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IDENTIFICATION OF LOW TORQUE STEP SIZES FOR THE DESIGN OF A SINGLE-CHANNEL MUSCLE-POWERED HYBRID ORTHOSIS FOR PEOPLE WITH SPINAL CORD INJURY

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ABSTRACT

In paraplegia due to complete or incomplete spinal cord injury, the connection from the brain to muscles in the lower limbs is severed but the muscles that act on signals from the brain to produce limb movement remain functional. Functional electrical stimulation (FES), which is the application of electric potential across a muscle group to artificially cause the muscle to contract, is a method that can be used alone or in conjunction with an orthosis to produce a gait cycle. Such FES based walking machines or devices have been studied and designed for several decades. However, their application in everyday exercise is limited by several factors, one of which is the rapid onset of muscle fatigue produced in the stimulated muscle. In this work, simulations were conducted in Simscape Multibody to lay the groundwork for the design of a next-generation FES based walking machine powered by the quadriceps femoris muscle group of each limb. The stimulation of the quadriceps femoris muscle causes the knee to extend while some energy is stored by the orthosis, which uses the stored energy to complete the gait cycle. In this study, we have analyzed the power requirements of each step in the hybrid FES-orthosis gait cycle for different stride lengths. These requirements can help identify small step sizes to reduce the power required from the stimulated muscle.

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NOMENCLATURE

SCI Spinal Cord Injury
FES Functional Electrical Stimulation
ESO Energy Storing Orthosis
CBO Controlled Brake Orthosis
HAT Head and Trunk

INTRODUCTION

An estimated 291,000 people in the United States live with SCI and their number increases by 17,730 people every year. [1] A significant number, 64 percent according to one study [2] , of people with SCI have ranked walking as their top choice.

Current medical technology helps people with SCI walk using several methods including motorized exoskeletons, spinal stimulation, FES implants [3], hybrid orthoses, etc. Motorized exoskeletons are commercially available, the most popular of which are the Indego, EKSO, and ReWalk. Notwithstanding the high costs and weight of a motorized exoskeleton, the key difference between a motorized orthosis and a muscle-powered machine is the physiological benefit of weight-bearing exercise and muscle contraction. When a user is in a motorized exoskeleton, they are able to ambulate which is a purpose also satisfied by a motorized wheelchair. In the case of muscle-powered machines, there is the added psychological benefit that the device is powered by the user's own muscles rather than an external source. A review of FES powered and hybrid devices can be found in [4].

Several iterations of FES hybrid orthoses have been developed at our lab [5–9]. The controlled brake orthosis (CBO) [5] incorporates the stimulation of and the controlled-brake orthosis provided trajectory guidance to the user. After the controlled-brake orthosis came several iterations of the FES-ESO hybrid orthosis which used single channel stimulation combined with an ESO to produce gait [6]. In the ESO, the gait cycle is divided into three phases and four states: The neutral state is the state in which the hip is flexed at 25 degrees and the knee is flexed at 60 degrees. When FES is applied transcutaneously to the quadriceps femoris muscle, the energy harvested from the muscle is used to extend the knee, and is also stored in two energy storage units. One unit is the knee-flexion unit, and the other is the energy-transfer unit. This state is the knee-extension state and the phase moving from the neutral state to the knee-extension state is the FES stimulation phase. Subsequently, the user effectively falls forward while leaning their head and torso (HAT) forward to place their foot on the ground. This is the first stance phase for the first leg. At the end of the first stance phase, the user is at the foot-on-ground state. The user then straightens their HAT with their upper body strength ending in a forward-leaning position, while the energy-transfer-unit transfers releases its stored energy to charge the hip-flexion unit resulting in hip extension. This is the HAT straightening phase. Following this, the user goes through the back-to-neutral phase, when the hip and knee units release their stored energy to respectively flex the hip and knee joints ending with the user in the neutral state.

