Development of a Functional Electrical Stimulation Rowing System and Coaching Application

by

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A thesis submitted in conformity with the requirements for the degree of Masters' of Applied Sciences
Graduate Department of Institute of Biomaterials and Biomedical Engineering
University of Toronto

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Abstract

Functional electrical stimulation (FES) involves electrically stimulating paralyzed muscle to contract and has been integrated into rehabilitation exercises for individuals with spinal cord injury (SCI). Unlike other exercises, FES-rowing requires many training sessions for beginners to learn the appropriate timing of administering FES to their legs through a manual push button, required for performing the exercise with correct technique. Therefore, the purpose of this study was to develop a coaching system to instruct new rowers on the optimal timing for administering FES.

We analyzed the temporal and spatial differences between leg muscle activation onset in relation to seat and handle position peaks in 10 able-bodied individuals on a custom-modified FES-rowing ergometer. We identified a target range of seat positions at which to administer FES. Based on this result, we developed and validated a coaching app in 7 individuals, which was shown to improve relative timing of manual button press.
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# Table of Contents

Acknowledgments ........................................................................................................ iii

Table of Contents ........................................................................................................ iv

List of Figures .............................................................................................................. vii

1 Introduction ............................................................................................................... 1

2 Literature Review: .................................................................................................... 3
  2.1 Benefits of FES-rowing ....................................................................................... 3
    2.1.1 Cardiovascular Benefits .............................................................................. 3
    2.1.2 Musculoskeletal Benefits .......................................................................... 3
    2.1.3 Bone Density Benefits ............................................................................... 4
  2.2 Rowing Technique ............................................................................................... 4
    2.2.1 Symmetric Force Production & Leg Length .............................................. 5
    2.2.2 Lower Back Strain: Lumbar Extension ..................................................... 5
    2.2.3 Sequential muscle activation patterns (Legs, Trunk, Arms): ..................... 6
  2.3 Disparities in Leg Muscle Activation During Rowing ....................................... 7
    2.3.1 Effect of Auditory Cues ............................................................................. 8
    2.3.2 Effect of controlled stroke rate .................................................................. 9
  2.4 FES During Rowing ............................................................................................ 10
  2.5 Coaching and Feedback Systems for Rowing .................................................... 12
    2.5.1 Auditory vs. Visual Feedback ................................................................... 12
    2.5.2 Frequency of feedback ............................................................................ 12
  2.6 Summary .............................................................................................................. 13

3 Research Objectives & Hypothesis .......................................................................... 15

4 Hardware: Development of FES-Rowing System .................................................... 16
  4.1 Custom footplates, back and shank support ..................................................... 16
  4.2 Mounting Position Sensors .............................................................................. 17
Experiment 1: Determine Timing of Leg Muscle Activation in Able-Bodied Rowers

5.1 Introduction

5.2 Methodology

5.2.1 Participants

5.2.2 Experimental Design

5.2.3 Data Analysis

5.3 Results:

5.3.1 Leg Muscles Activation Patterns

5.3.2 Determining Optimal Timing of Stimulation

5.4 Discussion:

5.4.1 Leg Muscle Activation Patterns:

5.4.2 Temporal Differences Between Kinematic Events and VL Onset:

5.4.3 Spatial Difference Between Kinematic Events and VL Onset:

5.4.4 Influence of Metronome:

5.4.5 Identifying the Target Seat Position Range for Manual Button Press:

6 Software: Development of a Coaching System to Instruct Users on the Optimal Timing to Administer Stimulation for FES-rowing

6.1 Appropriate Timing for Manual Button Pressing

6.2 FES-Rowing with Coaching System

6.3 Feedback Modality for GUI of Coaching System

6.4 Delays

7 Experiment 2: Testing and Validation of Coaching System

7.1 Introduction:
7.2 Methodology ................................................................................................................. 49
  7.2.1 Participants: ............................................................................................................. 49
  7.2.2 Protocol: .................................................................................................................. 49
  7.2.3 Data Analysis .......................................................................................................... 51
7.3 Results: ............................................................................................................................ 51
  7.3.1 Effect of Coaching System on Seat Position During Button Press ....................... 51
  7.3.2 Effect of Coaching System on Timing of Button Press ........................................ 53
7.4 Discussion: ....................................................................................................................... 57
  7.4.1 Summary and Explanation of Main Findings ......................................................... 57
  7.4.2 Coaching Significantly Influences Timing of Button Press for FES ..................... 57
  7.4.3 Coaching does not significantly influence seat position during button press for applying FES ......................................................................................................................... 58
  7.4.4 Relate the findings to similar studies ........................................................................ 59
  7.4.5 Implications for Testing in Individuals with SCI .................................................... 60
  7.4.6 Alternate Explanations for Findings ......................................................................... 60
  7.4.7 Visual and Auditory Feedback Systems: ................................................................. 60
  7.4.8 Limitations ............................................................................................................... 61
  7.4.9 Suggestions for Further Research ........................................................................... 62
8 Conclusions .......................................................................................................................... 63
9 References ............................................................................................................................ 64
10 Appendix: Closed Loop Automatic Stimulator Option .................................................. 70
  10.1 Introduction: .............................................................................................................. 70
  10.2 Rationale for Seat Position Based Controllers Rather than Time Dependent Controllers ................................................................................................................................. 71
  10.3 Testing and Validity of Automatic Stimulator ......................................................... 72
  10.4 Benefits of Closed Loop Stimulator .......................................................................... 73
  10.5 Drawbacks of Closed Loop Stimulator .................................................................... 74
List of Figures

Figure 1. Handle and foot force distributions between AB and SCI throughout one stroke [5]. Able-bodied individuals initiate the rowing stroke with their legs (peak of dotted line), while leg forces are delayed in individuals with spinal cord injury [5]. Cited and modified from [5]. ........... 7

Figure 2. Vastus lateralis muscle activation after normalization and averaging from 13 individuals during the drive (left) and recovery (right) phase. The x-axis is percent of rowing stroke, and y-axis is the percent of maximal EMG activity. Cited and modified from [20]. ........... 8

Figure 3: A collection of stimulation parameters from 28 FES-rowing articles demonstrating the stimulator used, target muscles, frequency chosen, maximum intensity level and pulse width. A pulse width of 450 us, a frequency of 40 Hz or 50 Hz, and a maximum intensity up to 115 mA was most commonly used as a threshold when the muscle fatigued. ................................................. 11

Figure 4. Modified Rowing Ergometer with Sensors. The modified rowing ergometer for individuals with SCI includes a back rest and shank support. This was instrumented with position and force sensors for data collection. Position sensors were used to determine the handle and seat position. Force sensors were used to determine the handle and foot forces. ......................... 16

Figure 5. Custom footplates. The first picture displays the outer casing of the footplate encasing the sensors. The second picture displays the normal force sensor (GS1240-250 Kg-SS), and the final picture displays the setup for the shear force sensor (SML-300; Interface). Cited and modified from [43]............................................................. 17

Figure 6. String Potentiometer Attachment. The string potentiometer was mounted at the posterior end of the rowing ergometer. The string was attached to a hook and flat metal piece that was screwed in to fit snugly underneath the seat. ................................................................. 18

Figure 7. Calibration results from sensors in custom built footplates: 2 normal force (top and bottom left) and 2 shear force sensors (top and bottom right). Each of these plots show the linear relationship between voltage and force................................................................. 19

Figure 8. Calibration of string potentiometer sensors using motion capture. Voltage reading converted to distance (red) is overlaid with the distance determined by motion capture (blue). . 21
Figure 9. Modified rowing ergometer with back support, spring and force and position sensors.

Figure 10. Leg muscle EMG patterns, handle and seat position, and foot and handle force in a typical rower during 1 stroke cycle. Solid line indicates average, and dotted lines indicate standard deviation after 30 strokes were normalized and averaged. Electromyograms from Rectus Femoris (RF), Vastus Lateralis (VL), Vastus Medialis (VM), Biceps Femoris (BF), Soleus (SOL), Medial Gastroc. (MG), Lateral Gastroc. (LG), Tibialis Anterior (TA). Kinematic and kinetic data such as Seat Position (SP), Handle Position (HP), Foot Force (FF), Handle Force (HF).

Figure 11. The timing difference (seconds) between various kinematic events (posterior seat position, posterior handle position, anterior handle and seat position) and VL onset across 4 different trials. Dark orange bar indicates preferred stroke rate, with a metronome. Light orange bar indicates preferred stroke rate without a metronome. Dark green bar indicates 30 stroke rate with a metronome. Light green bar indicates 30 stroke rate without a metronome.

Figure 12. Vastus lateralis EMG (red), seat position (green), and handle position (blue) during one sample strokes across 1.75 seconds. Vertical colored lines show EMG onset (red), the anterior kinematic events are shown by the green seat trough and blue handle trough, and the posterior kinematic events are shown by the green seat peak and blue handle peak.

Figure 13. The average seat and handle position during VL onset from 10 participants are shown in the graphs above for each trial type.

Figure 14. Sensor data is used to provide real time feedback on the optimal timing of button press to administer FES to the legs, which sends a signal to the electrical stimulation and activates the quadriceps for knee extension during the drive phase. Releasing the button when the seat reaches the posterior end of the ergometer will cause stimulation to the hamstrings only for the recovery phase.

Figure 15. The coaching system interface provides real time feedback on manual button press for administering FES, average power per stroke and stroke rate. A second window provides power output feedback for the duration of a single stroke.
Figure 16. Effect of Coaching System on Seat Position During Button Press in a Typical Subject. The graphs show the seat position during every single button press in no coaching (blue) and coaching (orange) trial respectively. Dotted line shows the acceptable target seat position range of administering FES based on experiment 1.  

Figure 17. Seat position during button press in 7 subjects in no coaching (blue) and coaching conditions (orange). Seat position values were not normally distributed so they are displayed as a box plot.  

Figure 18. Effect of Coaching System on Timing of Button Press in a Typical Subject. Graphs show timing of button press relative to the anterior most seat position (time diff. of zero sec; vertical black bar) in no coaching (blue) and coaching (orange) condition respectively. Positive timing difference values indicate the correct button presses which happened prior to anterior seat peak. Negative values indicate incorrect/delayed button presses.  

Figure 19. Timing difference between button press and seat peak in 7 subjects. Positive timing difference values indicate correct button presses that occurred before the seat reached the anterior peak. Negative time difference values indicate incorrect button presses that occurs after the seat passed the anterior peak.  

Figure 20. Closed loop automatic stimulator. The computer reads the position of the seat at a sample rate of 200 Hz. It sends a signal to the electrical stimulator to administer FES to the appropriate muscle when the seat reaches a target seat position. For example, when the seat reaches the target anterior-most position, it stimulates the quadriceps for knee extension during the drive phase. When the seat reaches the target posterior-most seat position, it stimulates hamstrings for knee flexion during the recovery phase.  

Figure 21. The seat position at the moment of stimulation was recorded across 30 consecutive strokes in 5 subjects. The target seat position for administering stimulation was based on the seat position during leg muscle activation as determined by the results from Experiment 1.  

Figure 22. The timing difference between button press and seat position peak was calculated across 30 consecutive strokes in 5 subjects. The positive difference values indicate that the stimulation occurred around 0.3 seconds before the seat reached the anterior most position.
1 Introduction

Spinal cord injuries (SCI) affect around 500,000 individuals per year around the world [1]. These individuals often experience a deterioration in their cardiovascular and musculoskeletal system due to physical inactivity after injury [1]–[7]. There are several aerobic exercises that aim to counter-act these comorbidities and decrease the risk of cardiovascular disease and fractures due to low bone mineral density [1]–[3], [6], [8], [9]. However, individuals with a higher level of injury had paralysis of a greater amount of muscle mass which significantly restricts an individual’s aerobic capacity [1]. This prevents them from exercising at the high intensities necessary for reducing cardiovascular risk [1]–[3]. For example, arm-crank ergometry for the upper limbs of individuals with incomplete paraplegia does not engage enough muscle mass to induce sufficient aerobic demand [1]. The next progression was incorporating the paralyzed leg muscles into rehabilitation exercises, which elicits higher aerobic demand.

Functional electrical stimulation (FES) induces artificial contractions in paralyzed muscle using an electric current. FES of leg muscles allows for engagement of greater muscle mass and has been used to promote cardiovascular health in individuals with SCI, by use of various exercises [1], [2]. FES-cycling exercise is the most popular, nevertheless it has been shown that the exercise intensity during FES cycling tend not to meet the aerobic demand for reducing cardiovascular risk. Consequently, hybrid FES exercises may be the key. Hybrid FES exercises involve voluntary movement of the upper limbs coupled with FES to the quadriceps and hamstrings to engage the lower limbs [1]–[3]. For example, hybrid FES-cycling involves the coupling of FES cycling with arm-crank ergometry. Hybrid FES exercises have shown greater aerobic demand measured by peak oxygen consumption (26.5 – 35.7 mL/kg/min) relative to other clinical exercises (i.e. arms only: 15.7 mL/kg/min or FES-cycling: 14.3 mL/kg/min exercises) [1]–[4], [10]. Another form of hybrid FES-exercise, known as FES-rowing, emerged with modifications to standard ergometers to enable individuals with SCI to perform this full body exercise [10]. In FES-rowing, the arms and legs are engaged in a coordinated manner, with a plethora of additional benefits including the activation of the lower limb muscle pump for venous return, cardiac hypertrophy, reduced leptin and glucose levels in the blood,
improvements in bone density, and greater muscle mass [1–3], [6], [8], [9], [11]. Thus, FES-rowing is one of ideal exercises improving the cardiovascular health in individuals with SCI.

