34th Annual International Conference of the IEEE EMBS
San Diego, California USA, 28 August - 1 September, 2012

Gait Planning and Double Support Phase Model for Functional Electrical Stimulation-based Walking

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***Abstract*— Joint or segment angle trajectories of able-bodied persons are often recorded or mimicked as reference trajectories for walking restoration in paraplegia. In this paper, lower limb segment angle trajectories are computed from simple mathematical models developed to represent functional electrical stimulation (FES) and a novel brace based walking. The new models incorporate the double support and single support phases of walking. Dynamic optimization is utilized to design walking trajectories that minimize muscle activations and arm reaction forces generated from the walker. Compared to the voluntary walking trajectories, the new trajectories are more representative of FES-based walking as only a limited number of muscle are stimulated to compute walking trajectories.**

I. Introduction

Various prototypes of orthoses have been developed and combined with FES (see [1], [2] and references therein) such as reciprocal gait orthosis (RGO) [3], controlled-brake orthosis [4], hybrid neuroprosthesis [1], joint couple orthosis [2] to restore standing/walking in persons with gait disorders [1], [2], [5]. The main advantage of utilizing an orthosis is that it can support body weight during standing/walking, thus excessive stimulation due to FES can be avoided during standing or stance phase of walking.

Speed and efficiency of FES and orthosis-based walking can be improved by using closed-loop control methods compared to open-loop strategies such as the push-button control approach [6]. However, closed-loop control generally requires knowledge of desired or reference trajectories. As these reference trajectories are unknown, FES-based walking restoration employs recorded walking trajectories from able-bodied persons [5]. Walking trajectories in able-bodied humans are generated through the activation of healthy muscles which in the case of persons with mobility disorders may have been atrophied or possess limited strength. Also, FES can only activate a limited number of muscles; able-bodied walking voluntarily recruits more muscles. Due to these aforementioned reasons, tracking a voluntary walking gait may not be feasible or can be energetically costly in rehabilitation of walking. Therefore, new walking trajectories that minimize fatigue due to FES and metabolic energy consumption need to be developed.

Under some conditions able-bodied persons choose their walking pattern through minimizing metabolic energy [7]. However, the nature of mechanisms involved in the human motor control system that minimize metabolic energy consumptions is still an open research question. Nonetheless, this minimization principle can still be practiced by employing the existing engineering methods such as dynamic optimization to compute walking trajectories. In [8], walking trajectories were computed by minimizing a cost function containing muscle activation variables (to minimize FES) and arm reaction forces (to minimize metabolic energy consumption). As calculation of these trajectories required a dynamic walking model, a simple three-link walking model was developed [8]. This model represented single support phase dynamics in sagittal plane and an instantaneous double support phase modeled as an algebraic equation derived through the principle of conservation of angular momentum. However, an instantaneous double support phase (DSP) model is not an appropriate representation as able-bodied persons spend 20% (10% in each of the two DSPs) of the gait cycle in DSP. Moreover, in persons using FES-based walking systems walking speed is significantly slower than able-bodied persons which implies greater time spent during DSP of the gait cycle.

In continuation of our earlier effort in [8], this paper focuses on incorporation of finite (instead of instantaneous) double support phase in the walking model. A further objective of this paper is to design new optimal walking trajectories (without utilizing normal gait data) that reduce muscle fatigue and minimize metabolic energy consumption using the new model. The developed human walking model accounts for muscle activations via FES, a knee ankle foot orthosis (KAFO) system, a walker, and finite DSP. The model includes muscle dynamics required to simulate hip and knee torques evoked through FES during the single support phase of walking. The optimization toolbox in MATLAB was utilized to solve the constrained nonlinear minimization problem and thus, find optimal stimulation and force profiles to produce walking. Algebraic impact equations were also derived to obtain a transition rule between the end of single support phase and the beginning of final DSP. Five walking steps were simulated to show the convergence of the optimization method to produce repeatable optimal walking trajectories.

