FORWARD DYNAMICS OF FUNCTIONAL ELECTRICAL STIMULATION (FES) ROWING FOR INDIVIDUALS WITH SPINAL CORD INJURY (SCI)

A Thesis Presented

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ABSTRACT

Spinal Cord Injury (SCI) is a devastating life-altering event that impacts the patient physically, emotionally, socially and economically. The population of individuals living with an SCI in the United States grows by about 18,000 each year adding up to an estimated 288,000 as of 2017. SCI and its repercussions introduce high levels of morbidity and low quality of life in these individuals. After SCI, immobilization due to paralysis is observed in a vast majority of cases making patients dependent on assistive devices to perform motor functions.

Immobilization severely affects bone health in SCI population due to lack of loading on legs. Studies have shown that about 60% loss of bone mineral density can be seen in just 1-3 years post injury. This phenomenon is known as disuse osteoporosis and makes bone highly vulnerable to low energy fractures while performing daily activities of living.

Functional Electrical Stimulation (FES) is a rehabilitation therapy that uses electrical signals to produce contractions in paralyzed muscles. When FES is used to flex and extend the legs during ergometer rowing (FES rowing), studies have shown cardio-respiratory benefits. However, biomechanical analysis showed that the force at the feet was modest compared to able-bodied rowers, indicating that FES rowing may be inadequate to prevent bone loss and promote bone growth.

This thesis presents a musculoskeletal forward dynamics computational simulation of FES rowing to probe the effects of muscle timing, assistive springs, and muscle atrophy on the knee joint load. A musculoskeletal model of a rower was built in OpenSim and forward dynamics tool was used to a perform muscle activation driven simulation. The work presented suggests that reaction forces through the feet are higher when the rower does not rely on assistive springs but uses higher intensity muscle activation to perform the rowing motion. Individual effects of assistive spring forces, muscle activation levels and timing of activation on foot reaction force and knee force were studied. Computational models of rehabilitative therapy may provide a guide to effective therapy for bone health.
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Chapter 1.

Introduction and Objectives

1.1 Introduction

Spinal cord injuries (SCI) are predominantly caused by trauma to the vertebral column affecting its ability to carry signals from the brain to the rest of the body. This usually results in a loss of sensory function and/or motor control over all (quadriplegia) or lower (paraplegia) extremities depending on the site and severity of injury. It is estimated that every year about 17,700 new cases are recorded in the United States, adding up to about 288,000 persons currently living with SCI. [22]

SCI often also leads to a loss of mobility which inhibits individuals with SCI to stand or walk. Decreased mobility prevents the lower limbs from getting loaded due to insufficient weight bearing. Mechano-adaptation is a property of bone that influences a change in bone density and shape when a change in mechanical strain is sensed. Higher mechanical strain promotes bone growth and lack of mechanical strain leads to bone resorption. Studies have shown that lack of loading on bones may result in osteoporosis. This is also known as disuse osteoporosis and it makes bone vulnerable to low energy fractures while performing daily activities. [18]

Functional electrical stimulation (FES) is a technique that uses external electrical stimuli to stimulate muscle contractions. FES therapies are used to rehabilitate patients with neurological disorders or traumatic spinal cord injuries. FES rowing is used in persons with paraplegia. It combines voluntary upper body exercise with externally stimulated lower body movements. Electrodes are placed on the patient’s quadriceps and hamstrings to perform hip and knee flexion and extension. Previous studies have shown that FES sowing can be an effective way to stimulate metabolic and cardiovascular activity in individuals with SCI. However, the therapeutic effect of FES Rowing on bone to prevent disuse osteoporosis is unknown. Previous studies have indicated foot forces during FES rowing are very small (10% BW), but forces around the knee are likely higher and may have therapeutic benefit.
1.2 Objectives

There have been studies that examined the effect of muscle activation in FES based therapies such as FES cycling. But the effect of muscle activation on motion and joint loading in FES rowing is unknown. We analyzed how muscle activity and external forces affect joint reactions, particularly at the feet and knees. We used a forward dynamics approach where muscle activity is specified and motion is analyzed. This is done using the Forward Dynamics tool in OpenSim which uses a mathematical model to describe how forces and moments applied through muscle contractions influence motion. The goal was to probe parameters such as muscle activation, assistive spring forces, and muscle atrophy in order to determine its effects on knee joint loading and understand therapeutic options for achieving higher loading on legs and joints for musculoskeletal benefit.
Chapter 2.