Two problems with these machines, which were addressed in this study, are the rapid onset of fatigue in the stimulated muscle, and toe clearance. Another problem with FES powered machine is the non-linearity and variability of the FES torque available for the gait sequence. [10] Currently, the device simulates normal walking angles where in the swing phase the hip flexion is 25 degrees and the knee flexion is 60 degrees resulting in a step size of 1.52 meters [9]. Another problem to address in these machines is that during the FES stimulation phase, the toe grazes the ground, and that can result in the user tripping and the energy from the quadriceps muscle being wasted to overcome the friction instead of being stored.

In this study, we aim to reduce the amount of torque required from the quadriceps femoris muscle by reducing the step size while maintaining the condition of toe clearance in the swing phase.

Methods

Figure 1 shows the Simscape model. A simulation for the back-to-neutral phase in the ESO gait cycle was set up in MATLAB Simscape Multibody package. Anthropometric data from [11] was used to model the length and inertia of each limb segment for a person of weight 180lbs and height 5’11”. This set was chosen to be the same as [9] for the sake of comparison. In

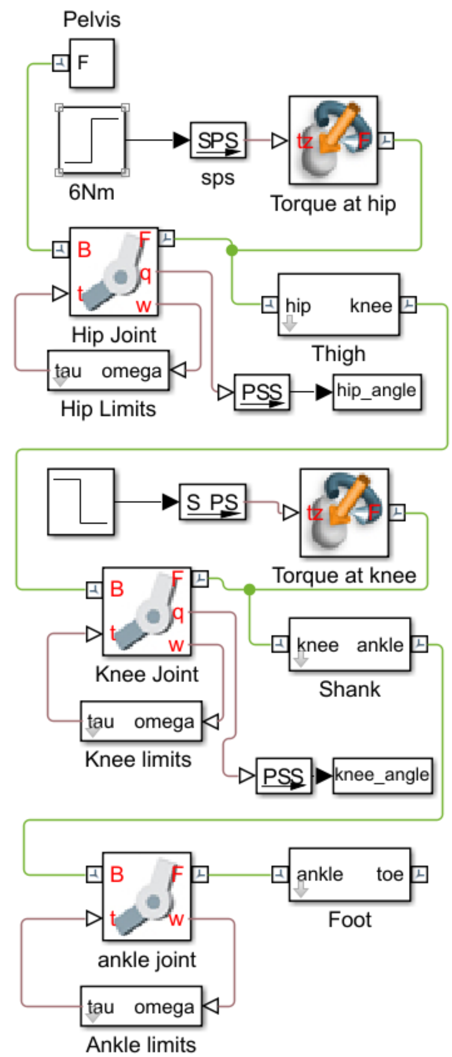

FIGURE 1. SIMSCAPE MULTIBODY MODEL OF 3-SEGMENT LIMB WITH JOINT LOCKS AND CLUTCH

Simscape, each limb is modeled as two reference frames at the two ends of the length of the limb segment with a solid rectangular brick chosen for graphics, but the inertia and center of gravity of the brick are over-ridden with anthropometric values to mimic the dynamics of the limb. The hip, knee and ankle joints are modeled using Revolute Joint blocks because the joints of the orthosis are also revolute joints. Furthermore, a parameter of the Revolute Joint block in Simscape allows setting limits on the joint to represent the rubber bumpers in the actual orthosis as a spring-damper. A unidirectional clutch block was coded in Simscape language to ensure that the limb segment moved only in one direction. The inputs to the block was angular velocity, and the output was a torque, which was fed back to the joint. This torque is the torque required to resist motion, and its value is zero when the joint is moving in the preferred direction, and equal to that of a reaction from a spring-damper similar to the joint limit when it is moving in the locked direction. Ground contact was detected by sensing the translation of the reference frame at the toe from the ground reference frame using a Transform Sensor block. Two sets of experiments were conducted. The first set was to obtain the initial estimates for torque requirements for hip and knee angle combinations ranging from 5 degrees to 30 degrees each using inverse kinematics. A sigmoid profile for the trajectory was provided to the hip and knee joints, while recording the torque required to carry out the motion as an output of the joint. During this set of experiments, neither the clutch nor the limits were used. This set of experiments was also used to reject the hip and knee angle combinations where the toe dug into the ground. However, in our device, the torque is provided by a constant moment arm as the radius of a pulley, and a constant force from a gas spring unlike the controlled variation of an electric actuator. After the initial estimates for torque were obtained for different knee and hip angle combinations, a constant torque of the maximum amount for the selected combination was used to verify if the motion is reproducible using our energy storage system.