Although hybrid FES-cycling is relatively easy to learn, FES-rowing is a complex full body exercise; individuals with spinal cord injury require many training sessions to obtain sufficient skill and learn the timing of administering FES to their lower limbs to perform the movement correctly[12]. Able-bodied individuals rely more on and initiate the rowing stroke with their legs, but individuals with SCI may initiate the rowing stroke with their arms leading to low force production and minimal involvement of the legs [5]. This is a critical issue because improper FES timing to the legs impairs coordination between upper and lower limbs, which prevents the individual from receiving the benefits of this exercise [5].
2 Literature Review:

2.1 Benefits of FES-rowing

2.1.1 Cardiovascular Benefits

The limb paralysis resulting from SCI can inhibit individuals from achieving the high exercise intensities that are necessary for counteracting cardiovascular disease [1]. Therefore, it is necessary that the individual’s rehabilitation plan involves exercises with high aerobic demand [1]. There are a plethora of studies comparing the cardiovascular benefits of FES-rowing relative to other clinical exercises in individuals with SCI [1–3, 10, 13–15]. An early study compared 1) arms only rowing, 2) legs only stimulation, and 3) the hybrid exercise combination: both arms-rowing and legs-stimulation. The combined arms-rowing and leg-stim was shown to have the greatest increases in oxygen uptake (VO2 of 16.38 nm ml/kg/min: 83% of maximum). These studies demonstrated that a coordinated full body exercise combining arms-rowing and legs-stimulation had increased oxygen consumption because of greater muscle mass involved in the FES-rowing exercise [2]. Invoking greater muscle mass is useful for aerobic training to reduce cardiovascular disease. Furthermore, hybrid training is superior to arms-only exercises because targeting the legs reduces venous pooling [3]. These benefits are greater in individuals with higher level injuries. They have greater paralyzed muscle mass and lower initial aerobic capacities; thus, they experience a greater relative increase in aerobic capacity. For example, an individual with a C5 injury doubled their peak aerobic capacity but an individual with a T4 injury only experienced a 20% increase [3].

Cardiac structure and function were also altered after SCI individuals underwent FES-rowing training. Five SCI individuals who consistently participated in FES-rowing had cardiac structure and function that was similar to able-bodied individuals. Ultrasound (echocardiogram) analysis demonstrated improvement in cardiac wall thickness during diastole after FES-training, which may indicate cardiac hypertrophy [7].

2.1.2 Musculoskeletal Benefits

Spinal cord injury results in a muscle fiber shift from type I (slow twitch, endurance) towards type II fibers (fast, anaerobic, highly fatigable) within weeks after SCI possibly due to
inactivity[16]. However, several studies have shown increases in muscle cross sectional area, changes in the myosin heavy chains expressed, and improvements in capillary density after FES training [16].

2.1.3 Bone Density Benefits

Spinal cord injury often results a decrease in physical activity, which causes a decrease in bone mineral density in the femur and tibia leading to fractures [6], [8], [9]. Counteracting this loss of bone can be partially accomplished by exercise involving loads of 1.5 times the individual’s body weight [6]. One study demonstrated that one FES-rowing trained participant had higher trabecular bone mineral density than expected in chronic SCI individuals[8]. Therefore, this exercise provided joint contact forces above the necessary threshold for preventing this loss of trabecular bone density [8]. Another article established a stress threshold value which would result in remodeling of the tibia during FES-rowing [6]. Specifically, Gibbons used positive quantitative computed topography (p-QCT) analysis to assess bone mineral density in one participant’s ultradistal tibia, who had been training with FES-rowing for over 2 years [9]. This participant’s bone mineral density was within the range of able bodied individuals (within 0.57 s.d.) and far above SCI individuals (1.66 s.d. away from able bodied). Therefore, the loading during FES-rowing has been shown to prevent the bone mineral density loss normally seen in SCI individuals [8], [9].

Safety considerations are essential for developing an exercise in the spinal cord injury population. 3D stress analysis demonstrated where the loading stress was applied on the tibia and it was revealed that there were not unsafe levels of stress applied [6]. Studies across several years also reveal that there have not been any skeletal issues or fracture risk reported across sample sizes of over 20 participants emphasizing that this is a safe exercise with bone related benefits [8].

2.2 Rowing Technique

There are several factors that contribute to optimal rowing technique including sequential muscle activation patterns, symmetric force production, safe lumbar extension and knee joint angles.
Cerne (2013) compared rowing technique between three groups of participants: non-rowers, junior rowers, and elite rowers [17]. It was found that elite rowers had more consistent technique irrespective of the stroke rate [17]. To elaborate, in elite rowers, the stroke length was not dependent on stroke rate and was consistent at all stroke rates [17]. Stroke length of junior rowers and non-rowers were only consistent at same stroke rate. Elite rowers had a normalized stroke length (stroke length / height) target of 0.83, junior rowers had normalized stroke length of 0.87, and non-rowers had shortest stroke length due to lack of trunk inclination, knee flexion at drive phase, and lack of pulling handle to abdomen [17]. This study emphasizes that 1) good rowing technique is classified by consistency to some extent, and 2) requires the handle to be pull to the abdomen. Therefore, when conducting experiments, it will be essential to monitor rowing consistency and recommend that individuals pull the handle to their abdomen.

2.2.1 Symmetric Force Production & Leg Length

Even within the same participant, Greene (2009) showed that variation in the shank length to thigh length ratio between each leg can result in altered timing and force production [18]. For example, individuals with shorter shank length had earlier power generation relative to those with a longer shank [18]. Furthermore, individuals with shorter shank length had greater lumbar extension during the drive phase and decreased rotation of the pelvic segment to compensate, but ultimately had similar performance as longer shank individuals [18]. In this study, these variations in timing did not affect overall power output. However, it highlights the idea that anthropomorphic variability can alter muscle activation timing during rowing. This is an important consideration for determining the optimal timing for administering stimulation to the legs during FES-rowing.

2.2.2 Lower Back Strain: Lumbar Extension

Greene’s (2009) study demonstrated that shorter shank length resulted in earlier power generation due to greater lumbar extension [18]. However, greater lumbar extension and knee flexion angles during rowing puts individuals at a greater risk of injury [18]. Therefore, even though 14 elite rowers have demonstrated variations in rowing technique to maximize force production, there may be trade-offs resulting in overuse injury. In a population involving individuals with spinal cord injury, it is important to prioritize safety over force output or
performance. The modified rowing ergometer used in this thesis study is equipped with a back rest and shoulder straps to prevent excessive lumbar extension and flexion.

2.2.3 Sequential muscle activation patterns (Legs, Trunk, Arms):

Sequential muscle activation is an important consideration during rowing. There are variations in upper and lower limb coordination between AB and SCI FES-rowers. Firstly, AB rowers initiated the stroke with arms and legs concurrently [5]. Accurately timed knee extension allows the seat to move posteriorly followed by the handle pull, allowing for a synchronization of the force generation [5]. Therefore, AB rowers demonstrated greater oxygen consumption, and double the mechanical efficiency (work over oxygen consumption) [5].

However, individuals with incomplete SCI are more reliant on their arms for force generation. In individuals with SCI, the drive phase was incorrectly initiated by the arms and the knee extension occurred afterwards as shown in the figure below [5]. Since the knee extension was delayed, the seat was in the anterior position for a longer duration of the stroke [5]. SCI individuals also had low foot force, which was approximately 10% of their body weight, regardless of stimulation intensity used and a faster stroke rate with a shorter drive phase [5]. Thus, they had lower oxygen consumption and less external work, which limits the potential benefits [5]. Importantly, the delayed knee extension (i.e. improper technique) may result in buckling of the knees during handle pull which would impair the individual’s ability to complete the drive phase during the rowing exercise. When developing the coaching system, it will be imperative to monitor seat position and encourage individuals to focus on initiating the drive phase with the legs to prevent the knees from buckling.
Disparities in Leg Muscle Activation During Rowing

The rowing exercise is a complex movement that demands full body coordination. Previous studies have investigated the muscle activation patterns during rowing in order to understand the timing of muscle activation, investigate fatigue, or compare various rowing ergometers. Electromyography and kinematic studies on standard rowing ergometers for use in able-bodied individuals have been shown that specific regions of the spine (lumbar L3 - sacral S1) are more vulnerable to soft tissue injury since this region shows the most movement during the rowing action. Our modified rowing ergometer has a back rest with shoulder straps, shank support, a spring for facilitating the recovery phase, and minor modifications to reduce movement of the spine in individuals with spinal cord injury. These modifications may result in alterations in the leg muscle activation patterns. Even various brands and types of standard rowing ergometers have shown differences in thoracic and leg muscle activation patterns. Therefore, it is important to consider the influence of the modified rowing ergometer on leg muscle activation patterns.

Firstly, in order to determine the optimal timing of administering stimulation to the legs, it would be useful to explore the leg muscle activation patterns in able-bodied individuals. The leg muscle activation patterns on standard ergometers (slide-based and stationary ergometers) have shown that the vastus lateralis muscles were active during the drive phase. One interesting feature

Figure 1. Handle and foot force distributions between AB and SCI throughout one stroke [5]. Able-bodied individuals initiate the rowing stroke with their legs (peak of dotted line), while leg forces are delayed in individuals with spinal cord injury [5]. Cited and modified from [5].

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was the early activation of the vastus lateralis muscle in the stationary rowing ergometer. As shown in the figure below during one stroke, the vastus lateralis is mainly active during the drive phase (initial 50% of the rowing stroke), and silent in part of the recovery phase (remaining 50% of the stroke stroke). The vastus lateralis onset occurs earlier than expected (i.e. at the end of the recovery phase) in the stationary rowing ergometer (open circle) set up, but not the slide-based ergometer (closed circle)[20]. On the other hand, FES-rowing articles instruct individuals to administer stimulation to the legs arbitrarily at the end of the recovery phase, when the seat is near the anterior most position. This highlights the importance of investigating the leg muscle activation patterns in our modified rowing ergometer in order to instruct users on the optimal timing of administering stimulation to their legs.

![Graph showing Vastus lateralis muscle activation](image)

**Figure 2. Vastus lateralis muscle activation after normalization and averaging from 13 individuals during the drive (left) and recovery (right) phase. The x-axis is percent of rowing stroke, and y-axis is the percent of maximal EMG activity. Cited and modified from [20].**

Leg muscle activation patterns can also be influenced by rhythmic auditory cues and controlled stroke rates.

### 2.3.1 Effect of Auditory Cues

Rhythmic auditory cues (e.g., metronome sounds) have been implemented during rehabilitation exercise by aligning an action to the sound [22]. These cues are delivered at equally spaced temporal intervals [22]. Functional magnetic resonance studies have shown that these rhythmic auditory cues increase activation in the basal ganglia, supplementary motor areas, dorsal premotor areas, and the cerebellum indicating connectivity between auditory and motor areas.
A meta-analysis also showed that auditory cues had higher basal ganglia activation in a finger tapping task relative to visual or no cued tapping tasks [22], [26]. Furthermore, auditory cues have been shown to reduce movement variability in individuals with Parkinson’s disease (who have degradation of dopamine neurons in the substantia nigra) [27], [28]. Individuals with spinal cord injury demonstrated issues with intralimb coordination shown by altered hip and knee angles during gait relative to able-bodied individuals [29]. The sensory impairment associated with spinal cord injury can also impair motor behavior and inter-limb coordination [30], [31]. Given the connectivity between auditory and motor areas, the addition of an auditory cue may help with timing and/or movement coordination during rowing.

### 2.3.2 Effect of controlled stroke rate

There is a discrepancy regarding recommended stroke rates for FES-rowing. Stroke rate is important because it may influence the power generated and technique during rowing. In rowing, increases in stroke rate occur by reducing the time spent in the recovery phase [32]. In other sports, such as swimming, if stroke continues to increase beyond an optimal level, the swimmer’s velocity begins to decrease [32]. One study has shown that increasing stroke rate generally resulted in increasing work output in sub-elite athletes [32]. However, they only tested stroke rates of up to 40 strokes per minute [32]. Discussions with elite rowers recommended relatively slow cadences of around 24 strokes/min. During preliminary testing on our modified rowing ergometer in non-rowers, we observed that power output was relatively subjective to the individual’s level of effort; it was possible for individuals to have greater work output during slower stroke rates, or lower work output during faster stroke rates. This may also be due to deterioration of rowing technique at faster stroke rates in non-rowers. Our modified FES-rowing ergometer has shoulder straps and back rest which restricts lumbar flexion, shortening the stroke length [4]. With a shorter stroke length, individuals naturally row at faster stroke rates. To support this claim, one study has shown that there are discrepancies in the preferred stroke rate between able-bodied individuals and individuals with spinal cord injury [5]. Higher stroke rates and variability of stroke rates were observed in FES-rowing relative to able-bodied rowing (FES-rowing: 36 ± 6; AB rowing: 25 ± 3, p < 0.01) [5]. Therefore, before the development of the coaching system, it is important to investigate the effect of controlled versus preferred stroke rates in order to determine if this will influence button press timing.
2.4 FES During Rowing

The rowing exercise is a complex motion involving coordination of both upper and lower limbs. Wheeler (2002) split the rowing stroke into three phases: the drive phase, the pull phase and the return phase [33]. The drive phase consists of knee extension. Next, the pull phase consists of handle pull by the arms, while the legs remain maximally extended. In some articles, the pull phase is considered part of the end of the drive phase. Finally, the return phase consists of pushing the arms forward, and flexing the knee to return to the anterior most position on the rowing ergometer [33].

