effort in three of the eight muscles while reaching towards the near target.
S7 Assistance – Direct Assistance Parameters, Damping, and Integrator

Participant S7 came in for two sessions to test unique assistance sets. In one case, the assistance can be summarized as

\[ F_{\text{assist}} = ((3 \times \text{est.}) + 12)^\uparrow + ((2.5 \times \text{est.}) + 4)^\downarrow, \]

and was tested both with and without a damping field. Results are shown in Figure 6.3. S7 was able to reach to all of the target positions without damping, but could not maintain the near position, as shown in Table 6.1. Adding damping enabled the participant to hold the near position, but also increased difficulty reaching to the up position. S7 performed the range of motion measurement for the assistance without damping and the no assistance condition, but not the damping condition. Reaching characteristics were not
different with and without the damping in the assistance condition. Less effort was required to hold the near target position with assistance in the damping field, but without damping, the participant had considerable difficulty maintaining a stable position. Direct amplification may enable participants to reach new target positions, but too much amplification cannot necessarily be held consistently without additional processing or smoothing. Damping produced by muscle mechanical properties and soft tissues may provide additional stability, increasing the ability to hold target positions.

Figure 6.3: Range of motion, reach and hold metrics, and EMG values for S7 with the following assistance, $F_{assist} = (3\times z_{est.})^2 + (2.5\times z_{est.})^4$ both with and without damping. Range of motion presented during the damping condition is the composite shape of all the reaching trials because range of motion was not separately measured during the damping condition. * - p<0.05.
S7 also tested a pure integrator with proportional gain, \( K_p = 0 \) and integrator gain, \( K_i = 0.5 \) in both the AP and SI directions. The thresholds were 3 and 2.5 in the AP SI directions respectively. Results are shown in Figure 6.4. The movements were slower than other participants for forward reach, but the participant was able to reach all of the targets with the integrator. Range of motion was not measured, but the reaching characteristics for the near directions were not different and the participant required less effort to maintain the near position. This indicates that the participant could control the force over time to change the bias force that is assisting with the movement. Part of the reason for the slow movements is due to the slow time that it took for the integrator to add to the force in the forward direction, making it slower to produce the movement. While the participant was able to more successfully reach and hold the target positions than with direction amplification and offsets, it took more time to change the bias force. This indicates that a combination of these two methods may be beneficial.
Figure 6.4: Range of motion, reach and hold metrics, and normalized EMG values for S7 with the integrator assistance. Range of motion presented is the composite of all the reach trials because range of motion was not measured. * - p<0.05.

**S5 Assistance – PI Control**

S5 used the PI control method described in Figure 6.1. PI control was tested with the following parameters: proportional gains, $K_p = 2$ and $K_p = 0.5$ in the AP and SI directions; integrator gains, $K_i = 1$ and a threshold of 1N in the AP directions and $K_i = 0.2$ and a threshold of 9N in the SI directions; and a constant forward bias $C=5N$. S5 was enabled to reach all of the target positions and was able to exert less effort while reaching to the near target position as shown in Figure 6.5. While the resulting movements may be slower than a pure amplification and offset, this method may provide a balance between speed and
stability. Considering the small sample size and the different assistance parameters used with each participant, it is not reasonable to compare participants.

![Graph showing active range of motion](image)

**Figure 6.5:** Range of motion, reach and hold metrics, and EMG values for S5. Range of motion presented is the composite of all the reach trials because range of motion was not measured. * - p<0.05.

**S8 Assistance – PI Control**

Interestingly, participant S8 was able to reach to and hold all of the target positions during the follow up session, despite not being able to consistently hold some of the target positions during the previous session. S8 also used a combination of the proportional and integral assistance and the following
parameters: proportional gains, $K_p = 0.5$ in both the AP and SI directions; integrator gains, $K_i = 0.25$ and a 3N threshold in both the AP and SI directions.

Results are shown in Figure 6.6. None of the reaching characteristics were different, while the only significantly different muscles responses while holding the near target position were the upper trapezius, anterior deltoid, and biceps. The range of motion was only different by 98cm^2, but only one loop for the range of motion was measured during the assistance condition. While the results do not look remarkably different for this participant, the participant’s perspective was that it was much easier to reach with the assistance and that reaching required a lot less effort.

![Active Range of Motion](image)

**Figure 6.6: Range of motion, reach and hold metrics, and normalized EMG results for S8. * - p<0.05.**
Summary

While the results in Chapter 5 show that reaching can be improved with assistance, the results in Appendix 1 demonstrate that higher levels of amplification, even as high as 5 in some cases, can be well controlled. Additionally, these results demonstrate the value of exploring other methods to control the force assistance over time, like the integrator approach.

Pure forces were applied to the arm in Chapter 4, but if these forces are used to simulate assistance from FES, there is value in incorporating damping to mimic muscle properties. The damping characteristics applied to participants S5 and S7 are similar to the damping response from contracted muscles. The damping provides stability to the assistive forces and may reduce reflex effects that the assistive forces evoke.
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