Results

The net FES torque requirement is shown as a combination of knee torque and hip torque required in Figure 2. As step length is determined by hip angle alone, hip angle is represented as step length in this figure according to

$$L_{step} = 2 \sin \theta_{hip} \tag{1}$$

Furthermore, negative knee torques are not visible on this chart, and it appears that in some cases only hip torque is being utilized to complete the back-to-neutral phase. In the current context, a negative torque value means that the joint in question does not require torque to move in the requisite direction. This is further observed in the second set of experiments. Figure 3 shows the trajectory generated by applying 7 Nm of torque at the hip, and 0 Nm of torque at the knee joint. The corresponding torque values obtained from the first set of experiments are 8 Nm at the hip joint and -0.01 Nm at the knee joint.

Discussion

We observed that when knee angle was smaller than hip angle, the torque required to flex the knee was negative, which means that the momentum of hip flexion also causes the knee to flex. To avoid the knee from retracing its trajectory due to gravity and inertia, a clutch was implemented. These results were thus

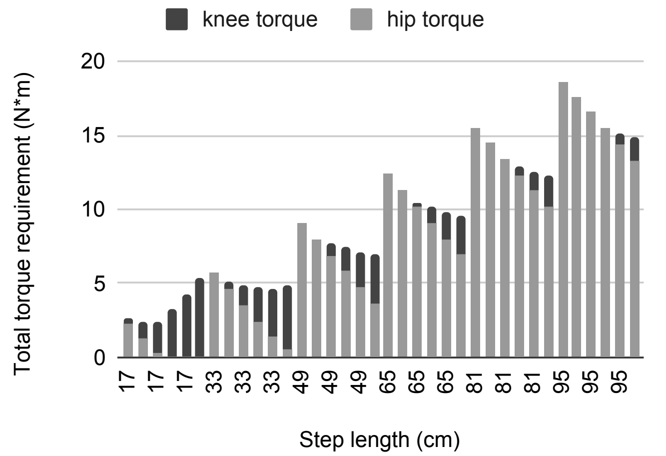

FIGURE 2. TORQUE REQUIREMENTS FOR DIFFERENT STEP LENGTHS AND KNEE ANGLE COMBINATIONS

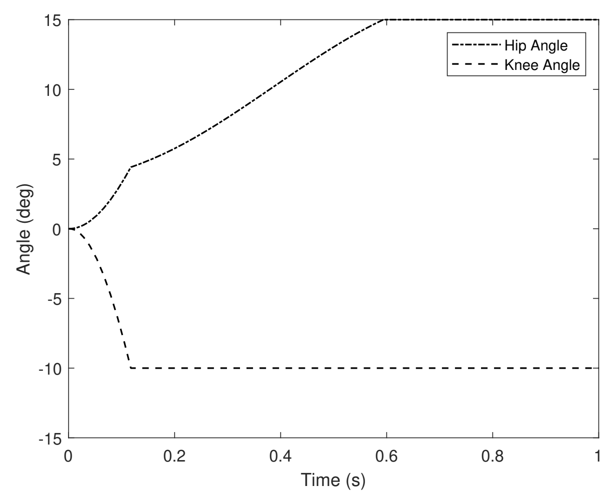

FIGURE 3. HIP AND KNEE ANGLE RESPONSE TO A CONSTANT INPUT OF 7Nm AT THE HIP AND 0 Nm AT THE KNEE JOINT

useful to determine the redundancy of the knee spring required for the back-to-neutral phase and may be eliminated in the next iterations.

