STABILITY AND GAIT PATTERNS FOR UNSUPPORTED OPEN-LOOP CONTROLLED WALKING

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Abstract

It was the purpose of this study to investigate the feasibility of unsupported FES-controlled walking with open-loop control. Walking was simulated with a two dimensional musculoskeletal model. Muscle stimulation patterns were found by optimizations that maximized the walking duration and required a speed of 1.4 m/s. Stable gait patterns were found, and the gait was robust against small perturbations in muscle stimulation parameters. The gait was typically stiff legged and had an early heel rise. We conclude that open-loop control is sufficient for stability in the sagittal plane, provided that stiff legged gait is acceptable. Additional feedback control may be needed for lateral stability.

Introduction/Background

Most current FES systems for walking employ open-loop control [1]. These systems typically require that the patient uses a walker to maintain balance. On the other hand, computer simulation studies have shown that intrinsic force-length-velocity properties of muscles provide sufficient mechanical stability to produce well-coordinated and unsupported walking movements of limited duration, when constrained to the sagittal plane [2]. This hints at the possibility that stable walking movements of unlimited duration may be possible with open-loop control. Once sagittal plane stability is achieved, lateral instabilities can be effectively controlled by a simple control algorithm for lateral foot placement [3]. This could lead to practical systems for unsupported FES-controlled walking.

In this study, we investigate (1) whether open-loop controlled walking of unlimited duration is theoretically possible, when constrained to the sagittal plane, and (2) to which extent a normal gait pattern can be produced.

Methods

A forward dynamic two-dimensional musculoskeletal model with nine skeletal degrees of freedom was used. The model consisted of seven rigid body segments (trunk, two thighs, two shanks, and two feet), connected by revolute joints. Equations of motion were formulated using SD/FAST (Symbolic Dynamics, Mountain View, CA). The model included eight muscle groups in each lower extremity, each represented by a three-element Hill model (Figure 1). Musculoskeletal properties were identical to those described in [2], with the exception of an alternative formulation of the force-length-velocity properties of muscles. Ground contact was modeled by an array of 10 viscoelastic contact elements on each foot. Equations of motion and constitutive equations for muscles and ground contact were implemented in software written in C.

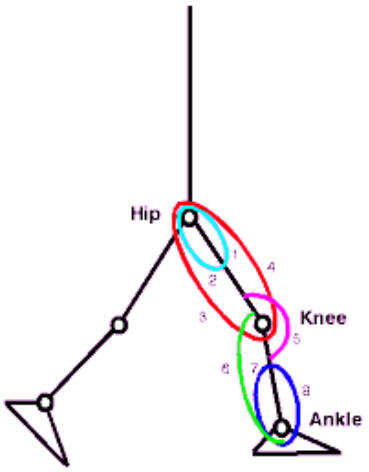

Figure 1: Musculoskeletal model

Open-loop neural control was applied using cyclic block-shaped muscle stimulation profiles that were identical for both legs, with one leg being 50% out of phase. The cycle time was set at 1.2 seconds. Each muscle stimulation pattern was parameterized by its start time relative to the start of the gait cycle, its amplitude and its duration, resulting in a set of 24 parameters describing the control system. These parameters were optimized for the walking task using a simulated annealing (SA) algorithm [4]. Parameters for the temperature reduction schedule in the SA algorithm were set to values recommended in [4], and repeated optimizations were done with different random number sequences in the SA algorithm.

A stability optimization was defined by maximizing the distance walked by the model at a required speed of 1.4 m/s. Simulations were terminated when a fall had not occurred after 15 seconds, which was taken as indication that a stable cyclic solution had been achieved. Gait initiation in the model was aided by a harness suspension system with stiffness decreasing linearly to zero over the first four seconds. This allowed

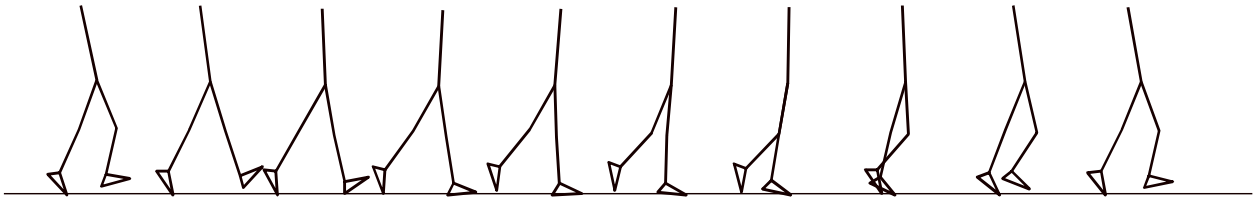

Figure 2: Stick figure representation of one step in a typical optimized walking movement, shown at 15 frames per second.

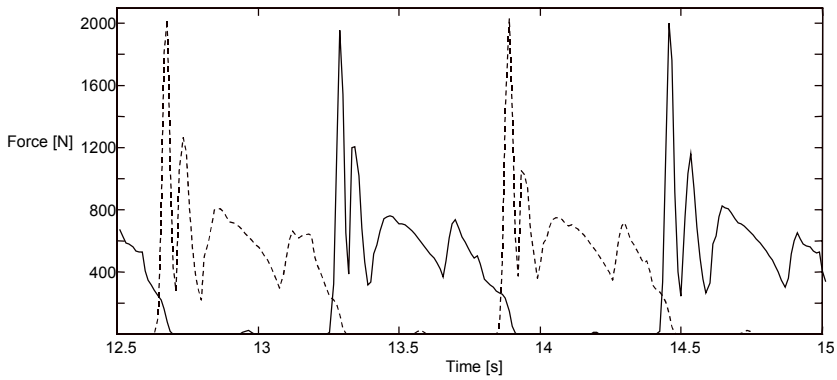

Figure 3: Ground reaction forces during the simulated movement shown in Figure 2.

muscle stimulation patterns for steady state unsupported walking to be found without having to solve the problem of gait initiation.