II. Three-link brace walking model

A dynamic walking model was developed to represent walking using a KAFO, FES, and an assistive device such as walker. For model development, the following assumptions and properties of the components in the FES-based walking system were used. The trunk of the user can be stabilized by the walker and thus, its dynamics were neglected in the model. The user pulls against the walker to produce required

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978-1-4577-1787-1/12/$26.00 ©2012 IEEE

propulsion to step forward and this reaction force acts at the hip of the user. The upper body (i.e., head, arm and trunk (HAT)) is considered as a point mass. The KAFO provides a rigid link around the ankle joint during the swing phase of the motion and limited rotation during the DSP of the motion. The KAFO provides kinematic constraints and bears the user's weight during standing and walking. The walking movement is considered only in the sagittal plane. Also, the KAFO system locks the knee joint just before the stance phase (i.e., when the swing leg is about to hit the ground) and remains locked during the stance phase. Similarly, during DSP the KAFO locks the knee joint of both legs. When swing phase is initiated, the knee joint is unlocked. Each leg has its own KAFO and the two braces work independently (e.g., the locking/unlocking of each brace is controlled by a push-button). The hip and knee muscles can be stimulated using electrodes (surface or percutaneous) to produce hip/knee flexion and extension. The walking model is divided into three phases: 1) *initial double support phase*, 2) *single support phase*, and 3) *final double support phase*. The terminology used for the model phase definitions is different from the definitions in the gait analysis literature.

A. Initial double support phase

Double support phase (DSP) is modeled as a closed-link chain as both legs are touching the ground. Since the brace locks the knee joints of both legs, each leg is considered as a single segment. The model has joints at hip, ankle, and toe during the initial DSP (see Fig. 1 (a)). The equations of motion are generated through the Lagrangian method. Essentially, the three link system is reduced to a single degree of freedom link chain. The equation of motion during the DSP are given as:

$$J\ddot{q_d} = C(q_d, \dot{q_d}) + G(q_d) + M_w, \tag{1}$$

where J is the inertial element and $\ddot{q_d}$ contains angular accelerations of one of the legs, C contains Coriolis terms, G contains gravitational components, and M_w is the moment generated by the arm reaction force F_w (see Fig. 2 and is defined as

$$M_w = F_w l,$$

where l is the total length of the locked segment. The equation of motion during the DSP was derived by considering the double support phase as a closed-link four bar mechanism with the assumption that the feet does not slip during the movement. The Lagrangian method was followed to derive the equation of motion of the four-bar mechanism. The equation of motion was parametrized by only one angle q_d using an approach described in [9]. Details of the equation and its derivation are available upon request to the authors.

B. Single support phase

Since the brace locks the knee joint during the stance phase, the stance leg is considered as a single segment rotating about the ankle joint as shown in Fig. 1 (b) The leg

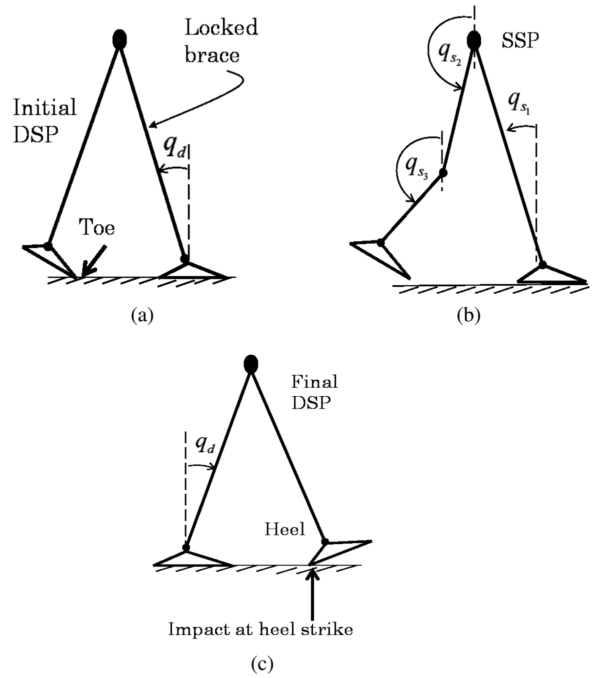

Fig. 1: (a) Initial double support phase (DSP) model: The model has three joints at the ankles of both legs, hip, and toe of the trailing leg. (b) A three-link model representing single support phase (SSP). The leg in stance phase is considered as a single segment as the brace locks the knee joint while the leg in swing phase is considered as two segments as the brace is unlocked. (c) Final DSP model: The model has three joints at ankles of both legs, hip, and heel of the forward leg.