In order to produce this sequence of movements using functional electrical stimulation: 1) The quadriceps are stimulated for knee extension to signal the end of the return phase and initiate the drive phase. 2) The quadriceps are continually stimulated through-out the drive phase, and during the pull phase. 3) Finally, the hamstrings are stimulated for knee flexion during the return phase. The return phase is also assisted by a spring in this FES-system [33]. Next, it is important to consider the parameters of stimulation during FES-rowing.

A literature review of 28 FES-rowing articles was conducted, and the most commonly used stimulation parameters of FES-rowing studies are summarized in Figure 3 to determine what stimulation parameters should be used during FES-rowing. Firstly, the four-channel stimulator was preferred over the two-channel stimulator. The four-channel stimulator targeted the quadriceps during knee extension and the hamstrings during knee flexion. The stimulation was self-administered through a manual switch (or switches) found on the handle. The quadriceps were stimulated for extension when the button was pressed, and the hamstrings were stimulated for flexion when the button was released. In some cases, stimulation to the hamstrings and quadriceps were controlled by separate switches. Figure 3 demonstrates that the most commonly used stimulation parameters for FES-rowing were a pulse width of 450 us, a frequency of 40 or 50 Hz, maximal stimulation intensity of 115 mA, and a four-channel stimulator targeting both quadriceps and hamstrings muscles.
Figure 3: A collection of stimulation parameters from 28 FES-rowing articles demonstrating the stimulator used, target muscles, frequency chosen, maximum intensity level and pulse width. A pulse width of 450 us, a frequency of 40 Hz or 50 Hz, and a maximum intensity up to 115 mA was most commonly used as a threshold when the muscle fatigued.

There are a variety of methods used to administer stimulation including manual stimulators (push buttons) and automatic stimulators (PD, PID controllers, fuzzy logic controllers) [33]–[36]. However, most studies resorted to manual self-administration of FES for several reasons. The primary reason is preference and comfort: the participant would be able to stop the stimulation immediately or choose not to stimulate if they so wished[4][37]. Also, they would be able to selectively control the duration of stimulation and minimize the electrical stimulation needed[4][37]. This is important because automatic stimulation systems administer FES for the
whole duration of the drive phase, which is unnecessary discomfort, results in quicker muscle fatigue and does not feel natural for the participants[37].

2.5 Coaching and Feedback Systems for Rowing

Coaching systems have used auditory or visual feedback systems to hasten learning in complex motor exercises. The optimal response to feedback is based on modality of feedback provided and the frequency/amount of feedback (i.e. user may become overwhelmed).

2.5.1 Auditory vs. Visual Feedback

Visual feedback has been shown to aid learning of complex tasks during the early phases of motor skill acquisition. Auditory feedback can also provide a lot of information to the user: In visually impaired elite para-rowers, acoustic feedback on boat speed was shown to improve boat speed and acceleration [38]. In a more complex method of auditory feedback, auditory movement sonification involves matching the movement error to the pitch or volume of the auditory feedback (i.e. the greater the error, the louder the auditory feedback). This form of auditory sonification was developed for a rowing-type movement, but its usefulness has not been explored [38]. These studies show that auditory feedback can improve rowing performance and communicate a plethora of information.

Furthermore, the combination of different modalities of feedback may improve motor learning. For example, Sigrist et al., (2014) compared multimodal types of feedback to assess whether they improved learning [39]. Specifically, they compared audiovisual feedback, visuo-haptic feedback, and visual feedback only and demonstrated that all groups were able to decrease movement error, but audiovisual feedback significantly improved learning relative to the other modalities [39].

Therefore, integrating both visual and auditory feedback into the coaching system for administering FES to the legs may accelerate learning.

2.5.2 Frequency of feedback

Anderson et al. (2005) and Bardy et al. (2013) examined real-time visual feedback interventions for teaching and improving performance in novice rowers[40][41]. Anderson et al. (2005)
explored three different frequencies of visual feedback interventions (detailed real-time feedback, summary feedback and no feedback) to investigate their effects on rowing performance and consistency[40]. They showed that detailed real-time feedback improved rowing performance and consistency relative to summary feedback and no feedback[40]. Furthermore, summary feedback was shown to improve rowing consistency relative to no feedback[40]. However, there were no significant differences in rowing performance between summary feedback and no feedback[40]. Bardy et al. (2013) investigated whether virtual reality technology could teach energy-management strategies to novice rowers[41]. The virtual reality coaching group was compared to a control group that underwent a generic coaching procedure instructed by an elite rower[41]. The virtual reality coaching group showed an increase in rowing performance relative to the control group[41]. These studies highlight that 1) detailed real time feedback can improve performance relative to summary and no feedback, 2) teaching new skills to novice rowers through game-related feedback systems may accelerate learning. By applying the findings from these studies: developing a real-time visual feedback coaching system to instruct novice rowers on the optimal timing of administering FES to their legs may accelerate learning (i.e. reduce the number of sessions required for individuals to learn FES timing).

In many FES-rowing studies, individuals with spinal cord injury often take many sessions to learn the optimal timing of administering stimulation. Reducing the number of sessions required for individuals to learn FES timing is important because accelerating their learning of the rowing exercise will allow them to train at the higher intensities necessary for cardiovascular benefits. Improving the level of fitness and reducing cardiovascular risk in individuals with spinal cord injury can have profound implications in their quality of life and autonomy.

2.6 Summary

It is important for new rowers to learn proper timing of leg muscle activation to perform the exercise correctly, maximize musculoskeletal and cardiovascular benefits and to prevent injuries while muscle patterns are still being developed. However, previous groups stimulate the legs arbitrarily and have not developed a target range to guide the timing of leg extension during FES-rowing. Previous groups developed and tested automatic controllers (PD, PID, fuzzy logic controllers) to administer FES based on the anterior most seat position. However, many groups
reverted to manual stimulators through a push button due to patient preference and in order to maximize the voluntary contribution toward the exercise [4], [34]–[36]. Furthermore, maximizing voluntary contribution using a manual push button is potentially superior to motor based systems because it may promote neural plasticity and accelerate learning of complex motor behaviors.

While Draghici (2017) studied two-dimensional rowing biomechanics of rowing using string potentiometers to estimate the handle and seat position [5], it is necessary to investigate the leg muscle activation patterns relative the seat position in order to determine the optimal timing of administering FES to the legs. Optimal timing of leg muscle activation during the rowing exercise is essential for performing the drive phase of the rowing stroke: the drive of the legs should be linked to the subsequent handle pull[42]. Optimal timing of leg muscle activation may also maximize force output, counteract muscle atrophy and improve cardiovascular function [5], [13]. To elaborate, AB rowers perform the rowing action in a coordinated manner by extending the legs and pulling the handle shortly after [5]. In contrast, SCI rowers initiate the rowing action with their arms, followed by a brief delay before knee extension using FES [5]. Other studies investigating rowing technique and communication with elite rowers has also highlighted the importance of initiating the drive phase with knee extension first, followed by handle pull to prevent the seat from moving anteriorly during the drive phase, in the case of weaker leg muscles. These differences in the biomechanics of rowing result in decreased work output and prevent individuals with SCI from performing the drive phase of the rowing exercise [5], [42].

The rowing action is a complex movement that requires proper technique to perform the movement correctly, maximize force production and enhance the cardiovascular and musculoskeletal benefits, while preventing injuries. Given that FES-rowing has a slow learning curve, it often takes several sessions (approximately 13±7) for an SCI individual to become accustomed to the movement and perform at a level with sufficient aerobic demand for reducing cardiovascular risks [2]–[8]. Furthermore, implementing a coaching system with multimodal real-time feedback may accelerate learning of button press timing. Therefore, individuals need to be properly instructed on the timing of a manual switch initiating electrical stimulation to the legs to perform the rowing movement effectively [5], [42].
3 Research Objectives & Hypothesis

Objective: The aim of this thesis study was to design a coaching system to instruct users on the optimal timing of FES-induced knee extension.

Specific Aims: The specific aims of this study were to: (1) modify a rowing ergometer to enable individuals with spinal cord injury to perform FES-rowing; (2) determine the optimal FES timing based on the timing of leg muscle activation relative to kinematic events in able-bodied individuals; (3) develop a coaching system to instruct users on the optimal timing of administering FES to their legs using a manual switch; (4) test the coaching system to assess whether it improves timing of button press with and without the coaching system’s guidance in 7 able-bodied individuals.

Hypothesis 1: The administration of FES to the legs should be directly correlated to seat position. Knee extension should be induced by FES before the seat is at the anterior-most position. Accordingly, knee flexion should be induced by FES when the seat is at the posterior-most position.

Hypothesis 2: The coaching system will help improve the timing of manual button press that administers FES to the legs to allow the user to effectively produce the rowing stroke.
4 Hardware: Development of FES-Rowing System

The FES-rowing system was developed to enable individuals with spinal cord injury to perform the rowing exercise with FES[4]. FES-rowing required these modifications to a standard rowing ergometer (Concept 2, Inc., Morrisville, Vermont, USA) because there were several challenges associated with FES-rowing that are not present in regular rowing. For example, SCI individuals require back support to stabilize their trunk and prevent lumbar flexion and shank support to alleviate the large mediolateral torques observed during FES-rowing [4]. Furthermore, the rowing ergometer was instrumented with sensors to enable data collection.

![Modified Rowing Ergometer with Sensors](image)

**Figure 4. Modified Rowing Ergometer with Sensors.** The modified rowing ergometer for individuals with SCI includes a back rest and shank support. This was instrumented with position and force sensors for data collection. Position sensors were used to determine the handle and seat position. Force sensors were used to determine the handle and foot forces.

4.1 Custom footplates, back and shank support

We installed back and shank support obtained from Dr. Brian Andrews, Oxford University. We added custom foot plates to precisely measure the force applied by legs onto the rowing machine.
The custom foot plates were initially designed and built by Travis Van Belle et al as his undergraduate summer research[43]. Each footplate consists of one shear force and one normal force sensor (GS1240-250 and SML-300, Interface Advanced Force Measurement Durham Instruments, Arizona, USA).

![Footplate Diagram]

**Figure 5. Custom footplates.** The first picture displays the outer casing of the footplate enclosing the sensors. The second picture displays the normal force sensor (GS1240-250 Kg-SS), and the final picture displays the setup for the shear force sensor (SML-300; Interface). Cited and modified from [43].

A few modifications were made to the custom-built footplates. Firstly, the normal force sensor was attached directly to the metal base, which restricted the necessary deflection of the load cell. This impaired its ability to measure the heavy loads expected from the legs during the rowing exercise. Therefore, 1) washers were added above and below the sensor to ensure the sensor had enough space to deflect for heavier loads and 2) the coarse and fine gain settings were adjusted in the amplifier.

### 4.2 Mounting Position Sensors

The position sensors were mounted in order to accurately capture the distance travelled by the handle and seat during rowing. For handle position, the string potentiometer (Measurement Specialties, Durham Instruments, Arizona, USA) was mounted near the flywheel, so the string was parallel to the handle chain[4]. For seat position, the string potentiometer (Measurement Specialties, Durham Instruments, Arizona, USA) was mounted at the posterior end of the ergometer and attached to the back of the seat using a small hook[4].

![String Potentiometer Diagram]
Figure 6. String Potentiometer Attachment. The string potentiometer was mounted at the posterior end of the rowing ergometer. The string was attached to a hook and flat metal piece that was screwed in to fit snugly underneath the seat.
4.3 Force Sensor Calibration

The rowing ergometer was custom mounted with force and position sensors that had to be calibrated to ensure proper functioning. The force sensors convert a compressive or tension force into a proportional electrical signal.

![Calibration results from sensors in custom built footplates: 2 normal force (top and bottom left) and 2 shear force sensors (top and bottom right). Each of these plots show the linear relationship between voltage and force.](image)

In each footplate, there was 1 normal force and 1 shear force sensor. The sensors output a voltage reading when they experience force. Therefore, in order to calibrate them, weights (25, 50, 75, 100 lbs) were placed on the surface and the voltage readings were recorded. The equations that define the relationship between the voltage and weight can be used to convert the voltage reading back into a force reading. All four sensors demonstrated a linear relationship between voltage and force.
The following equations were used to calculate the force exerted based on the output voltage from the sensors.

Shear Force Sensor Equations:

\[
\text{Force} = \text{Voltage} \times \left( (-20 \times 4.44822) / (2^{16} \times 0.0415) \right)
\]

\[
\text{Force} = \text{Voltage} \times \left( (-20 \times 4.44822) / (2^{16} \times 0.0411) \right)
\]

Normal Force Sensor Equations:

\[
\text{Force} = \text{Voltage} \times \left( (-20 \times 4.44822) / (2^{16} \times 0.011) \right)
\]

\[
\text{Force} = \text{Voltage} \times \left( (-20 \times 4.44822) / (2^{16} \times 0.0417) \right)
\]

### 4.4 Position Sensor Calibration

The position sensors converted the linear position into a proportional electrical signal. The cable was spring loaded to maintain tension and prevent sagging or deflection.

In order to calibrate the position sensors, the voltage reading was compared to the position determined using a motion capture system (six infrared cameras: Rapter-E, Motion Analysis Corp, Santa Rosa, USA). The position sensors in this set up were used to measure handle and seat position during rowing.
Figure 8. Calibration of string potentiometer sensors using motion capture. Voltage reading converted to distance (red) is overlaid with the distance determined by motion capture (blue).

As shown in Figure 8, after determining the coefficient to convert the voltage (V) into position (mm), this derived position value (red line) was compared to the actual string length (blue line) as determined by motion capture. As the seat was moved back and forth, we can see slight deviations at the extremities. Root mean square error of 38 mm.