Background

2.1 Spinal Cord Injury

The human spinal cord is a complex arrangement of nerves that run caudally through the vertebral canal from the medulla oblongata of the brain to the lumbar region of the spine. This slender, long, and curvy cylinder is a part of the central nervous system (CNS) and carries 31 pairs of spinal nerves. The spinal cord is protected by bony structures called vertebrae. This stack of vertebrae makes up the Spine and is divided in four regions: cervical, thoracic, lumbar and sacral. These vertebrae are numbered starting from the top to the bottom as: 7 cervical vertebrae (C1-C7), 12 thoracic vertebrae (T1-T12), 5 lumbar vertebrae (L1-L5), 5 sacral vertebrae (S1-S5) and the coccyx. The nerves in the spinal cord exit the spine from between the vertebrae through the intervertebral foramen.

The spinal cord is responsible for carrying messages from the brain to other parts of body and vice versa through nerves. These messages control many vital functions of the body. It controls our motor functions by commanding voluntary muscles to contract and relax. It also carries sensory signals of touch, pressure, temperature and pain from the body to the brain and can also control involuntary functions such as digestion, heart rate and contraction/dilation of blood vessels.

Spinal cord injury (SCI) is a devastating life-altering event that impacts the patient physically, emotionally, socially and economically. In the United States, about 92% cases are caused due to trauma to the spinal cord. [22] Vehicular crash is the most dominant cause of SCI making up to 38.3% of all causes. Falls, violence (mostly gunshot wounds) and sports are other leading causes. Since, the cause in almost all cases is trauma related, the victims range from all ages. Quality of life after a spinal cord injury is severely affected. [30] Spinal cord injury impairs the sensory function of the nerves. Sensory score is required to be assessed after a spinal cord injury. Sensory response from dermatomes for each 28 key sensory points on each side is tested and scored for pin prick and for light touch. [15] Therefore, a maximum achievable sensory score would 56 for pin prick, 56 for light touch and 112 for both. Motor score is required to document change in motor function after the injury. Five key muscle functions of each extremity are tested and graded on a scale of 1-5, adding up to a maximum score of 25 per extremity or 100 for all
extremities. [26] The American Spinal Injury Association (ASIA) has also devised an impairment scale to assess the extent of neurological damage from a spinal injury. The scale has 5 degrees of impairment as [taken verbatim from ASIA International Standards for Neurological Classification of Spinal Cord Injury]:

A = **Complete** - No sensory or motor function is preserved in sacral segments S4-S5

B = **Sensory Incomplete** - Sensory but not motor function is preserved below the neurological level and includes the sacral segments S4-S5, AND no motor function is preserved more than three levels below the motor level on either side of the body.

C = **Motor Incomplete** - Motor function is preserved below the neurological level, and more than half of key muscle functions below the single neurological level of injury have a muscle grade less than 3 (Grades 0–2).

D = **Motor Incomplete** - Motor function is preserved below the neurological level, and at least half (half or more) of key muscle functions below the NLI have a muscle grade >3.

E = **Normal** - If sensation and motor function as tested with the ISNCSCI are graded as normal in all segments, and the patient had prior deficits, then the AIS grade is E. Someone without a SCI does not receive an AIS grade.

This impairment scale helps clinicians understand and evaluate the severity of the injury and plan treatment and rehabilitation for the patient.

Recent estimates show that about 17,700 new spinal cord injury cases are documented every year. [22] The number of people currently living with a spinal cord injury can be anywhere between 247,000 and 358,000, majority of these being males.

### 2.2 Disuse Osteoporosis

Bone is an active living connective tissue that undergoes remodeling and reshaping throughout its life with the help of osteoblasts and osteoclasts. Osteoblasts are cells responsible for bone formation and osteoclasts are responsible for bone resorption. Old bone is constantly replaced with new bone. [12] This process of bone formation and resorption, under equilibrium, is a vital function since it helps build new bone in an event of a fracture or even micro-damage. Bone remolds its internal structure and mass to adapt to the mechanical environment to maximize strength and reduce the risk of fractures. [16] It has been shown that new bone formed
is aligned along the direction of mechanical strain. Osteocytes are the cells that are responsible for sensing this mechanical strain. When osteocytes sense a change in the mechanical environment, osteoblasts and osteoclasts are deployed to either form or resorb bone.