Solutions that combined stability and efficiency (SE) were found by including a penalty for energy cost in the optimization. Energy cost was defined as the time integral of the muscle stimulation patterns, summed over all muscles.

All results were examined for stability (i.e. whether or not the model could walk for 15 seconds) and gait pattern. Robustness of the gait in the presence of errors in the open-loop muscle stimulation patterns was quantified by Monte Carlo simulations (N=25000) where the stimulation parameters were perturbed by normally distributed random numbers.

Results

Computation time for walking simulations on a Pentium 450 MHz processor typically was 0.7 to 5 times real time, depending on the muscle stimulation patterns. Optimizations with the simulated annealing method typically required 10000 to 50000 simulations, and converged after 2-5 days of continuous computation.

All muscle stimulation patterns found by the stability optimization resulted in symmetric cyclic walking movements of infinite duration, and achieved exactly the required speed. Repeated optimizations, with different random number seeds, resulted in different solutions, which indicates that there were many open-loop muscle stimulation patterns that all resulted in stable walking movements at a given speed.

Gait patterns found by the different optimizations were different, but all showed a stiff-legged toe walking gait: the knee was fully extended throughout the stance phase and the pressure between foot and ground shifted to the forefoot very early in the stance phase. A typical example is shown in Figure 2. Vertical ground reaction forces and muscle forces for this simulation are shown in Figures 3 and 4.

These gait patterns were robust against small perturbations in the neural control inputs. Random perturbations in stimulation timing, with an RMS value of 1% of the gait cycle (12 ms), resulted in a walking gait that was still stable in 54% of the 25000 simulations. Random perturbations in stimulation magnitude, with an RMS value of 10% of maximal stimulation, resulted in 37% of the 25000 simulations still being stable.

Optimizations where energy cost was added to the optimization resulted in stiff legged gait patterns when the energy cost term was given low weighting. Increased weighting of energy cost resulted in solutions that had lower muscle activation, tended to flex the knees during stance, and could never walk for more than 10 seconds.

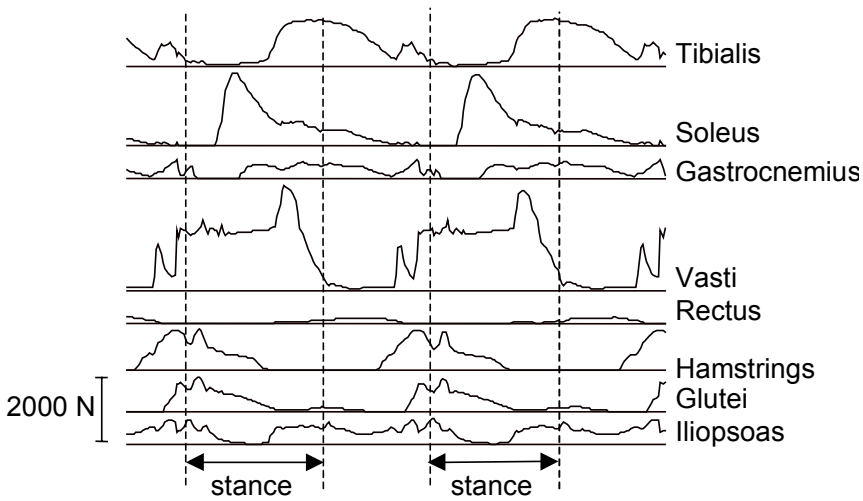

Figure 4: Muscle forces during the simulated movement shown in Figure 2.

Discussion/Conclusions

Our modeling results suggest that open-loop neural control is sufficient for anterior-posterior stability of walking gait. There was considerable robustness against random variations in stimulation magnitude, which suggests that fatigue effects will not compromise the short term stability of the movement. Sensitivity to stimulation timing was greater, but timing should be more easily controlled in an FES system. These conclusions only apply to steady state walking, because gait initiation in the model was aided by a weight support system.

The gait patterns that were found by the optimization were clearly abnormal. There was no motion in the knee joint during the stance phase, while during normal human gait the knee joint has 10-20 degrees of flexion in the knee during stance. The model accomplished this gait pattern by activating the plantar flexors, which causes pressure to move to the forefoot early in stance. This then causes the ground reaction force vector to pass anterior to the knee joint, effectively locking the knee joint and making knee motion insensitive to muscle coordination. Large knee extensor forces were often seen (Figure 4) and this would also contribute to locking of the knee joint. Although the hip and ankle joints still require control, open loop muscle stimulation appears to be sufficient. The ground reaction force (Figure 3) showed excessive impact forces, which may be related to the lack of knee flexion during weight acceptance.

Stiff-legged gait and toe walking may be an effective way to deal with lack of control. This type of gait is often seen in patients with sensory deficits [5]. Ankle foot orthoses, which cause an early shift of pressure to the forefoot, are sometimes successful in patients with neurological problems.

If stiff knee gait is so stable, why do healthy humans not use this type of gait? We propose that humans may flex their knee during stance to avoid shock or because it is more economical. In an attempt to produce such gait patterns, we added a variable related to energy cost to the optimization criterion. When the weighting of energy cost was increased, we saw a tendency to flex the knee during stance, but those simulations always resulted in a fall before 10 seconds. This suggests that feedback control of knee angle will be needed to produce this feature of normal gait with FES.

We conclude that open-loop FES can theoretically produce stable gait in the sagittal plane. We have not investigated lateral stability, but analysis of a passive stiff-legged walking mechanism has shown that lateral foot placement, based on sensory information from the previous step, can control such instabilities [3].

References

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