The abrupt change in direction during rowing could be responsible for the deviations seen at the extremities and during the initial 1-2 second period. The motion capture markers were placed at 1) the tip of the string potentiometer and 2) the contact point of the string potentiometer and the seat. Therefore, during abrupt changes in direction, the string may have some slack, increasing the actual length of the cable, while the distance between the seat and string potentiometer as determined by motion capture remains constant.

Position Sensor Equations:

\[
Handle \text{ Position (mm)} = \frac{Voltage}{15.911};
\]

\[
Seat \text{ Position (mm)} = Voltage \times \left(\frac{10}{0.0769}\right)\times \left(\frac{20}{2^{16}}\right);
\]
4.5 Other Modifications on FES-rowing Ergometer

Other rowing ergometers have been developed for individuals with spinal cord injury to address some issues seen during FES-rowing[4]. For individuals with spinal cord injury, one limiting factor is that the knee flexion resulting from hamstring stimulation may be insufficient to complete the reset / recovery phase of the rowing stroke [4]. One study reported that they had a physiotherapist present to physically push the rower back to the front after each stroke to help them reset [4]. Techniques used to alleviate this issue include 1) administering stimulation to the hamstrings, 2) mounting a spring at the posterior position of the ergometer, and 3) raising the incline of the posterior end of the rowing ergometer[4].

The most common practice involves administering stimulation to the hamstrings coupled with the spring at the posterior position[4]. Proper rowing technique involves the drive phase and the recovery phase. The individual begins the rowing stroke with their elbows extended and knees flexed. Firstly, the drive phase consists of 1) knee extension to move the seat backwards, 2) followed by elbow flexion for handle pull. Following this motion, the recovery phase consists of 1) elbow extension/reset handle and 2) knee flexion for returning to the initial starting position[15]. The drawback of this posterior spring-mounted set-up means that once knee extension is complete, the rower must slightly resist the spring to complete the handle pull, before being pushed forward to reset. Furthermore, the spring is mounted on the monorail which limits the maximum range that the seat can travel[4].

One example involves an inclined system to raise the posterior end (the monorail) of the rowing ergometer by around 4 degrees [4]. The raised posterior end uses gravity to its advantage, as the rower will slide down and forward to reset / return to the anterior most position. Drawbacks to this system include a higher necessary knee extension force to push upwards against the incline during the drive phase. This system does not require a spring.

When comparing both rowing ergometer-setups, the spring set up is more suitable for naïve FES-rowers since they likely would have insufficient leg strength to drive against the inclined system. Furthermore, the spring can be adjusted for each individual to maximize the seat’s range of motion and account for anthropomorphic differences in thigh and shank length. For the following
experiments, the rowing ergometer was used with a spring and hamstring stimulation to aid with the recovery phase as shown in Figure 9.

Figure 9. Modified rowing ergometer with back support, spring and force and position sensors.
5 Experiments

5.1 Introduction

Previous studies have investigated the muscle activation patterns on stationary and dynamic rowing ergometers in order to explore how the timing of muscle activation is altered by different types of ergometers [20]. EMG and kinematic studies in able-bodied individuals on standard ergometers have highlighted that the lumbar and sacral regions (L3-S1) of the spine showed higher movement during the rowing action, which may increase the risk of soft tissue injury in this area[21]. Another study showed that vastus lateralis activation for knee extension occurred earlier in the stationary ergometer relative to the dynamic ergometer [20]. The rowing ergometer for FES-rowing used in our experiment has several modifications which may influence the leg muscle activation patterns. Therefore, understanding the timing of leg muscle activation on the modified FES-rowing ergometer will help determine the optimal range of FES timing for coaching system.

One idea to improve consistency during a periodic movement is to use rhythmic auditory cues (RACs), which are sounds at regularly spaced time intervals[22]. They can be used with rehabilitation exercises by aligning an action to the sound [22]. RACs have been used in individuals with Parkinson’s disease to reduce movement variability [27][28]. Functional magnetic resonance imaging studies have shown that RACs increase activation in the basal ganglia relative to visually cued or uncued conditions during a finger tapping task[22], [25]. Individuals with spinal cord injury have also demonstrated altered motor behaviors and interlimb coordination possibly due to sensory impairment [29]–[31]. Therefore, the addition of a metronome may improve timing and rowing consistency in individuals with spinal cord injury.

When considering timing and consistency, the stroke rate selected for rowing will influence the time spent in each phase of rowing, the power generated and rowing technique[42]. In general, as with other sports, increasing the cadence or number of cycles per minute results in greater work output in athlete[32]. However, a swimming study highlighted that there are trade-offs: swimming velocity may decrease at high stroke rates beyond an optimal range [32]. The FES-rowing ergometer restricts lumbar flexion and stroke length, which results in higher than normal
stroke rates. Furthermore, able-bodied rowing has a cadence of around $25 \pm 3$ strokes per minute, while FES-rowing has a cadence around $36 \pm 6$ strokes per minute[5]. Therefore, it will be important to investigate whether controlling the stroke rate influences the timing of leg muscle activation.

Understanding the leg muscle activation timing in able-bodied individuals may help develop a coaching system that instructs users on the recommended range at which to administer electrical stimulation. Furthermore, understanding the influence of auditory cues and controlled stroke rates on the timing of button press will help us make decisions on which factors influence button press timing and tailor the coaching system accordingly.

5.2 Methodology

5.2.1 Participants

Ten ($n=10$) able-bodied individuals volunteered for the study with no history of neurological disorders. Their rowing expertise ranged from individuals with no rowing experience to one varsity rowing athlete. All were instructed in proper rowing technique prior to participation in the experiment. Participants were recruited to Lyndhurst Rehabilitation Center through posters, email and by telephone. Participants signed a consent form detailing the experimental protocol and set up. These documents were reviewed by the Research Ethics Board at Toronto Rehabilitation Institute before commencing the experiment.

This study was approved by Research Ethics Board of the University Health Network. Each participant visited Lyndhurst Rehabilitation Center once for two hours to participate in the following experiment outlined below.
5.2.2 Experimental Design

Protocol Overview:

1. Setup with electromyography on bilateral leg muscles.
   a. Resting electromyography
   b. Maximum voluntary contractions
2. Warm-up and familiarization.
3. Rowing tests:
   a. Maximal work output test: 10 hardest strokes *
   b. Preferred stroke rate test *
4. Four randomized trials of rowing with metronome and stroke rate conditions at 50% of max work output:
   a. 30 strokes/min rate, no metronome *
   b. 30 strokes/min rate, metronome *
   c. Preferred strokes/min rate, no metronome *
   d. Preferred strokes/min rate, metronome *

*Each trial lasts two minutes in duration. Participants took two to five minutes of rest in between trials. Muscle activation patterns, kinematic and kinetic data were collected as outlined in the following paragraphs.

Electromyogram (EMG) were measured using an EMG amplifier (Bagnoli, Delsys Inc., USA). Raw electromyography data were collected at 2k Hz sampling rate using a 16-bit resolution; amplitude range of 10V. After removing the mean from the raw EMG data, it was full wave rectified, and then filtered. A 2\textsuperscript{nd} order 10 Hz low pass Butterworth filter was selected to obtain the envelope of the EMG signal given that one movement cycle is approximately 2 seconds or 0.5 Hz. Zero-phase digital filtering was conducted by using the filtfilt function in MATLAB to process the data in the forward and reverse directions as this is important to account for delays when calculating muscle onset timing. Electrodes were placed on bilateral leg muscles (rectus femoris, vastus lateralis and medialis, biceps femoris, medial and lateral gastrocnemius, soleus and tibialis anterior muscles). Ground electrodes were placed on the patella. The electrodes were taped and wrapped to the leg to prevent them from shifting during the rowing trials.
Previous studies often only collected muscle activation data from one side since they assumed symmetric leg muscle activation patterns during rowing exercise. This is a viable assumption if individuals are athletes or have previous rowing experience as they are likely to exhibit high symmetry. Preliminary testing on the rowing ergometer with able-bodied novice rowers showed that the electromyography recordings did not necessarily display a high degree of symmetry in all subjects as determined by Pearson’s Correlation. Therefore, in the analyses, the electromyography from both legs was averaged to account for slight asymmetry.

While the participant was seated comfortable on a chair, EMG were collected during resting in order to establish a baseline. EMG during the maximum voluntary contraction was also determined for each muscle as follows. The participant was seated, and maximally contracted the corresponding muscle for 5 seconds, while the researcher manually resisted against the movement. 10% of the maximal EMG activity during maximal voluntary contraction was used to determine the muscle activation onset during rowing for each stroke.

All rowing trials were conducted on a rowing ergometer (Concept 2, Inc., Morrisville, Vermont, USA) with back support, shoulder straps, shank straps, and mounted force and position sensors.

As mentioned in the hardware section, the back support and shoulder straps acted to prevent hyperextension and flexion of the lumbar spine respectively.

The shank straps exist to ensure that leg muscle force is directed primarily in the sagittal plane. This feature is for individuals with spinal cord injury to ensure that FES induced knee extension occurs in the sagittal plane. When this experiment was conducted in able-bodied individuals, the shank straps, shoulder straps and back rest were used in order to represent the leg muscle activation patterns expected during FES-rowing.

The force data was collected from custom built foot plates which contain both normal and shear force sensors (Interface Advanced Force Measurement Durham Instruments, Arizona, USA). The seat and handle position data were collected from string potentiometers (Measurement Specialties, Durham Instruments, Arizona, USA) mounted to FES-rowing system to measure handle and seat position. The analog signals from the sensors were collected using a data acquisition system with a sampling frequency of 2000 Hz (PowerLab 16SP, Cortex and LabChart, ADInstruments, Inc.,

27
Colorado Springs, CO, USA). The force data was also filtered by a 10 Hz low pass filter since each movement cycle during rowing lasts around 2 seconds or 0.5 Hz.

The participant performed 5 to 10 minutes of warm-up and familiarization on the modified rowing ergometer.

Participants were instructed to perform their ten hardest strokes for the maximal power output test. The work output during these strokes were recorded using the Concept II performance monitor. For the following trials, individuals performed at around 50% of their maximum power output.

The preferred stroke rate test was conducted at 50% of maximum power output for 1 minute.

Four trials of rowing lasting two minutes each were conducted. Two trials were conducted at the individual’s preferred stroke rate, with and without the metronome. Two trials were conducted at a controlled stroke rate of 30 strokes/min, with and without the metronome. The order of the trials was randomized among participants, to account for the learning and/ or fatigue effects in novice rowers. Participants took rest for 2 minutes after each trial to minimize the influence of fatigue or the order of the trails.

The preferred stroke rate and controlled stroke rates were conducted in order to determine the effect of controlling the stroke rate. Furthermore, a controlled stroke rate at 30 strokes per minute may not be suitable for all participants. For example, anthropomorphic differences such as shorter limb length may result in smaller stroke length and thus higher stroke rates[18].

The auditory cue / metronome provided periodic sounds to help individuals keep pace at their preferred and controlled stroke rates.

5.2.3 Data Analysis

5.2.3.1 Leg Muscle Activation Patterns

After obtaining the raw EMG signal, the mean was removed, full wave rectified, and filtered. EMG signal was full wave rectified to remove any negative values. A 2nd order low pass Butterworth filter at 10 Hz was applied to the EMG signal to obtain an envelope since each
movement cycle was two seconds or 0.5 Hz. Next, in order to normalize each rowing stroke from 0% to 100% of the movement, each stroke cycle was segmented based on the anterior-most seat position. Each stroke cycle was down sampled from roughly 3000 points to 1000 points in order to normalize. The ensemble average and standard deviations were calculated for all aforementioned leg muscles for each subject. The first and last five strokes were excluded from the analyses due to anomalies in the data since individuals were adjusting their positions or catching their rhythm.

5.2.3.2 Electromyography Onset Threshold

In order to determine the threshold value for EMG onset during rowing, 10% of EMG amplitude during maximum voluntary contraction was calculated for each muscle of each participant. Since EMG was collected bilaterally, the average EMG signal and onset threshold of the left and right muscles were used.

5.2.3.3 Temporal Difference Calculation Between Kinematic Events (Seat Position & Handle Position) and Vastus Lateralis Onset:

The kinematic events of interest include the 1) anterior handle position, 2) posterior handle position, 3) anterior seat position, and 4) posterior seat position. The timing differences between the kinematic events (anterior and posterior, seat and handle positions) relative to VL muscle onset were calculated.

The quadriceps muscles (rectus femoris RF, vastus lateralis RF, vastus medialis VM) are active for knee extension during the drive phase. VL was selected as the muscle of interest because it is solely used for knee extension while the RF occasionally had two peaks since it is active for both knee extension and hip flexion[44]. Furthermore, one article compared the envelope of the RF and VL muscles. The RF peak corresponding to knee extension during the drive phase is only apparent at high exercise intensities of around 95% of VO2 peak. However, the VL EMG signal had only one peak due to its singular role for knee extension. Secondly, the shape of the VL EMG envelope and number of peaks were not influenced by exercised intensity[44]. Given that the rectus femoris onset was harder to determine due to the periodic occurrence of the double
peak (dual role as knee extension and hip flexor), the vastus lateralis onset time was chosen to
guide the timing of the manual button for administering FES [45].

For each two-minute rowing trial, the sensor and EMG data were segmented based on stroke
cycle as defined by the anterior most seat position peak. The anterior most seat position peak
signals the beginning of the stroke’s drive phase. 30 consecutive strokes from the middle of the
trial was used for the analyses. Muscle onset was marked as the part of the EMG envelope that
exceeded 10% of the EMG value from maximum voluntary contraction.