**Figure 1 Model for the transduction of mechanical strain to osteocytes in bone [16]**

Disuse osteoporosis is defined as loss in bone density caused due to insufficient loading or mechanical strain on bone. Osteocytes sense the disuse or lack of loading on bone and respond to it by triggering osteoclasts that are in turn responsible for bone resorption or bone loss making the bone fragile or osteoporotic. Cancellous bone is highly vulnerable to disuse osteoporosis compared to cortical bone and hence, bones with a higher proportion of cancellous bone are adversely affected. Disuse osteoporosis makes the bone brittle and prone to fractures or damage. In most cases, disuse osteoporosis is caused by absence or insufficiency of weight bearing, ground reaction forces and dynamic loading from movements like walking, running or standing. Prolonged bed rest, paraplegia or tetraplegia due to paralysis and microgravity are some mechanisms through which disuse osteoporosis can be induced.

After an incident of a spinal cord injury, bone mineral density at some specific sites drop at an extreme rate of about 1% per week in the first 3-4 months. Bone mineral density continues to drop at this increased pace for the next 12 months to up to 3 to 8 years in some cases before slowing down.
In a study by Vestergaard P. et al [29], it was found that low energy fractures in the lower extremities were common in SCI patients. Patients with paraplegia are at a greater risk of fractures compared to patients with tetraplegia. [27] The reason behind that might be higher levels of activity such as getting in and out of the wheelchair.

### 2.3 Functional Electrical Stimulation (FES)

A spinal cord injury can affect the central nervous systems and paralyze a set of muscles depending on the location and severity. This results in an inability of the brain to voluntarily produce contractions in these skeletal muscles. Functional Electrical Stimulation or FES employs electrical pulses of low power to externally produce contractions in paralyzed muscles. FES can either be executed using surface electrodes (transcutaneous) or subcutaneous electrodes (percutaneous or implanted) electrodes. Surface electrodes are most commonly used since they are non-invasive, easy to apply, and inexpensive. They are applied over the muscles that need to be stimulated. However, stimulating profound muscles such as the hip flexors may be challenging using surface electrodes. [2]

FES based therapies have been monumental in rehabilitating individuals with spinal cord injury, drop foot, multiple sclerosis, cerebral palsy, and stroke patients. Over the last few decades, FES cycling has been used to provide cardiorespiratory exercise to individuals with SCI. However, the cardiovascular demand in Hybrid FES Cycling (Arm Crank Ergometry + Leg Cycling Exercise) was found to be higher than just FES cycling. [13] FES cycling may also have the ability to restore voluntary leg muscle strength and function. [4]. Although it can prevent bone loss and lower the risk of fractures, it is only limited to actively loaded femur. Passively loaded tibia does not have a significant effect on skeletal parameters from FES cycling. [8]

FES rowing uses rowing ergometer that is slightly modified for paraplegic persons. It uses voluntary upper body function and FES assisted lower body function to produce a rowing motion. Surface electrodes are used to externally stimulate the quadriceps and the hamstrings. This is controlled by the rower via a switch on the handle that triggers an electric signal that activates the corresponding muscles. Activating the quadriceps initiates drive that helps the rower translate from the front to the back and activating the hamstrings initiates recovery that helps the rower translate from the back to the front again. The seat has a back rest and straps to keep the trunk stable. It also has safety stops at both ends that are spring loaded to assist the rower during
change of momentum. Legs are constrained to move in the sagittal plane by the use of straps and a telescopic arm.

![Figure 2 Schematic of FES rowing ergometer][31]

FES rowing has been successfully used to provide cardiorespiratory exercise to individuals with SCI lowering the risk factors for cardiovascular diseases such as Type 2 diabetes and obesity. [14] A single subject study found that FES rowing could induce loading on legs and have a therapeutic effect on osteoporosis. [11] Biomechanical analysis by Draghici A. [5] et al suggested that loading on legs is modest in SCI population compared to able bodied rowers. Poor coordination between the upper and lower body and muscle atrophy might be reason for the low levels of loading.