For cases where torque was required to flex the knee, with the constraint of a stiff ankle, the toe dug into the ground. This could be remedied with the provision of a dorsiflexion degree of freedom.

For a suitable hip and knee angle combination, we chose the hip angle 15° and knee angle 15° which satisfied a qualitatively sufficient step length of 49 cm, and a total FES torque requirement of 7 Nm. A step length of 49 cm means that the heel of the forward foot is almost the same line as the toe of the rear foot. The foot length used in this experiment is 27.6cm.

Conclusions

We have shown a process to calculate torque requirements for different step lengths, and test the torque requirements for a constant torque input. With this model, we were able to reduce the FES torque requirement from 23 Nm [12] to 7 Nm. For future work, we will implement the values obtained from these experiments to build a bench-top prototype and confirm the ability of the torque mentioned to overcome the losses in the system not covered in the simulation. Additionally, it might also be possible to eliminate the knee flexing spring altogether to result in a lighter orthosis as suggested by both set of experiments.

REFERENCES

[1] National Spinal Cord Injury Statistics Center spinal cord injury facts and figures 2019. https://www.nscisc.uab.edu/Public/Facts Accessed: 2019-11-17.

[2] Brown-Triolo, D. L., Roach, M. J., Nelson, K., and Triolo, R. J., 2002. "Consumer perspectives on mobility: Implications for neuroprosthesis design". *Journal of Rehabilitation Research and Development,* ***39***(6), pp. 659–669.

[3] Popovic, M. R., Curt, A., Keller, T., and Dietz, V., 2001. "Functional electrical stimulation for grasping and walking: Indications and limitations". *Spinal Cord,* ***39***(8), pp. 403–412.

[4] Del-Ama, A. J., Koutsou, A. D., Moreno, J. C., De-los Reyes, A., Gil-Agudo, Á., and Pons, J. L., 2012. "Review of hybrid exoskeletons to restore gait following spinal cord injury". *Journal of Rehabilitation Research and Development,* ***49***(4), pp. 497–514.

[5] Durfee, W. K., and Goldfarb, M., 1992. "Design of a controlled-brake orthosis for regulating FES-aided gait". *Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS,* ***4***(1), pp. 1337–1338.

[6] Durfee, W. K., and Rivard, A., 2005. "Design and simulation of a pneumatic, stored-energy, hybrid orthosis for gait restoration". *Journal of Biomechanical Engineering,* ***127***(6), pp. 1014–1019.

[7] Kangude, A., Burgstahler, B., and Durfee, W., 2010. "Engineering evaluation of the energy-storing orthosis fes gait system". In 2010 Annual International Conference of the IEEE Engineering in Medicine and Biology, IEEE, pp. 5927–5930.

[8] Boughner, K., and Durfee, W., 2014. "Preliminary design of an energy storing orthosis for providing gait to people with spinal cord injury". In 2014 36th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBC 2014, Vol. 2014 of *2014 36th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBC 2014*, Institute of Electrical and Electronics Engineers Inc., pp. 2581–2584.

[9] Katti, V., and Durfee, W., 2018. "Preliminary design and testing of a muscle-powered walking exoskeleton for people with spinal cord injury". 2018 Design of Medical Devices Conference, DMD 2018 ; Conference date: 09-04-2018 Through 12-04-2018.

[10] Durfee, W. K., and MaClean, K. E., 1989. "Methods for Estimating Isometric Recruitment Curves of Electrically Stimulated Muscle". *IEEE Transactions on Biomedical Engineering,* ***36***(7), pp. 654–667.

[11] Winter, D. A., 2009. *Biomechanics and Motor Control of Human Movement: Fourth Edition*.

[12] Project, A., 2018. "Design of a Muscle-Powered Walking Exoskeleton for People With Spinal Cord Injury".