The kinematic events of interest (i.e. peaks of seat and handle position) were determined using
the findpeaks function in MATLAB. The time difference between the muscle onset and the
kinematic events of interest were calculated for each stroke.

5.2.3.4 Spatial Difference Calculation Between Kinematic Events and
Vastus Lateralis Onset

The seat and/or handle position during VL onset can be used to guide the application of electrical
stimulation. For each stroke, the seat and handle position during the onset of VL EMG was
calculated relative to the anterior most seat and handle position (zero). Based on 30 consecutive
strokes, the mean and S.D. of seat and handle positions during VL onset for each trial was
calculated.

5.3 Results:

5.3.1 Leg Muscles Activation Patterns

Figure 10 demonstrates that the quadriceps muscles (rectus femoris, vastus lateralis, vastus
medialis) were active during the drive phase (first 50% of the rowing stroke) for knee extension.
The hamstring muscles (biceps femoris) were active for knee flexion during the recovery phase.
The tibialis anterior was active for ankle dorsiflexion during the recovery phase (last 50% of the
rowing stroke). The remaining shank muscles appear to be active during the drive phase in this
participant, but displayed high standard deviation which may be attributed to our modified
rowing ergometer set up as the shanks are strapped in. Foot force peaks earlier than handle force
demonstrating expected sequential muscle activation patterns (initiate with legs and follow up
with arms).
Figure 10. Leg muscle EMG patterns, handle and seat position, and foot and handle force in a typical rower during 1 stroke cycle. Solid line indicates average, and dotted lines indicate standard deviation after 30 strokes were normalized and averaged. Electromyograms from Rectus Femoris (RF), Vastus Lateralis (VL), Vastus Medialis (VM), Biceps Femoris (BF), Soleus (SOL), Medial Gastroc. (MG), Lateral Gastroc. (LG), Tibialis Anterior (TA). Kinematic and kinetic data such as Seat Position (SP), Handle Position (HP), Foot Force (FF), Handle Force (HF).
5.3.2 Determining Optimal Timing of Stimulation

There were two types of analyses or calculations conducted to determine the optimal timing of stimulation: 1) Temporal Difference Analyses 2) Spatial Difference Analyses.

5.3.2.1 Temporal Difference Between Kinematic Events (Seat Position and Handle Position) and Vastus Lateralis Onset.

First, we looked at the timing difference between each kinematic event and leg muscle activation onset. The kinematic events of interest include the 1) posterior-most seat position, 2) posterior-most handle position, 3) anterior-most seat position, and 4) anterior-most handle position based on each participant’s range of motion. The average timing difference between each kinematic event and the VL onset for each stroke was calculated for each trial per participant (i.e. one timing difference value representing the average of 30 consecutive strokes was calculated for each trial per participant). Then, the average timing difference across 10 participants are displayed in Figure 11.

Figure 11. The timing difference (seconds) between various kinematic events (posterior seat position, posterior handle position, anterior handle and seat position) and VL onset across 4 different trials. Dark orange bar indicates preferred stroke rate, with a metronome. Light orange bar indicates preferred stroke rate without a metronome. Dark green bar indicates 30 stroke rate with a metronome. Light green bar indicates 30 stroke rate without a metronome.
First, the timing difference between seat position (anterior and posterior) and VL onset in 10 subjects across 4 trials is shown in the two left-most graphs of Figure 11. The average time difference between anterior seat position and VL onset in 30 stroke rate, no metronome trial was 1.692 ± 0.176 seconds, 30 stroke rate, metronome trial was 1.736 ± 0.265 seconds, preferred stroke rate, no metronome trial was 1.662 ± 0.145 seconds, preferred stroke rate, metronome trial was 1.553 ± 0.262 seconds. The mean across all 4 trials was 1.661 ± 0.078 seconds.

The average time difference between posterior seat position and VL onset in 30 stroke rate, no metronome trial was 0.901 ± 0.159 seconds, 30 stroke rate, metronome trial was 0.917 ± 0.189 seconds, preferred stroke rate, no metronome trial was 0.857 ± 0.114 seconds, preferred stroke rate, metronome trial was 0.769 ± 0.225 seconds. The mean across all 4 trials was 0.861 ± 0.066 seconds.

Next, the timing difference between handle position (anterior and posterior) and VL onset in 10 subjects across 4 trials is shown in the two right-most graphs of Figure 10. The average time difference between anterior handle position and VL onset in 30 stroke rate, no metronome trial was 1.715 ± 0.147 seconds, 30 stroke rate, metronome trial was 1.773 ± 0.162 seconds, preferred stroke rate, no metronome trial was 1.675 ± 0.071 seconds, preferred stroke rate, metronome trial was 1.638 ± 0.185 seconds. The mean across all 4 trials was 1.700 ± 0.058 seconds.

The average time difference between posterior handle position and VL onset in 30 stroke rate, no metronome trial was 0.807 ± 0.158 seconds, 30 stroke rate, metronome trial was 0.843 ± 0.180 seconds, preferred stroke rate, no metronome trial was 0.794 ± 0.111 seconds, preferred stroke rate, metronome trial was 0.685 ± 0.223 seconds. The mean across all 4 trials was 0.782 ± 0.069 seconds.

There were no significant differences between conditions (non-parametric, paired/repeated measures, Wilcoxon Signed Rank Test).

Figure 11 shows similar average timing differences between VL onset and anterior kinematic events (seat and handle positions: 1.661 seconds and 1.700 seconds respectively). However, when considering posterior kinematic events, the posterior seat position occurs first followed by the posterior handle position. This is because in order to perform the rowing action, the
individual extends their legs first, then the handle pull follows immediately afterwards. The average timing difference between VL onset and posterior handle position is 0.782 seconds, while the average timing difference between VL onset and posterior seat position is 0.861 seconds. These values reinforce the idea that the knee extension peaks first to drive the seat to the posterior end of the ergometer, followed by handle pull, followed by the leg muscle activation for the subsequent stroke as shown in figure 12.

Figure 12. Vastus lateralis EMG (red), seat position (green), and handle position (blue) during one sample strokes across 1.75 seconds. Vertical colored lines show EMG onset (red), the anterior kinematic events are shown by the green seat trough and blue handle trough, and the posterior kinematic events are shown by the green seat peak and blue handle peak.

The drive phase consists of knee extension (red EMG peak) to drive the seat backwards (green seat position peak) followed by handle pull (blue handle position peak). At the end of the drive phase (around <50% of the rowing stroke), the seat and handle are at the posterior-most position of the rowing ergometer. At this point, the rower begins the recovery phase and returns to the anterior-most position of the rowing stroke, and their leg muscles prepare to activate for an explosive drive and repeat the rowing cycle.
It is important to note that there were large individual differences in timing differences likely due
to anthropometric differences such as shank to thigh ratios [18] and relative time spent in drive
versus recovery [44].

The addition of a metronome increased the standard deviation in all cases. However, comparison
using Wilcoxon Signed Rank test revealed no significant differences in the standard deviation
between trials.

Exercising at a controlled stroke rate (30 strokes/min) resulted in a slightly higher timing
difference relative preferred stroke rate. However, these differences were not significant as
determined by the Wilcoxon Signed Rank Test (non-parametric, paired/repeated measures).

Furthermore, the standard deviations of the timing difference between the kinematic events and
VL onsets did not have a significant difference. These standard deviations of timing differences
between kinematic events and VL onset are likely due to variations in VL onsets of up to 10% of
the movement cycle, that were also seen in other studies involving cyclical movements such as
walking and cycling [46]. This is due to small changes in the way the body performs the periodic
movement or slight alterations in the intensity and pattern of muscle activations during multiple
strokes [46].

These results show that the addition of a metronome, or controlling the stroke rate, or the
interaction (both metronome and controlled stroke rate) did not significantly affect the timing
difference between kinematic events and VL onset.
5.3.2.2 Spatial Difference Between Kinematic Events and VL Onset

Figure 13. The average seat and handle position during VL onset from 10 participants are shown in the graphs above for each trial type.

30 consecutive strokes were used from each trial for each participant. The spatial difference between the mean seat/handle position during VL onset relative to the anterior most seat/handle position from 10 participants are shown above for each trial. Given that the anterior most seat/handle position was set to zero for each stroke, taking the average seat/handle position during VL onset is equivalent.

The average seat position during VL onset in the 30 stroke rate, no metronome trial was 80.8 ± 38.2 mm, 30 stroke rate, metronome trial was 66.7 ± 38.3 mm, preferred stroke rate, no metronome trial was 112 ± 30.6 mm, preferred stroke rate, metronome trial was 112 ± 38.5 mm. The mean across all 4 trials was 93.2 ± 23.2 mm.

The average handle position during VL onset in the 30 stroke rate, no metronome trial was 110 ± 51.0 mm, 30 stroke rate, metronome trial was 80.1 ± 48.1 mm, preferred stroke rate, no metronome trial was 137 ± 38.1 mm, preferred stroke rate, metronome trial was 136.7 ± 48.3 mm. The mean across all 4 trials was 116 ± 27.2 mm.
There was no significant difference between conditions based on Wilcoxon Sign Rank Test (non-parametric, paired/repeated measures). Therefore, regardless of the condition chosen for the rowing exercise (i.e. controlled stroke rates or metronome), there is no significant difference in seat or handle position during VL onset.

5.4 Discussion:

This experiment was conducted to determine the leg muscle activation patterns during rowing in able-bodied individuals. The timing of leg muscle activations was used to develop a coaching system which guides the administration of FES to the legs in SCI individuals at the optimal time.

5.4.1 Leg Muscle Activation Patterns:

Our results demonstrated that the quadriceps were activate for knee extension during the drive phase of the rowing stroke. Biceps femoris was activate for knee flexion during the recovery phase. The tibialis anterior was active for ankle dorsiflexion during the recovery phase.

5.4.1.1 EMG activations in rowing studies: difficulty determining RF onset

In Neil et al., (2014)’s study, comparing leg muscle activation patterns on standard rowing ergometers and on-water rowing, they also demonstrated quadricep muscle activation during the drive phase[44]. However, they showed that exercise intensity influences the shape of the rectus femoris envelope [44]. The rectus femoris occasionally has two peaks because it demonstrates activity during 1) the drive phase for knee extension and 2) initiation of the recovery phase for hip flexion. Even though we are primarily interested in stimulating the rectus femoris for knee extension, this first peak is more prominent only at high exercise intensities (95% of VO2 peak) which is likely not achievable by individuals with spinal cord injury. The vastus lateralis demonstrated one peak for its role in knee extension and the shape of the EMG envelope was not largely affected by exercise intensity. The double peak in rectus femoris activity relative to the single peak in vastus lateralis muscles were also shown in a similar study [45]. On our modified rowing ergometer, we also saw the occurrence of a double peak in rectus femoris activity, while the vastus lateralis activity had one peak. Given that the rectus femoris onset was harder to determine due to the periodic occurrence of the double peak (dual role as knee extension and hip
flexor), we chose to use the vastus lateralis onset time to guide the timing of the manual button for administering FES.

5.4.1.2 Biceps Femoris EMG activation patterns

In our study, the BF was probably active for knee flexion during the recovery phase. However, some EMG rowing studies conducted on standard ergometers demonstrated BF activity in phase with VL for stabilization [44], [45]. However, 13 out of 15 FES-rowing studies chose to stimulate the BF or hamstrings for knee flexion during the recovery phase (as shown in section Stimulation Parameters), because the recovery phase is especially challenging for individuals with spinal cord injury[4]. Some studies had a researcher or physiotherapist physically pushing the individual’s seat to aid with recovery[4]. Other design methods to help the individual recover (return to the initial anterior position) includes the addition of a spring mounted at the back of the ergometer and raising the back of the ergometer at a 4-degree incline [4]. [33],[47].

Our modified ergometer has a spring mounted on the back, which would reduce the need for BF activation during the recovery phase. The inter-subject differences in BF activity patterns seen across other EMG rowing studies may be attributed to different types or modifications to the ergometers[20].

Even though the biceps femoris may be active during both phases, we chose to mirror other articles and our results from experiment 1, and stimulate the BF only for recovery since this FES-rowing system is most useful for individuals with spinal cord injury. Stimulation during both phases may contribute to early fatigue onset; stimulation during only the drive phase would not solve the problem of recovery that many FES-rowing studies have reported[4].

The tibialis anterior (TA) muscle was active for ankle dorsiflexion during the recovery phase. However, on our modified ergometer, the shank is strapped in rather than the foot. Therefore, the activation of the TA for ankle dorsiflexion on our modified rowing is smaller. The contribution of the TA is also smaller relative to larger muscle groups (quadriceps and hamstrings).
5.4.2 Temporal Differences Between Kinematic Events and VL Onset:

Our results showed that average timing differences between VL onset and anterior kinematic events were relatively similar (i.e. anterior handle and anterior seat position were similar) because the anterior seat and handle peaks occur closely in time. Furthermore, the posterior handle position has a sharper peak and occurs after posterior seat position so the timing differences between these kinematic events and VL onset were similar but not always overlapping to the same degree as shown in Figure 12.

It is important to note that timing differences across subjects was not consistent, which made it difficult to determine the optimal timing difference for administering FES. These differences in the mean timing difference of each individual may be attributed to anthropomorphic differences and drive to recovery ratios [18]. For example, one study showed that shorter individuals with smaller shank to thigh ratios had earlier power generation[18]. Furthermore, some individuals may have a long, slow recovery phase, while others prefer shorter recovery phases. Therefore, applying stimulation based on temporal differences may not be inclusive of different rowing strategies.