### 2.4 Forward Dynamics of FES based therapies

Forward dynamics simulations are muscle driven simulations that help researchers study human movement resulting from varying muscle activity. It can be instrumental in analyzing FES based therapies since they employ different muscle activation patterns to achieve certain motion.

In FES cycling, forward dynamics has been used to analyze the effect of parameters such as seating position, physiological parameters, pedaling rate and muscle activation patterns. [19]
FES rowing is a possible means to multi-system rehabilitation individuals with SCI but there are no studies that examine the effect of muscle activity and other parameters on the motion of rower. Forward dynamics can provide an insight into the kinetics and kinematics of FES rowing. Through simulation we can determine the joint loads, which have implications for the therapeutic effect on bone.
Chapter 3.

Methods

3.1 FES Rowing Data Acquisition

A previous study acquired kinetic and kinematic data from six SCI subjects who performed FES rowing [5]. Subjects had injuries at different levels ranging from neurological levels C6 to T4. Force data acquired from these subjects helped us establish a benchmark for handle and spring forces. We used data from a male subject who weighed 87.36kg and was 1.702m tall.

![Figure 3 Instrumented rowing ergometer [5]](image)

The experimental setup was at Spaulding Rehabilitation Hospital and consisted of a modified Concept2 rowing ergometer. Four force transducers were placed on the foot board to record reaction forces from the left and right toes and heels. One force transducer was used to measure the upper body force via the handle. To record the translation of the seat with respect to the feet, a linear string potentiometer was used. Two linear string potentiometers recorded the deflection of the front and rear assistive springs which was used to calculate the spring forces. Figure 3 shows the experimental setup with instrumentation.
3.2 Musculoskeletal Modeling

Musculoskeletal modeling uses mathematical models of the muscle and bone system to analyze the kinetics and kinematics under the action of muscle activation and external forces. The skeletal system is considered to be a system of interconnected rigid bodies while the muscles are considered to be force producing elements. The movement of the skeletal system is based on the laws of mechanics whereas the force produced by muscles is based on the force-length property and force-velocity property of muscles. Musculoskeletal modeling is vastly used by healthcare researchers since it can non-invasively provide an insight on the mechanics of the human body.

We used OpenSim to build the musculoskeletal model, simulate muscle contractions via forward dynamics and analyze motion, muscle forces, reaction forces and joint forces. OpenSim is freely available open source software that allows users to mathematically model and simulate the musculoskeletal system in a dynamic environment. [25]

3.2.1 Musculoskeletal Model of FES Rower

A musculoskeletal model in Opensim consists of two major elements: bones and muscle tendon units. Bones are modeled as rigid bodies that do not deform under load. Bones are interconnected using joints that are defined by their degrees of freedom rotation or translation. Muscle tendon units are modeled using Hill-type contractile element in series with a tendon. The force produced by the contractile element is determined by a predefined force-length and force-velocity relationship as shown in the figure. The muscle and the tendon also have inherent stiffness that is responsible to produce passive muscle forces. To characterize each muscle tendon unit in the model, maximum isometric force, optimal muscle fiber length, tendon slack length, maximum contraction velocity and pennation angle are the parameters used. [28] These parameters are obtained from cadavers.
The model developed by Rajagopal et al. [24] is a full body model with high-fidelity representation of the lower limbs to produce muscle driven simulations. However, we found out that this model is not suitable for analyzing motions that involve a high degree of knee flexion. It fails to produce accurate results in such cases. Lai et al refined this model so that it can be used for motions that involve high degrees of hip flexion (≤120°) and knee flexion (≤140°). [17] Active force generating properties and passive force generating properties of the previously problematic muscles such as the knee and hip flexors were altered in order to make the model compliant in cases of high flexion. Optimal muscle lengths were also increased within limits.