in the swing phase is considered as a two link segment with joints at knee and hip only while the ankle joint is considered as rigid (i.e., foot on the contra-lateral leg is fixed with the lower leg segment). The angle which the leg in stance phase makes with the vertical is denoted as q_{s_1} as shown in the Fig. 2. The thigh angle and the shank angle of the leg in the swing phase are denoted as q_{s_2} and q_{s_3}, respectively (see Fig. 1). The details of the dynamic equation for single support phase (SSP) are given in [8].

C. Impact phase

After the completion of swing phase and just before the start of final DSP, the impact of the swing leg with the ground is accounted in the model. In Fig. 1, the stance angle before and after the heel strike is denoted as $q_{s_1}^-$ and q_d^+, respectively. The thigh and shank angles just before the heel strike are denoted as $q_{s_2}^-$ and $q_{s_3}^-$, respectively. However, as the brace locks just before the impact, the thigh and shank angles just before the heel strike are considered to be equal (i.e., $q_{s_2}^- = q_{s_3}^-$). Since the final DSP model has one degree of freedom, $\dot{q}_d^+$ is the only unknown variable. With the assumption that the angular momentum about the impact point is conserved [10]), the following algebraic equation can

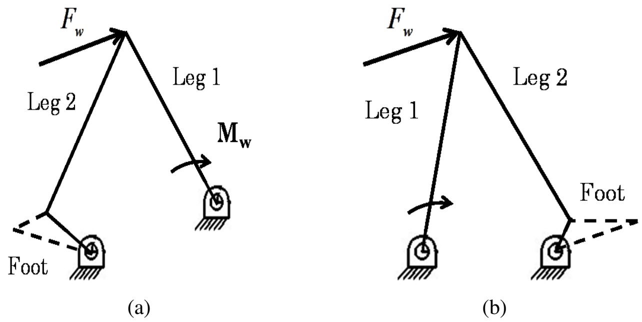

Fig. 2: Four-link closed chain representation used to derive dynamic equations for initial (a) and final DSP (b) model.

be derived:

$$\dot{q}_d^+ = \frac{H^-}{H^+}, \tag{2}$$

where H^- is the function dependent on $q_{s_1}^-$, $q_{s_2}^-$, $\dot{q}_{s_1}^-$, $\dot{q}_{s_2}^-$ and $H^+$is the function dependent on q_d^+. $q_d^+$is determined from the geometrical configuration (i.e., $q_{s_1}^-$, $q_{s_2}^-$) at the end of SSP. Details of the equation in (2) and its derivation are available upon request to the authors.

D. Final double support phase

The final DSP is modeled as shown in Fig. 1 (c). The model has joints at the hip, both ankles, and heel and its equation of motion can be derived as in (1).

III. Optimization results

The objective was to find the trajectories that minimize a function that contains cost on muscle activations of hip/knee flexor and extensors and arm reaction force from the walker. The goal of the optimization problem was to minimize objective function Π

$$\Pi = \int_{t_0}^{t_1} F_w^2 dt + \int_{t_1}^{t_2} (F_w^2 + u_{h_f}^2 + u_{h_e}^2 + u_{k_f}^2 + u_{k_e}^2) dt + \int_{t_2}^{t_f} F_w^2 dt,$$

where u_i, $(i = h_f,\, h_e,\, k_f,\, k_e)$ are the muscle stimulation variables of hip flexors, hip extensors, knee flexors, knee extensors [8], F_w is the arm reaction force [8], and t_0 is the initial time of the step, t_1 is the end time of the initial DSP, t_2 is the end time of the single support phase, and t_f is the time taken to complete one step. The optimization is subject to the following constraints