5.4.2.1 Standard Deviation:

A low standard deviation is important for determining the optimal timing of administering stimulation because it shows consistent rowing pattern. The standard deviations in timing differences are primarily caused by variations in EMG onset times. The EMG onset times appear at similar instants during cyclical periodic movements but variations of around 10% of the movement cycle are possible because the untrained individual does not produce a perfectly cyclic movement [46]. The timing difference from each stroke were calculated from over 30 strokes from each individual, because using only the ensemble average EMG envelope has been shown to be unreliable since it results in loss of information [46].

Previous studies have shown that individuals may have drive to recovery ratios between 0.9 and 1.7, but the optimal drive to recovery ratio was estimated to be around 1.0 [44]. Furthermore, on standard ergometers, the drive phase changes based on the stroke rate [44]. The drive phase during a slower stroke rate is around 33% of the rowing stroke, while the drive phase during a faster stroke rate was around 44% of the rowing stroke. Even though these studies were
conducted on standard ergometers, the change in drive to recovery ratio based on stroke rate reflects that individuals increase the relative time spent in the drive phase at faster stroke rates [44]. We had controlled stroke rate trials at a constant 30 strokes/min to address this problem, but the ranges of timing differences were still inconsistent across participants. This highlights the weakness of using time dependent features to guide FES-timing: it does not account for individual variability and forces individuals to complete the phase within a restricted time interval.

5.4.3 Spatial Difference Between Kinematic Events and VL Onset:

Previous literature allowed participants to arbitrarily administer FES to the legs when the seat was around the anterior most position of the rowing machine[37][4]. However, our results showed that VL activation occurs slightly earlier than the anterior most seat position. Furthermore, determining the optimal time to administer FES based on the temporal difference between the posterior peak of handle or seat position and VL muscle onset were inadequate. This is because temporal differences were highly variable across participants and forced the individual to complete the rowing phases in a restricted time interval. Furthermore, relying on temporal differences does not account for fatigue or weak muscles often seen in individuals with spinal cord injury. Therefore, we chose to administer FES based on spatial difference data rather than temporal difference data. Unlike previous FES-rowing articles, we demonstrated that VL activation occurs before the end of the recovery phase for a braking effect as shown in our results and EMG rowing studies on standard stationary ergometers [45].

The seat and handle position during VL muscle onset were at the anterior most position of the individual’s range of motion. Handle position had slightly larger variability than seat position, but handle position also has a larger range or distance of travel. The seat travels a maximum range of 600 mm during the drive phase and 600mm during the recovery phase since the rowing ergometer has a spring and a stopper which limit the range of motion. The handle travels approximately 1000 mm during the drive phase and 1000mm during the recovery phase, but this stroke length is also subjective and based on the participant’s anthropomorphic characteristics[18]. Other groups that have developed automatic controllers have used seat position or handle position or both to guide the timing of FES[36]. In our case, the target
population involves individuals with incomplete spinal cord injury who seek to improve function in their legs. Thus, we chose to focus on seat position, which is directly related to knee angle and leg function, would fall in line with their goals [35].

5.4.4 Influence of Metronome:

Our results showed that the addition of the metronome increased average standard deviation in all trials, but this was not a significant difference. This implies that the addition of a metronome did not improve movement variability (i.e. consistency of rowing pattern). Other studies showed that the addition of an auditory cue increased activation in motor areas (basal ganglia, supplementary motor areas, dorsal premotor areas, cerebellum) [22]–[24], decreased movement variability in individuals with Parkinson’s Disease [28] and increased basal ganglia activation in a finger tapping task relative to visual and no cues [26]. This is contradictory to our findings since we did not see a significant difference in the temporal or spatial differences between kinematic events and VL onset, or even in the standard deviation with the addition of a metronome.

Even though we did not see a significant difference with the addition of a metronome. Some explanations for increased variability in the metronome trials include the increased cognitive load due to the novelty aspect of metronome-rowing. The able-bodied rowers in the experiment were familiar with regular rowing but were not previously exposed to metronome-based rowing training. Therefore, the addition of the metronome may have a negative impact initially due to the increased cognitive load of coordinating their body to an auditory cue. Perhaps long-term use of the metronome may decrease timing difference variability, even if first-time exposure to the metronome increases variability. To test this, a longer interventional study comparing two groups, with and without metronome may reveal the benefits of adding the metronome.

In this regard, one study in able-bodied individuals demonstrated that long term learning benefits of auditory cues were not seen during a wrist flexion task [22]. However, when the metronome was paired with motor evoked potentials, it caused increases in cortico-motor excitability (possibly related to increased plasticity) relative to the uncued movement [48]. Therefore, even if the metronome alone did not show benefits in our results, pairing the electrical stimulation with a metronome may decrease variability and increase plasticity.
The influence of the metronome is also dependent on the type of movement performed. One study showed that the addition of a metronome did not increase brain activation during a dance step-like movement. During the uncued movement, there were increases in putamen activation due to the cognitive attention on self-pacing [49].

Previous literature shows that the metronome has differing effects in various neurological disorders as well. During gait, even though the metronome decreased movement variability in Parkinson’s Disease [28], it caused attentional deficits in individuals with Huntington’s Disease, and impairment in executive function in individuals with Alzheimer’s Disease, which is not characterized as a movement disorder [50].

Ultimately, pairing the metronome with electrical stimulation in spinal cord injured individuals may decrease variability and increase plasticity. In our results, the metronome may have increased variability in the rowing trials due to attentional or executive function impairments or increased cognitive load or due to the novelty aspect[22]. Although the metronome may be beneficial for individuals with spinal cord injury with distorted interlimb coordination [29]–[31], it may be difficult for individuals to coordinate their movement with multiple stimuli (visual feedback and auditory cues).

5.4.5 Identifying the Target Seat Position Range for Manual Button Press:

Our results show that VL activation begins slightly before the end of the recovery phase, i.e. before individuals reach the anterior most seat position. Thus, the coaching system should ensure that the VL muscle activation occurs prior to anterior most seat position, within a target range. EMG studies on standard stationary rowing ergometers also reinforce the idea that VL muscle activation should happen earlier in order to brake against the forward motion during the end of the recovery phase and stabilize the rower on the stationary ergometer [45]. Accordingly, we observed a similar pattern on our modified FES-rowing ergometer with back and shank support.

In the spatial difference results section, we showed that on our modified ergometer, the vastus lateralis activation is initiated when the seat and handle are 93.2 ± 23.2 mm and 116.2 ± 27.2 mm from the anterior most position of the individual’s range of motion based on one mean value from each of the 10 subjects. However, the standard deviation range did not encompass the wide
inter-subject variability seen during rowing. Therefore, in order to determine the target seat position range for administering FES, all the strokes were pooled and the global mean and global 1 S.D. of the seat positions during VL onset were calculated to be 96.5 mm ± 93.3 mm. This target seat position range was able to encompass the wide range of seat positions demonstrated by participants due to anthropomorphic variability. Furthermore, subsequent preliminary testing involving FES-rowing with manual button press showed that able-bodied individuals tend to button press prior to the anterior most position within this target range: 1.0 global S.D. of the global mean. Therefore, this target seat position range was selected for the subsequent development and testing of the coaching system.

6.1 Appropriate Timing for Manual Button Pressing

The coaching system was developed to instruct naïve FES-rowers on the optimal timing to press the manual push button to administer FES to the legs. Based on the results in the previous chapter, the appropriate timing of FES manual button was determined to be 96.5 ± 93.3 mm based on the global mean and standard deviation of seat positions from all the pooled strokes. This target range was implemented in the software we developed, which was written in Python (Python Software Foundation, Python Language Reference, Version 3.5).

6.2 FES-Rowing with Coaching System

The coaching system was run on a laptop (Dell Inspiron 7559, Texas, USA) and displayed in front of the individual on the modified rowing machine as shown in Figure 14. As outlined in the hardware section, there are force and position sensors (Durham Instruments, Arizona, USA) mounted on the FES-rowing ergometer. A manual push button was mounted on the handle to send a signal to the electrical stimulator (Compex Motion 2, Compex Motion SA, Switzerland) and subsequently produce a contraction in the leg muscles.

The coaching system monitors the sensor data at a sampling frequency of 200 Hz through the data acquisition system (USB-6002 Multifunction I/O Device, National Instruments, South Portland, ME, USA), and displays the seat position, button press timing, stroke rate and power output in real time. Based on this sensor data, the primary function of the coaching system is to provide real time feedback and instruct the user on the optimal time to press the manual push button based on the target seat position range determined from experiment 1. Once the user presses the button, a signal is sent to the electrical stimulator to produce a contraction via FES in the quadriceps for knee extension to perform the drive phase of the rowing stroke. Releasing the button at the end of the drive phase, stimulates the hamstring muscles for knee flexion to perform the recovery phase. This recovery phase is facilitated by the spring since the hamstring stimulation is often insufficient to result in adequate knee flexion[4].
Figure 14. Sensor data is used to provide real time feedback on the optimal timing of button press to administer FES to the legs, which sends a signal to the electrical stimulation and activates the quadriceps for knee extension during the drive phase. Releasing the button when the seat reaches the posterior end of the ergometer will cause stimulation to the hamstrings only for the recovery phase.

6.3 Feedback Modality for GUI of Coaching System

Next, we considered different types of feedback for instruction of FES timing. The integration of auditory and visual feedback has been shown to be accelerate learning in a rowing task compared to visual only and visuo-haptic feedback[39]. Therefore, the coaching system provides auditory and visual real time feedback on whether the stimulation was administered within the recommended range. A specific noise is emitted for correct button presses and a different noise is emitted for incorrect button presses. Furthermore, the score is displayed and increased during correct button press, while misses are increased during incorrect button presses.

In the coaching system interface shown in Figure 14, the blue seat transverses across the grey bar at the bottom and displays the seat position in real time. Extrinsic feedback such as displaying
the seat position, rather than solely relying on intrinsic feedback such as proprioceptive information, might be beneficial for individuals with SCI who may have lack of afferent sensory input.

In terms of visual feedback, the FES symbol turns yellow when the participant is within 2 standard deviations of the optimal range of seat positions to instruct participants to get ready to apply FES. The FES symbol turns green when the participant is within 1 S.D. of the mean to instruct participants to press the button and administer FES to their legs. For each participant, the target seat position range for FES administration will be modified based on their range of motion to account for anthropometric variability. Thus, the optimal target range is $96.5 \pm 93.3$ mm from each individual’s anterior most seat position.

![Figure 15](image)

**Figure 15.** The coaching system interface provides real time feedback on manual button press for administering FES, average power per stroke and stroke rate. A second window provides power output feedback for the duration of a single stroke.

The stroke rate is displayed to allow individuals to develop a consistent rowing pattern. The stroke rate was calculated by 60 seconds divided by the duration of the current stroke (sec). The duration of each stroke was calculated by comparing the time stamps of the beginning of each rowing cycle defined as the anterior-most position of the seat. Our modified ergometer has
shoulder straps and back support which aims to minimize lumbar flexion and extension and thus decreases the stroke length. A small stroke length requires higher stroke rates in order to produce similar work output to a standard rowing ergometer. Furthermore, FES-rowing studies with back supports report that individuals row at higher stroke rates relative to standard rowing studies [5]. Therefore, based on user preference, monitoring and maintaining a constant stroke rate may be useful for developing a consistent pattern of button press timing in some individuals.

The average power output feedback is useful for individuals to track their power over time and meet a set target/goal. The average power output in the drive phase is based on the (force (N) * distance of handle travelled(m)) / time spent in drive phase (sec).

6.4 Delays

When developing the coaching system, it was important to consider the delays in the system. The coaching system has a wide target range for FES timing which accounts for the delays in the system. First, there may be a reaction time delays to the visual feedback presented. The FES symbol turns yellow to instruct the user to get ready, before turning green to instruct the user to apply FES. This allows the individual to anticipate FES timing and minimize this reaction time delay. Second, there will be a small electromechanical delay, between when the FES is triggered, and the muscle contraction is produced. Previous research investigating electromechanical delays in the quadriceps muscles after electrical stimulation have demonstrated minimal delays of 17.2 ms in able-bodied individuals [51]. The target range of seat positions at which to administer stimulation is 96.5 ± 93.3 mm, which allows for a total 186 mm range for individuals to button press. During the end of recovery phase, when the user is expected to apply FES, the seat is moving slowly, so these minor delays will not drastically impact the individual’s ability to button press within this wide target range.

Ultimately, the coaching system provides audiovisual feedback to instruct users on the optimal timing of administering FES for knee extension via a manual push button. The power and stroke rate feedback are supplementary features. Next, it was important to investigate the effect of the coaching system on button press timing.
7 Experiment 2: Testing and Validation of Coaching System

7.1 Introduction:

Individuals with spinal cord injury experience a decline in their cardiovascular and musculoskeletal systems due to paralysis of their limbs[1]. This can be counteracted by several rehabilitation exercises including functional electrical stimulation (FES) rowing [14]. FES-rowing involves voluntary movement of the upper limbs with FES applied to the legs. However, FES-rowing is a complex movement and previous studies have shown that it takes 13 ± 7 sessions for individuals to learn the optimal timing of pressing the button for FES [5]. Therefore, we developed a coaching system to instruct users on the optimal timing to administer FES to their legs through a manual push button.