We adapted this high flexion model for forward dynamics of FES rowing. This is a whole body model with a 3D skeleton geometry that includes 22 articulated rigid bodies. The lower body consists of the pelvis, right and left femur, patella, tibia, talus, calcaneus, and toes. The upper body consists of a combined head, torso and arms. The original model had 23 degrees of freedom (20 degrees of freedom for the lower body and 3 degrees of freedom for the upper body). Figure 5 shows a topology view of the model. It weighs 75kg and is 175cm tall. Since the subject the force data was based on weighed 87kg, we scaled the model by weight. No changes in height or length of segments were made.
Figure 5 Musculoskeletal model developed by Lai et al (left) and Topology view of the model indicating bodies and joints (right)

Rowing is a planar motion that occurs mostly in the sagittal plane. In order to limit the complexity of the model, we locked all translational movement in the Z-direction and all rotation in the X-Z and Y-Z planes. Hip adduction, hip rotation, metatarsophalangeal (mtp) angle and subtalar angle were constrained to fixed values throughout the motion. Lumbar bending, extension and rotation were also locked. We introduced a new translational degree of freedom along the X-axis on the pelvis that represented the sliding of the seat on the rails. The slider was limited to move within a predetermined range. The hip joints were free to flex and extend from -30° to 120° whereas the knee joints were allowed to be able to flex and extend from 0° to 140°.

To constrain the feet, we added a point constraint at the toes. The location of the toes with respect to ground was fixed but rotation in the X-Y plane was allowed. Figure 6 shows the model with constraints at the starting position.
Figure 6 FES rower model in OpenSim illustrating directions of slider motion and the external forces; and spatial constraints

3.3 Forward Dynamics

The model described above was used to run forward dynamics simulations. Fig. 7 describes the simulation. Muscle excitation, assistive spring forces and handle force were the inputs for the simulation. We varied the muscle properties depending on the type of model and observed the effect of these in motion, foot force and knee force.

Figure 7 Layout of the forward dynamics simulation showing inputs and outputs.
3.3.1 Muscle Activation

During FES rowing, quadriceps and hamstring muscles are stimulated externally to achieve the desired motion. To simulate external stimulation in OpenSim, quadriceps and hamstring muscles of the model were activated using the forward dynamics tool. The quadriceps muscles used were vastus medialis, vastus lateralis and vastus intermedius. Rectus femoris, that is a part of the quadriceps, was not activated since the placement of electrodes on the thigh is such that it does not trigger this muscle. The hamstring muscles that were activated were bicep femoris (long and short heads), semimembranosus and semitendinosus. The quadriceps muscles were activated at the beginning of the cycle during the drive phase and the hamstring muscles were activated during the recovery phase of the motion using a step-type input. The rest of the muscles were retained in the model but were not activated.

3.3.2 External Forces

The external forces were recorded from an SCI subject rowing on an instrumented ergometer. The ergometer measured handle force, front spring and back spring displacement, and force at the feet. We estimated the following external forces from this experimental data:

**Handle Force**: 150N

**Front Spring Force**: 300N

**Rear Spring Force**: 450N

All external forces to the model were applied along the X-axis through the center of mass of the pelvis to avoid complications and any resulting moments. This data provided baseline around which we examined changes in front spring force. The foot force was an output of the model and was compared to the experimental values.

3.4 Model based on muscle atrophy

Not only do individuals with SCI undergo severe bone loss, but also experience drastic muscle atrophy. [10] These effects are even more prominent in the lower extremities. [21] To understand the effects of FES rowing on patients with atrophied muscles, we built a model that simulated muscle loss. We modified the passive force generating ability of the model by lowering
the peak isometric force of the gluteus maximus, gluteus medius, soleus, piriformis and adductor magnus to 50% of a healthy muscle. Fig. 8 and Fig. 9 show passive forces in the non-atrophy and atrophy models when the motion was consistent whereas the muscle activation was changed. Effects of muscle activity and assistive spring forces were analyzed in this model as well.