$$q_d(t_0) \ = \alpha_1, \quad q_d(t_f) = \alpha_2,$$

where α_1, α_2, are known constants. Further, constraints were applied to avoid knee hyperextension and ground penetration by the foot. Optimization for five steps was performed by utilizing the function fmincon in the optimization toolbox in MATLAB. For the first step, the initial angular velocity (i.e., $\dot{q}_d$) was zero. For the succeeding steps, the initial angular velocity was obtained from the final angular velocity of the leg 2 in the preceding step. α_1, α_2 were chosen as $15°$ and $-15°$, respectively. These values for α_1, α_2 were chosen as these angles provided minimum cost function value per distance. Stimulation and force profiles for each step during the single support phase were divided into ten grid points where the profiles were linearly interpolated between the grid points. In each DSP, the force profiles for each step were divided into six grid points. The mass, length, and inertia parameters in the equations (1) and (2) were obtained from [11]. The computed trajectories and their corresponding stimulation and force profiles are shown in the Figs. 3, 4, and 5, respectively.

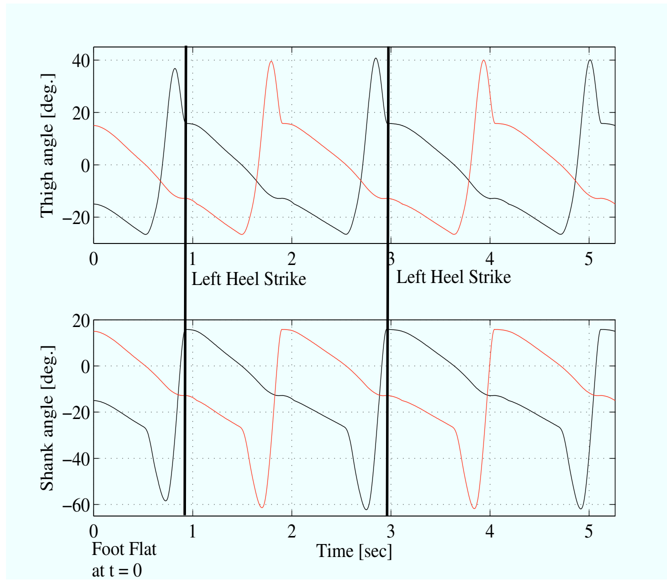

Fig. 3: Thigh and shank segment angle trajectories over five steps. The left leg is shown in black solid line (thin) and the right leg is shown in red solid line. The vertical black solid lines denote the time when left heel strike occurs (LHS).

IV. Discussion

In able bodied persons, the DSP approximately equals 20% of the gait cycle and it tends to become higher in the case of walking with a FES-brace system. In general, the existing models represent DSP as an instantaneous phase or require ground reaction models for modeling dynamics. In the paper, the main motivation behind modeling DSP was to provide the correct representation of a finite DSP in the gait cycle of able bodied persons and paraplegics using an FES-brace walking. We also made efforts to model the DSP in a simple manner through minimizing the degree of freedom in the model and considering motion only in the sagittal plane. Particularly, the model can be visualized as a closed four-bar mechanism (assuming that during DSP the feet do not slip) with ground links at the one of the ankle joints and at the heel or toe depending on the initial or final DSP. Further, the dynamics were reduced to minimal coordinates (only one angle, q_d) by using the approach given in [9]. Another benefit of the closed four mechanism formulation is that it does not require ground reaction models (e.g., spring damper models) to simulate the dynamics. Thus, this paper deals with the problems in the existing models