The optimal timing to administer FES can be delivered to the user through visual or auditory feedback. One study comparing various types of multimodal feedback demonstrated that audiovisual feedback significantly improved learning relative to visuo-haptic and visual feedback only [39]. Therefore, the combination of both visual and auditory feedback in the coaching system may accelerate learning of FES timing.

When considering the frequency of feedback, there is the option of providing summary feedback at the end of the exercise or real time feedback. In specifically rowing studies, Anderson et al. (2005) showed that real-time visual feedback improved rowing performance and consistency relative to summary feedback and no feedback conditions. Therefore, in the coaching system, implementing real time feedback after every button press may be more useful for learning FES timing.

Given that naïve FES-rowers often require many sessions to learn the optimal timing of administering FES to the legs, providing real time audiovisual feedback on the appropriate range of seat positions for administering FES may accelerate learning of manual button press timing[5].
7.2 Methodology

7.2.1 Participants:

7 young non-elite able-bodied individuals participated in this experiment.

7.2.2 Protocol:

- Part A: Warm-up
- Part B: Determine maximum tolerable intensities of quadriceps and hamstrings for administering FES
- Part C: Randomized A-B Protocol
  - A: FES-rowing without coaching system
  - B: FES-rowing with coaching system

First, the participants conducted a warm-up trial on the rowing ergometer. The maximum and minimum seat positions during rowing for each subject was recorded to calibrate the limits of coaching system. This was followed by determining the maximum tolerable intensity of stimulation for each of the target muscles (quadriceps and hamstrings). 70% of the maximum tolerable intensity was used during the subsequent FES-rowing trials. Each trial lasted 2 minutes in duration, followed by 2 minutes of rest.

The adhesive gel electrodes (5 cm x 9 cm) used for FES were placed on the quadricep and hamstring muscles. For each muscle, the cathode was placed proximally over the primary motor point, while the anode was placed distally. Rectangular biphasic pulses were delivered to each muscle using a programmable 4-channel electrical stimulator (Compex Motion 2, Compex SA, Switzerland) to determine the maximal tolerable intensity. For each participant, the intensity of stimulation began at 0 mA and increased at 1 mA intervals until the maximum tolerable intensity was reached for each muscle. 70% of the maximum tolerable intensity was used for each muscle. Based on the results from experiment 1, only the quadriceps were stimulated during the drive phase and only the hamstrings were stimulated during the recovery phase. The stimulation parameters were 40 Hz frequency, 400 us pulse width based on the most common parameters determined from FES-rowing articles [4].
Similar to the first experiment, all rowing trials were conducted on a modified rowing ergometer (Concept 2, Inc., Morrisville, Vermont, USA) with back support, shoulder straps, shank straps, and mounted force and position sensors for data collection.

The force data was collected from custom built foot plates which contain both normal and shear force sensors (Interface Advanced Force Measurement Durham Instruments, Arizona, USA). The seat and handle position data were collected from string potentiometers mounted to FES-rowing system to measure handle and seat position (Measurement Specialties Durham Instruments, Arizona, USA). The analog signals from the sensors were collected using a data acquisition system with a sampling frequency of 2000 Hz (PowerLab 16SP, Cortex and LabChart, ADInstruments, Inc., Colorado Springs, CO, USA).

The manual push button is used to trigger the electrical stimulation to the legs. Holding down the button sends a voltage signal to the electrical stimulator (Compex Motion 2, Compex Motion, Switzerland) activates the quadriceps, and releasing the button opens the circuit which results in hamstring stimulation.

The previous chapter outlined the set-up and features of the coaching system. In brief, the sensor data is read through the 16-bit data acquisition system (USB-6002 Multifunction I/O Device, National Instruments, South Portland, ME, USA) at a sampling frequency of 200 Hz. A laptop (Dell Inspiron 7559, Texas, USA) running the coaching system presented audiovisual feedback to instruct users on the optimal FES timing through a manual push button based on the target seat position range determined from experiment 1. When the user holds down the push button, the electrical stimulator administers FES to the quadriceps muscles to execute the drive phase. They receive auditory feedback on whether they administered FES within the target seat position range. When the user releases the button, the electrical stimulator administers FES to the hamstring muscles to execute the recovery phase.

Simultaneously, the time stamp of the button press and the seat position during button press is recorded in the coaching system. This allows us to determine if the time of button press and seat position during button press are influenced by the feedback provided by the coaching system.
7.2.3 Data Analysis

Timing of button press: First, at the moment of button press for FES to the legs, a signal is sent to the laptop that records the time stamp of the button press. Second, the time index of the anterior most seat position peak (i.e. the start of the drive phase) was also determined. Then, the timing difference between button press and the anterior most seat position was computed for 30 consecutive strokes in the middle of the two-minute rowing trial. This was done for both no coaching and coaching conditions, in the 7 able-bodied participants. Each individual completed the no coaching and coaching trial, so the data was paired. The timing data was not normally distributed, so the median values from each no coaching and coaching condition per participant were compared using the Wilcoxon Signed Rank Test (paired, non-parametric test).

Seat position during button press: Similarly, at the moment of button press, the laptop records the seat position during the button press. 30 consecutive strokes from the middle of the two-minute rowing trial was used. These analyses were repeated for each of the 7 able-bodied participants in the no coaching and coaching conditions. The seat position during button press was not normally distributed as well, and one median value from each no coaching and coaching condition per subject were compared using the Wilcoxon Signed Rank Test.

7.3 Results:

7.3.1 Effect of Coaching System on Seat Position During Button Press

First, the absolute seat position during button press was compared in the no coaching relative to the coaching trial in a typical individual. The seat position during button press did not follow a normal distribution, so the median values of seat position during button press in either trial were compared. In this typical individual, the median seat position in the no coaching trial was 0.71 metres, while the median seat position in the coaching trial was 0.65 metres from the posterior end of the ergometer. In Figure 16, the graphs depict the seat position (m) during every single button press in no coaching and coaching conditions. The results show that all button presses happen within the target range of seat positions as determined by the global mean and global S.D. of seat positions during VL onset from the results of experiment 1.
Figure 16. Effect of Coaching System on Seat Position During Button Press in a Typical Subject. The graphs show the seat position during every single button press in no coaching (blue) and coaching (orange) trial respectively. Dotted line shows the acceptable target seat position range of administering FES based on experiment 1.

Next, the group data was analyzed to assess if there was a significant difference between conditions. The mean, standard deviation and median of seat position during button press and was calculated for each of the 7 participants in the coaching and no coaching condition. For each participant, one trial consisted of around 40 to 60 button presses. After removing the first and last few strokes or corresponding button presses, 30 consecutive seat position values during button press were selected from each participant. Given that the spatial data were paired and not normally distributed, one median value from each condition per participant were used in the
Wilcoxon Signed Rank Test comparing the seat position during press in no coaching and coaching conditions.

Figure 17. Seat position during button press in 7 subjects in no coaching (blue) and coaching conditions (orange). Seat position values were not normally distributed so they are displayed as a box plot.

The coaching system encourages users to press the button at earlier, smaller seat position values, based on the target seat position range determined from experiment 1. But, the Wilcoxon Signed Rank Test on the group data of 7 subjects revealed that there was no significant difference between no coaching and coaching conditions (p=0.375). Secondly, only looking at absolute seat position is insufficient because it does not reveal if button press happens earlier or later than seat peak. In some cases, individuals appear to button press in the appropriate range of seat positions, but in reality, they press the button after they rebound off the anterior stopper. Therefore, it is important to consider the timing of button press relative to the timing at the anterior most seat position to ensure that button press is not delayed.

7.3.2 Effect of Coaching System on Timing of Button Press

Solely analyzing the seat position during button press is insufficient to determine if button press is occurring in a pattern that matches able-bodied leg muscle activation. This is because able-bodied individuals activate their legs prior to reaching the anterior most position of the ergometer
during the recovery for a braking effect as described in the results from experiment 1. In preliminary testing of the coaching system in able-bodied individuals, it was observed that the timing of button press sometimes occurs after the individual has reached the anterior-most position, i.e. after they have finished the recovery phase. Delayed button pressing causes the individual to rebound off the anterior barrier and permits the knees to flex at an unsafe and extreme angle[52], [53].

Therefore, the coaching system encourages individuals to button press before the seat reaches the anterior most position. In order to check whether the button press occurs before the seat reaches the anterior most position, the timing difference between seat peak and button press is shown in the graphs below. The zero value on the y-axis represents the anterior most seat peak. Positive timing difference values indicate that the button press happened earlier than the seat peak. Negative timing difference values indicate that the button press was delayed and happened after the seat peak. In this individual, the results show that 10 button presses without the coaching system were delayed (negative timing difference values). However, with the coaching system, none of the button presses were delayed (all positive timing difference values). This matches the way that leg muscle activation patterns from able-bodied individuals occurs before the anterior most seat position peak as shown by experiment 1.
Figure 18. Effect of Coaching System on Timing of Button Press in a Typical Subject.
Graphs show timing of button press relative to the anterior most seat position (time diff. of zero sec; vertical black bar) in no coaching (blue) and coaching (orange) condition respectively. Positive timing difference values indicate the correct button presses which happened prior to anterior seat peak. Negative values indicate incorrect/delayed button presses.

The mean, standard deviation and median of timing of button press and was calculated for each participant in the coaching and no coaching trial. Given that the timing data were not normally distributed, one median values from each condition per subject were used in the Wilcoxon Signed Rank Test comparing the timing during button press in no coaching and coaching trials for 7 able-bodied subjects.
The time difference between button press and anterior most seat position peak in 7 able-bodied subjects is shown in Figure 19. A timing difference of zero indicates the anterior most seat position (i.e. the beginning of the stroke). Similar to Figure 18, positive time difference values indicate correct button presses and negative time difference values indicate incorrect button presses. In the no coaching condition, there are many negative timing difference values, indicating that few individuals were consistently delayed in button pressing. In the coaching condition, the button presses occurred prior to reaching the anterior most seat position peak. The Wilcoxon Signed Rank test revealed that there was a significant difference in the timing difference between button press and anterior-most seat position peak in the no coaching vs. coaching condition (p = 0.0469).

Figure 19. Timing difference between button press and seat peak in 7 subjects. Positive timing difference values indicate correct button presses that occurred before the seat reached the anterior peak. Negative time difference values indicate incorrect button presses that occurs after the seat passed the anterior peak.

7.3.2.1 Summary of Results

There was no significant difference between absolute seat position during button press (p = 0.3750). However, the main result is that there was a significant difference in button press timing between coaching and no coaching trials (p = 0.0469).
7.4 Discussion:

7.4.1 Summary and Explanation of Main Findings

Using the developed coaching system based on the leg muscle activation patterns from able-bodied individuals, we tested and investigated whether the coaching system would affect button press timing.

The main findings of the current experiment include the following: 1) coaching significantly influences the timing of a button press for administering FES, 2) coaching does not significantly influence the seat position during button press.

7.4.2 Coaching Significantly Influences Timing of Button Press for FES.

Firstly, the results showed that coaching system can influence the timing of the button press. Without the coaching system, individuals pressed the button arbitrarily when the seat was near the anterior most position of the ergometer to stimulate their quadriceps and produce knee extension during the drive phase. Previous studies in FES-rowing with manual push buttons also instructed rowers to press the button arbitrarily when the seat was approximately at the anterior most position [4]. As shown in our results without the coaching system, many of the button presses were delayed and occurred after the seat position reached the anterior most position. However, the coaching system monitors the seat position and instructs users to press the button prior to reaching the anterior most seat position peak based on the results from experiment 1. In the trial with the coaching system, at both the individual and group level, we can see that individuals pressed the button prior to reaching the seat position peak (shown by positive timing difference between seat peak and button press).

These results are important because it shows that the timing of button press can be altered with only one two-minute trial of coaching in novice or non-rowers. Prior studies have claimed that it takes individuals with spinal cord injury around 13 ± 7 sessions to learn the timing of button press [5]. The stark difference demonstrates that the coaching system 1) has the potential to influence button press timing and 2) may demonstrate observable changes with only one two-minute trial.
Using the coaching system to ensure that button press occurs before the seat reaches the peak is essential because otherwise the individual will activate their leg muscles too late and will rebound off the anterior stopper. The leg muscles must activate prior to the seat reaching the anterior most position in order to successfully brake against the forward movement and prevent this rebound motion off the anterior stopper[45]. Secondly, if the individual does not brake, their knees will flex to an extreme angle. When the individual subsequently loads the knees with compressive forces to initiate the drive phase, while in this vulnerable position (maximally flexed knee angle), it may lead to injury[52], [53]. Although, elite rowers may want to travel to the anterior-most seat position to maximize the stroke length and power output, the target of the coaching system is for individuals with spinal cord injury, so it is important to prioritize safety over performance.

7.4.3 Coaching does not significantly influence seat position during button press for applying FES.

In the individual participant’s data, the seat position during button press during the no coaching and coaching conditions both look very similar. The coaching system instructs users to administer stimulation based on the target seat position range (global mean ± 1 S.D.) during leg muscle activation in able-bodied individuals as determined from experiment #1. In both the no coaching and coaching conditions, all the seat position values seem to fall within this expected seat position range. Wilcoxon Signed Rank Test on the group data suggest that there is no significant difference in seat position during button press in no coaching or coaching conditions (p = 0.3750).

An alternate explanation is that these results may be due to low sample size because in the box plot, we can see that coaching may slightly influence individuals to button press at smaller seat position values. Since the seat position sensor is mounted at the posterior end of the ergometer, smaller seat position values imply that the individual is closer to the posterior end. Therefore, even though there was no statistical difference in seat position during button press between conditions, a higher sample size may reveal that coaching influences individuals to button press at smaller seat position values (i.e. further away from the anterior end and at safer knee angles).
These results are important because they reveal that only analyzing absolute seat position is insufficient in determining whether the individual stimulated at the appropriate time to mimic able-bodied leg muscle activation patterns as determined by Experiment 1. This is because individuals tended to stimulate after reaching the anterior-most seat position, and only comparing seat position would not be able to easily detect these mistakes.