Figure 8 Passive muscle forces in the non-atrophy model
Figure 9 Passive Muscle Forces in the atrophy model
Chapter 4

Results

4.1 Foot force in subject versus models

When forward dynamics was run on the non-atrophy model described in section 3.2.1, we observed very high passive forces in some of the muscles. These large passive muscle forces contributed to a foot reaction force that was approximately twice the magnitude of that recorded by the subject FES rower. For the atrophy model, the passive forces generated were substantially lower compared to the non-atrophy model and comparable to the experimental data from the FES rower. The muscle activation levels for both models is shown in Fig. 10 and Fig. 11 shows the total foot force normalized by body weight for both models and the subject. Motion was kept consistent in these tests as shown in Fig. 12. The atrophy model better represented the subject FES rower whereas the non-atrophy model had foot forces more similar to an able-bodied subject. (Experimental Data not shown)

![Figure 10 Muscle Activation levels in Non-Atrophy and Atrophy models](image_url)
Figure 11 Normalized Total Foot Reaction Force in Non-Atrophy Model and Atrophy Model versus that observed in subject FES rower.

Figure 12 Comparison of motion in Non-Atrophy Model and Atrophy Model versus subject FES rower using seat position co-ordinate.
4.2 Effect of varying spring force on Foot Force, Knee Force and Motion in Non-Atrophy Model

To examine the effects of varying front assistive spring force in the non-atrophy model, we ran three forward simulations. In this case, the muscle activation in the quadriceps and hamstrings was same in these simulations. The front spring force was varied by increments of 100N from 200N to 400N. Foot forces and knee forces were normalized by body weight. Fig. 13 shows the results. The range of motion of the rower (i.e. seat displacement) increased as the front spring force increased. However, no significant effects were seen in the foot reaction force and knee joint reaction force, showing that the spring only assisted the rower motion and did not have a significant effect on bone loading.
Figure 13 Normalized Total Foot Reaction Force and Normalized Knee Joint Reaction Force over one rowing cycle with varying front spring force and constant muscle activation in Non-Atrophy Model
4.3 Effect of varying muscle activation on Foot Force, Knee Force and Motion in Non-Atrophy Model

To examine the effects of varying muscle activation in the non-atrophy model, we ran three forward simulations. The front spring force was kept consistent at 300N throughout these tests. Quadriceps and hamstring activation was varied as shown in Fig. 14. The quadriceps were active during the initial 30% of the rowing cycle whereas the hamstrings were active for the remainder of the cycle. These activation levels were controlled such that the starting position and the finishing position of the seat remained constant. Range of motion of the seat changed. Results were normalized by body weight. We observed that when the muscle activation levels were increased, the foot force and the knee force increased. However, higher hamstring activation reduced foot force in the latter half of the cycle.
Figure 14 Total Foot Reaction Force and Knee Joint Reaction Force over one rowing cycle with varying muscle activation and constant front spring force of 300N in Non-Atrophy Model.
4.4 Effect of varying muscle activation and spring forces on Foot Force, Knee Force and Motion in Non-Atrophy Model

We also analyzed the combined effect varying muscle activation and spring forces in the non-atrophy model. The front spring force, quadriceps activation and hamstring activation were controlled such that the motion remained consistent. Varying spring forces and muscle activation levels are shown in Fig. 15. Results were normalized by body weight. We noticed that as the activation levels went up and front spring force lowered, the foot force and the knee force significantly rose. During the latter half of the cycle, these forces followed a fairly similar curve. However, the results in the non-atrophy model were comparable to the able-bodied rower and not the FES rower.

4.5 Effect of varying muscle activation and spring forces on Foot Force, Knee Force and Motion in Atrophy Model

Combined effect varying muscle activation and spring forces was examined in the atrophy model as well. The front spring force, quadriceps activation and hamstring activation were controlled such that the motion remained consistent. Varying spring forces and muscle activation levels are shown in Fig. 16. Results were normalized by body weight. We observed a similar trend as seen in 4.4. As the activation levels went up and front spring force lowered, the foot force and the knee force went up. Similarly, as seen in the non-atrophy model, the forces in the latter half followed a similar curve for all three tests in the atrophy model. Lastly, the forces seen in this model were in the range of those recorded by the FES rower with SCI.
Figure 15 Total Foot Reaction Force and Knee Reaction Force over one rowing cycle with varying front spring force and quad activation while keeping the rowing motion consistent in Non-Atrophy Model
Figure 16 Total Foot Reaction Force and Knee Reaction Force over one rowing cycle with varying front spring force and quad activation while keeping the rowing motion consistent in Atrophy Model.
Chapter 5.