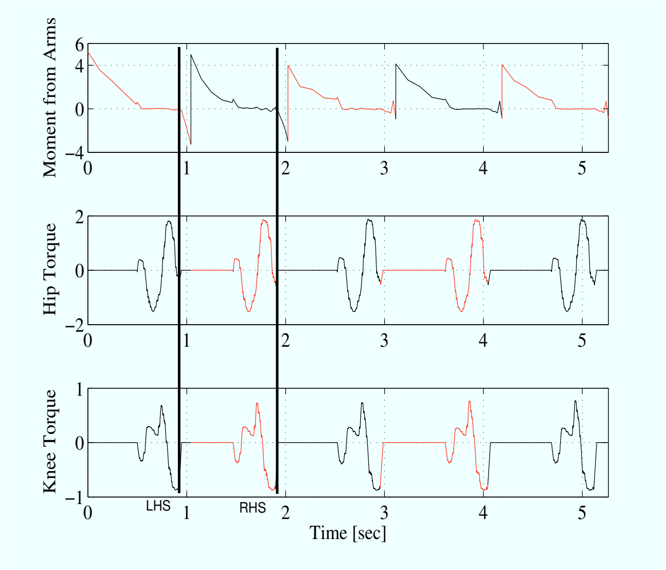

Fig. 4: Torque and moment (moment from the walker, M_w) profiles of right leg (red solid line) and left leg (black solid line) over five steps. The vertical black solid lines denote the time when left heel strike (LHS) and right heel strike (RHS) occur.

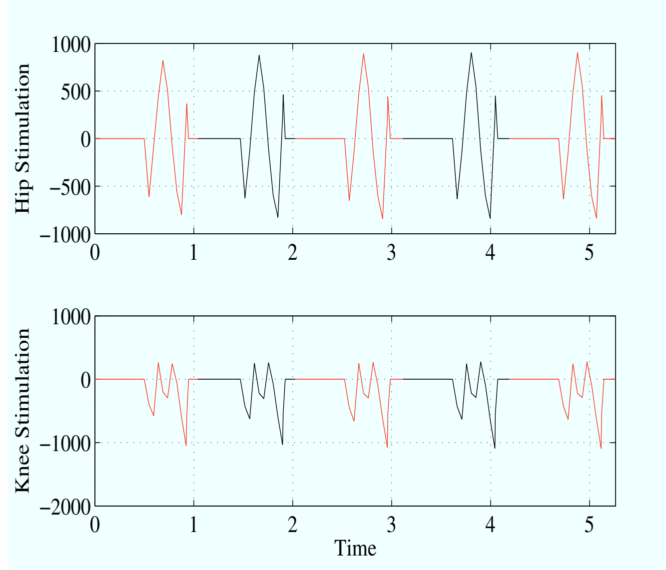

Fig. 5: Stimulation intensity profiles (in arbitrary units) of five steps. Stimulation profiles that generate hip extension and knee flexion are shown as negative values

by developing a simple DSP model for FES-brace walking which does not require a ground reaction model. Our last objective was to utilize the developed models in this paper and in [8] to generate optimal segment angle trajectories and torque/moment profiles via dynamic optimization. The optimization results shown in Figs. 3, 4, and 5 depict that the trajectories tend to converge to an equilibrium after the third step. Nonetheless, these models will be verified through experiments on the paraplegics using the FES and electronic KAFO (modified KAFO bought from Otto Bock, USA) system. Upon verification, we plan to utilize these models for modifying/improving FES-brace system and improving walking function of paraplegics through incorporation of closed-loop control strategies.

V. Conclusion

A simple double support phase model was developed to represent paraplegic walking utilizing FES and a KAFO. The double support phase was reduced to the four bar closed linkage and then the Lagrangian method was employed to obtain the dynamic equation in (1). These models were integrated with an earlier developed single support model in [8]. These combined models were utilized to obtain convergent optimal walking trajectories by minimizing muscle activations and arm reaction force from the walker. Future efforts will focus on model verification and utilizing these models to obtain optimal step lengths and step frequencies for the hybrid brace-FES walking.

VI. ACKNOWLEDGMENTS

This research was supported by an Alberta Innovates Health Solutions Postdoctoral Fellowship Award. We thank Dr. Vivian Mushahwar for helpful suggestions.

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