Furthermore, the implications of these results suggest that coaching feedback systems that rely on only seat position would be lacking. One way to combat this issue is to analyze the timing of button press to ensure that it happens prior to the seat peak. A more feasible method, which was used in the current coaching system, is to ensure that the seat is moving forward. If the seat is still moving forward during button press, this guarantees that the individual is still finishing the recovery phase and has not yet reached the anterior most seat position. On the other hand, if the individual presses the button after the anterior seat position peak, the seat will be moving backwards, and the coaching system can notify the user of an incorrect button press.

7.4.4 Relate the findings to similar studies

An early study by Page et al. (2003) involved the development of a training system to provide real time biomechanics feedback (kinematic and kinetic data) for rowers. They had instrumented an ergometer with a load cell at the handle and a series of string potentiometers to determine the handle and seat positions. The system provided real time feedback on the kinetics and kinematics using a stick figure animation. This study demonstrated that it was possible to provide real time biomechanical visual feedback to rowers. Alternatively, our coaching system is targeted for individuals with incomplete spinal cord injury and is instrumented with several load cells underneath the foot plates. Displaying foot force may be more clinically applicable for individuals with spinal cord injury as it can allow them to keep track of foot force improvements over time.

More modern studies have shown that virtual reality can also be used to accelerate learning, modify energy management and improve rowing performance in 7 novice rowers exercising on an indoor rowing ergometer in 8 training sessions. These studies have shown that gamification and visual feedback systems have the potential to modify energy management strategies and improve rowing performance in able-bodied individuals. Comparatively, our
results showed that the addition of a coaching system could improve timing of button press in only one two-minute training trial. The discrepancy in number of sessions required to master a specific skill may be based on the complexity of a skill. Our coaching system accepted a wide range of acceptable seat positions at which the individual could press the button for FES-induced knee extension. The wide range and relative ease of skill may explain the fewer number of sessions required to learn the button press timing.

7.4.5 Implications for Testing in Individuals with SCI

Furthermore, when testing our coaching system in individuals with spinal cord injury, it may take several sessions for them to learn the target range at which to button press for FES due to lack of interlimb coordination, lack of finger control, decreased aerobic capacity, and rapid muscle fatigue due to atrophy[37]. Previous studies have shown that it requires 13 ± 7 sessions for individuals to naturally learn the proper timing[5][12]. Even though the coaching system was able to instruct novice able-bodied rowers on the timing of button presses relatively quickly in our study, it will likely require multiple sessions for individuals with spinal cord injury to learn to button press at an appropriate time to produce this complex movement.

7.4.6 Alternate Explanations for Findings

It is unlikely that individuals may have learned the proper timing of administering stimulation through practice or the order of trials because the trial order was randomized. Furthermore, without the coaching system, most of the participants pressed the button to administer stimulation arbitrarily: both earlier and later than seat position peak.

One participant had similar results in both no coaching and coaching conditions. This may be due to the novelty, lack of familiarization with the coaching system or the complex rowing movement. If the user felt overwhelmed by the amount of feedback, this could impair performance as well.

7.4.7 Visual and Auditory Feedback Systems:

Previous research demonstrated that feedback has the potential to accelerate learning in various tasks[55]. However, it is important to regulate the amount of feedback provided to prevent cognitive overload [55]. The literature is controversial about the optimal method and modality of
delivering feedback, but this is possibly due to the differences in the complexities of tasks studied (i.e. rowing a is relatively complex task that requires coordination of the upper and lower limbs, while wrist flexion tasks are examples of simpler tasks)[55]. Multimodal feedback is encouraged in many studies to enhance complex motor learning[55]. It is also recommended that the amount of feedback provided could be tailored to the individual’s level of experience and personal preferences[55].

7.4.8 Limitations

Some participants had a lower threshold for pain, so they were not able to achieve a full involuntary contraction. They reported that part of the rowing movement was indeed voluntary. This is difficult to address because able-bodied individuals are likely to have some voluntary contribution, but more exposure to FES may enable them to tolerate higher current intensity and produce a contraction that is largely involuntary.

On the other hand, subject 7 reported that administering FES using a push button successfully initiated the contraction, even if the remaining component of the knee extension was voluntary. For the purposes of this experiment, we investigated whether the coaching system can influence the timing of button press. Therefore, even if the whole contraction is not voluntary, it is important and useful that the FES button press successfully initiates the contraction. This allows us to modify the initiation of the contraction in individuals with spinal cord injury, who likely would be able to tolerate higher intensities of stimulation due to decreased sensory perception. In individuals with spinal cord injury who experience denervation, this FES-rowing exercise may not be suitable. Multiple sessions of FES may improve the contraction response to FES.

Another limitation is the broad range of rowing experience. One individual had never rowed before, so they had trouble coordinating button presses and rowing technique. Therefore, in the beginning, an automatic stimulator might be useful for them to get an idea of the movement, and then progress to the manual stimulator once they are comfortable performing the movement.

Finally, it is important to note that the FES-rowing literature does not address or account for early leg muscle activation seen on modified ergometers. Previous studies involving electromyography of the legs in able-bodied individuals supported the results from Experiment
demonstrating that leg muscle activation onset occurs before the anterior most seat position peak. However, only a few FES-rowing studies have considered the requirement of braking by early leg muscle activation to counteract the forward motion at the end of the recovery phase[20], [42]. One possible explanation for the lack of literature on early leg muscle activation in FES-rowing is that in individuals with SCI, the recovery phase may be slow since their hamstring muscles are weaker, which may imply that there is less of a requirement for braking, i.e. the individual may naturally stop before they hit the anterior most stopper, and thus prevent themselves from going to extreme knee angles. However, this is unlikely to be seen in our custom FES-rowing setup, since the spring mounted at the posterior end would facilitate the recovery phase and increase the braking required.

7.4.9 Suggestions for Further Research

There are a plethora of studies investigating the influence of feedback on improving a skill or performance. The modality of feedback (visual, auditory, haptic, multimodal), the time window that the feedback is provided in (duration and delay of feedback), the complexity of the task, the skill level of the individual, and many other factors can affect the improvement due to feedback[56]. In order to accelerate learning, the coaching system should be intuitive to use, provide important information as feedback without overwhelming the user, increase user engagement and volitional input.

Furthermore, the current coaching system has been developed such that the target range at which to administer stimulation can be altered. Personalizing the recommended range of seat positions at which to administer stimulation could be beneficial due to anthropomorphic variability (i.e. variability in shank to thigh ratio or limb length).

Future testing in individuals with spinal cord injury may reveal design or accessibility modifications such as larger text or providing auditory-sonification feedback (the degree of error is related to the pitch of the auditory feedback; ex. worse mistakes result in a higher pitches / tones of auditory feedback)[39]. For individuals with spinal cord injury, external spinal fixation using a rod and straps may also help prevent soft issue injury in the lumbar and sacral region of the spine[21].
8 Conclusions

Spinal cord injury results in compromised cardiovascular and musculoskeletal systems due to limb paralysis and inactivity[1]. FES-rowing is a complex movement that aims to counteract the comorbidities associated with spinal cord injury by incorporating paralyzed leg muscles using electrical stimulation. Other studies in FES-rowing state that participants arbitrarily button press to administer stimulation to the legs when the seat is near the anterior most position. Thus, it takes many sessions for individuals to learn the timing of button press to administer FES to the legs and consequently receive the benefits of the exercise[5]. Therefore, it was necessary to identify the timing at which to administer FES, develop a coaching system to instruct new rowers on the optimal FES timing, and subsequently test the coaching system.

First, we instrumented a rowing ergometer with sensors for data collection and minor modifications to stabilize individuals with SCI on the rowing ergometer. In order to identify the optimal timing of administering FES, our first experiment investigated the timing of leg muscle activation patterns in able-bodied individuals on our modified ergometer. The quadriceps were active slightly prior to the drive phase for a braking effect, and the hamstring muscles were active during the recovery phase. Based on the timing of leg muscle activation onset from experiment 1, we developed a coaching system in Python to instruct users on the optimal timing range of manual button press using visual and auditory feedback. Individuals should administer stimulation to their legs at the target range of 96.5 ± 93.3 mm before the seat reaches the anterior most position (i.e. before the end of the recovery phase). This is important to induce a braking effect and prevent extreme knee flexion which may result in injury[52]. In experiment 2, we showed that there was a significant difference in FES button press timing between no coaching and coaching conditions in able-bodied individuals. The developed coaching system guided users to administer FES to their legs prior to the anterior-most seat position peak. Future studies in FES-rowing involving individuals with spinal cord injury can use the coaching system to accelerate learning of FES timing enabling them to receive the benefits of the FES-rowing exercise.
9 References


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10 Appendix: Closed Loop Automatic Stimulator Option

10.1 Introduction:

There are two main types of stimulators used for FES-rowing that are prevalent in the literature. The manual stimulator involves pressing a button or a switch to administer electrical stimulation to the legs. Automatic controllers also exist that deliver stimulation based on handle or seat position.

We developed an automatic stimulator to administer stimulation to the quadriceps during the drive phase and to the hamstrings during the recovery phase. This is based on the muscle activation patterns from various papers and Experiment 1. The seat position data is monitored constantly, and electrical stimulation to the appropriate muscle is provided accordingly. Just before the seat reaches the anterior most target position on the ergometer, this indicates the start of the drive phase and the laptop outputs a signal to the DAQ and then the Compex stimulator to administer FES to the quadriceps. Once the seat reaches the target posterior-most position on the ergometer, it indicates the start of the recovery phase and the laptop outputs a signal to the DAQ and then the Compex to administer FES to the hamstrings.

Even though the DAQ is able to sample at 200 Hz, the python code is only able to update every 30 ms and is a limiting factor. However, for the purposes of this experiment a sampling rate around 33 Hz is not detrimental to the system and was not noticeable by participants. Also, one stroke rate is around 2 seconds in duration, and the optimal range of recommended seat positions for administering stimulation encompasses at least 5 sensor readings.
Figure 20. Closed loop automatic stimulator. The computer reads the position of the seat at a sample rate of 200 Hz. It sends a signal to the electrical stimulator to administer FES to the appropriate muscle when the seat reaches a target seat position. For example, when the seat reaches the target anterior-most position, it stimulates the quadriceps for knee extension during the drive phase. When the seat reaches the target posterior-most seat position, it stimulates hamstrings for knee flexion during the recovery phase.

10.2 Rationale for Seat Position Based Controllers Rather than Time Dependent Controllers

Previous stimulators relied on seat position rather than timing difference because it allowed individuals to complete each state of rowing at their own pace with no fixed time intervals[35]. This is more beneficial for beginners who may not have a consistent stroke rate and for individuals with spinal cord injury who may have muscle weakness impairing their ability to coordinate their upper and lower limbs[35]. Furthermore, this allows participants to row at their
preferred stroke rate, and to adjust their stroke rate during the trial in case they are fatigued. Therefore, the current automatic stimulator administers electrical stimulation based on seat position.

The graphs below show the validation of the automatic stimulator in 5 able-bodied individuals across a two-minute FES rowing trial. When the individual reaches the target anterior and posterior seat positions, the stimulation switches to either quadriceps or hamstrings respectively. These target seat positions were calibrated based on the individual’s range of motion[35]. As shown in the graphs below, the seat positions at the moment of stimulation was constant because it was initiated by the computer. The timing of stimulation was also constant because individuals were rowing at a constant stroke rate.

### 10.3 Testing and Validity of Automatic Stimulator

![Graph showing seat position at moment of stimulation](image)

**Figure 21.** The seat position at the moment of stimulation was recorded across 30 consecutive strokes in 5 subjects. The target seat position for administering stimulation was based on the seat position during leg muscle activation as determined by the results from Experiment 1.
Figure 22. The timing difference between button press and seat position peak was calculated across 30 consecutive strokes in 5 subjects. The positive difference values indicate that the stimulation occurred around 0.3 seconds before the seat reached the anterior most position.

The graphs displaying seat position at the moment of stimulation and the timing difference both show that the automatic closed-loop stimulator can allow individuals to perform FES-rowing in a consistent manner.

10.4 Benefits of Closed Loop Stimulator

The benefits of the closed loop stimulator option improve accessibility of the FES-rowing machine for individuals 1) who had difficulty with manual button press due to lack of voluntary control or 2) who had trouble coordinating the button press for knee extension at the correct time[37]. The system reads the seat position and administers stimulation to the appropriate muscles. It stimulates the quadriceps for knee extension (drive phase) just before the seat reaches the target anterior most position and stimulates the hamstrings for knee flexion (recovery phase) when the seat reaches the target posterior most position. These targets are determined based on each individual’s range of motion.
10.5 Drawbacks of Closed Loop Stimulator

The drawbacks of this closed loop system are that it decreases the voluntary contribution of the individual (the manual button press) and requires less attention. Button pressing requires concentration in order to learn the optimal timing, and this voluntary contribution may be important for maximizing the CNS plasticity during rowing. Anecdotal evidence from some articles claim that individuals may prefer to use manual button pressing systems because it allows for better control of the stimulation [33], [37], [47]. The automatic controllers felt like individuals were being jerked back and forth [33], [37], [47]. Therefore, the coaching system was developed to instruct individuals on the timing of initiating a manual switch.