Discussion, Conclusion and Future Work

5.1 Discussion and Conclusion

*Non-Atrophy Model*

The forces observed in this model corresponded to those recorded by able bodied rowers. Although, the front spring forces help the rower move faster and have a bigger range of motion, it does not affect the forces on foot and knees significantly. This suggests that the use of assistive springs may not have therapeutic effect on preventing osteoporosis.

Higher muscle activation does provoke a significant increase in the foot and knee forces. Although, higher hamstring activation leads to a lower foot reaction force in the latter half. This shows that higher muscle activation can have a positive effect on the foot reaction only during the drive phase.

When both, change in muscle activation and spring forces, are incorporated at the same time, significantly better results are seen. This also subdues the effect of higher hamstring activation on the foot force in the latter half of the cycle keeping it fairly consistent through all the trials. This is indicative of the fact that if FES rowers rely less on assistive springs and more on stronger muscle activation to generate motion, there may be a therapeutic effect on osteoporotic bone.

*Atrophy Model*

After SCI there is severe loss and atrophy of muscle due to inactivity. The force generating ability of muscles is affected in individuals with SCI. When this was reproduced in the model by lowering the peak isometric force of muscles, the results seen in forward simulations were comparable to that recorded by FES rowers. This model responded to varying muscle activation and spring forces in a similar way that the non-atrophy model did. The foot and knee forces were in the range of subject FES rowers. Further investigation using this model may help explore the therapeutic effect on bone health and prevention of disuse osteoporosis.
Forward dynamics has been instrumental in the evaluating the cardiorespiratory effect in FES cycling. [13] It has also been used to study the effect of parameters such as pedaling rate, seat position and muscle stimulation pattern on FES cycling performance. [19] However, it has not been used to study its effect on joint loads. FES rowing is a novel method of providing a cardiorespiratory workout to individuals with SCI. Previous studies indicate that it may have a potential to address disuse osteoporosis. It is justifiable to use a muscle driven simulation to analyze to the effects of parameters that influence FES rowing because the approach functions on muscle stimulation that can be controlled externally.

Although this study provides an insight on how muscle excitation and spring forces can affect FES rowing and its ability to address disuse osteoporosis, there are some limitations that need to be considered. The model may not best represent the physical foot constraint. While the foot is fixed to a point but allowed to rotate in the sagittal plane, there is no control on the angle it can rotate. This may produce foot forces that are not accurate.

Handle force are applied on the model through the pelvis and not through the arms of the rower. This may result in slight inaccuracy in the foot and knee forces. The spring forces imparted on the model are time dependent. But in reality, these springs contract and relax depending on the position of the seat. We discovered that modeling muscle atrophy resulted in lower passive forces. But the way it is modeled may not best represent the atrophy in SCI patients.

In spite of these limitations, forward dynamics can non-invasively provide means to assess forces in the knee. This is critical since most low energy fractures in SCI patients are reported around the knee. If the therapeutic effect of FES rowing on bone health is investigated further using forward dynamics, the risk of these fractures can be mitigated.

5.2 Future Work

Musculoskeletal Modeling

Although the models in this thesis provided valuable insight on muscle driven simulations of FES rowing, refinement of some of the elements of the model would lead to a more accurate representation of the FES rower. A constraint defined at both the toes and heels instead of pin joint at the toes would replicate the real life physical foot constraint. In the current models the spring forces are defined with respect to time. Developing a model where the spring forces are defined with respect to the position of the seat is necessary.
Post SCI the extensor muscles of the lower extremities develop spasticity due to excessive muscle spasms. [Management of Spasticity After Spinal Cord Injury: Current Techniques and Future Directions]. The next approach would entail incorporating spasticity in the model and how its parameters affect the motion and forces. Muscle atrophy in current model is designed by lowering the peak isometric force to 50%. A more accurate way of modeling this atrophy based on clinical data would be beneficial.

Reproduction of simulation in experimental setup

As FES rowing is a fairly novel method of rehabilitation for SCI patients, effects of parameters that influence it are not well-known. Reproducing these iterations of forward dynamics simulations in an experimental setup would provide necessary validation and would help explore further possibilities of this therapy. Therapeutic effect of FES rowing on bone health and addressing disuse osteoporosis can be investigated.
REFERENCES


