33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

A method to simulate motor control strategies to recover from perturbations: Application to a stumble recovery during gait

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Abstract— **Perturbations during human gait such as a trip or a slip can result in a fall, especially among frail populations such as the elderly. In order to recover from a trip or a stumble during gait, humans perform different types of recovery strategies. It is very useful to uncover the mechanisms of the recovery to improve training methods for populations at risk of falling. Moreover, human recovery strategies could be applied to implement controllers for bipedal robot walker, as an application of biomimetic design. A biomechanical model of the response to a trip during gait might uncover the control mechanisms underlying the different recovery strategies and the adaptation of the responses found during the execution of successive perturbation trials. This paper introduces a model of stumble in the multibody system framework. This model is used to assess different feedforward strategies to recover from a trip. First of all, normal gait patterns for the musculoskeletal system model are obtained by solving an optimal control problem. Secondly, the reference gait is perturbed by the application of forces on the swinging foot in different ways: as an instantaneous inelastic collision of the foot with an obstacle, as an impulsive horizontal force or using a force curve measured experimentally during gait perturbation experiments. The influence of the type of perturbation, the timing of the collision with respect to the gait cycle, as well as of the coefficient of restitution was investigated previously. Finally, in order to test the effects of different muscle excitation levels on the initial phases of the recovery response, several muscle excitations were added to selected muscles of the legs, thus providing a simulation of the recovery reactions. These results pave the way for future analysis and modeling of the control mechanisms of gait.**

Keywords: gait simulation, gait control, stumble, recovery strategies.

I. Introduction

Falls during gait or stance result in a number of problems, ranging from psychological to physical, such as hip fracture, that affect frail populations such as the elderly [1]. Falling during gait occurs after an unsuccessful recovery attempt after some kind of gait perturbations such as a trip, a slip or a step down [2]. A trip or stumble is a perturbation resulting from the collision of the swinging foot with an obstacle. Two major recovery strategies have been reported in the literature, the lowering and the elevating strategies [3], [4]. The lowering strategy consists of quickly lowering the perturbed foot after the trip and is often immediately followed by a step of the contralateral leg. This strategy is observed when the trip occurs in mid or late swing. The elevating strategy is characterized by an elevation of the swing leg after the trip and is usually observed for perturbations occurring in early swing.

The recovery strategy success seems to be largely dictated by the ability of the body to counteract the forward inclination of the trunk [5], [6]. A successful recovery strategy prevents the body from reaching a critical inclination that might lead to a fall. This is achieved by means of an external counteracting moment. During gait, any external moment is due exclusively to foot-ground contact forces and the point where they are applied. Therefore, the time and position of the subsequent foot contact after the trip have a major influence on the outcome of the recovery.

In this paper a musculo-skeletal model of a trip during gait is proposed. This model is capable of predicting the response, including time and position of the next foot contact, in the first hundreds of milliseconds after a simulated trip. These simulations are performed on a detailed musculo-skeletal model through a three-step procedure. Firstly, the normal gait patterns, assumed as periodic, are reproduced by solving an optimal neuromuscular control problem minimizing an appropriate cost function [7]. Secondly, the reference normal gait obtained previously is perturbed. The perturbation (trip) is modeled by the application of forces on the foot during swing, such as an instantaneous collision of the foot with an obstacle, resulting in an instantaneous jump in the generalized velocities of the model. The dynamic response of the model after the trip is then predicted by forward integration starting from the new state of the model just after the impact. In the first simulations, the normal gait muscle excitations predicted by the optimization procedure were left unchanged. The use of the normal neural excitations in the simulation following the trip is acceptable only during the first few hundreds of milliseconds after the perturbation, due to the delay in the muscle response. For instance, long-latency responses start after 100 ms [8] and these delays can be modified with learning [9]. This model of the stumble is used to investigate the influence of different perturbation parameters such as the time of collision with respect to the gait cycle and the nature of the collision (toes or ankle collision, and coefficient of restitution) on the time and position of the next foot placement after the trip.

Manuscript received April 15, 2011. This work was supported in part by the Fundação de Amparo a Pesquisa do Estado de São Paulo (FAPESP, grant number 2010/17181-0).

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Finally, the model is used to assess the possible mechanisms of recovery strategies based on directly altering muscle excitation levels. It was shown previously that the recovery strategies were largely determined by the "uncontrolled" body dynamics in the first hundreds of milliseconds after the perturbation [10]. With the addition of a simple recovery strategy, e.g. increasing joint impedance, it is hypothesized that the body configuration at the subsequent double stance phase will be more favorable to reject the perturbation [6].

II. Material and Methods

This section introduces briefly the experimental results that guided this research and details de musculo-skeletal model used to interpret and explain these data.

A. Experimental data

Five young and healthy male subjects participated in a series of trip experiments. While a subject was walking comfortably on a treadmill at 1.1 m/s, an unexpected blockage of the left ankle was applied with controlled onset and duration as fully described in [4]. A rope attached at one end to the subject's ankle and at the other end to a locking mechanism with a load cell was used to apply a perturbing force for approximately 200 ms. The movement was measured with a video system while the perturbing force was measured by a load cell as shown in the Figure 1. The perturbation data were compared to normal gait data recorded during the same session. Although stumbling on treadmill is not exactly the same as tripping on the ground, the responses were identical [4].

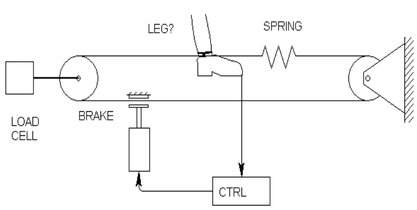

Fig.1. System designed to elicit a experimental stumble and measure the perturbing force. See [4] for details.

The horizontal force applied to the ankle of a subject to elicit an experimental trip was recorded separately. The force data set selected for this study corresponded to a subject whose anthropometric characteristics are similar to those of the model utilized in the simulations. The results are then compared to other two simulations using different contact formulations, which will be described as: 1) impulse with same magnitude as integral of applied force over time acting at the instant of initial perturbation; 2) collision at same instant modeled as inelastic with restitution coefficient $e = 0$.

B. Musculoskeletal Model

A planar musculoskeletal model of the body [11] consisting of seven rigid body segments (trunk, thighs, shanks and feet) with f = 9 degrees of freedom was used. The equations of motion are presented in (1):

$$\mathbf{M}(\mathbf{y})\,\ddot{\mathbf{y}} + \mathbf{k}(\mathbf{y},\dot{\mathbf{y}},t) = \mathbf{q}(\mathbf{y},\dot{\mathbf{y}},t) \quad (1)$$

where $\mathbf{y}$(t) is the vector (dimension f) of generalized coordinates, $\mathbf{M}$ is the mass matrix, $\mathbf{k}$ is the vector of Coriolis and gyroscopic forces and $\mathbf{q}$ is the vector of generalized applied forces and includes the muscle forces. Eight muscle groups are included in each lower limb: Iliopsoas, Glutei, Hamstrings, Rectus Femoris, Vasti, Gastrocnemius, Soleus, and Tibialis Anterior. Each muscle is represented by a three-element Hill-type muscle model and includes the first order activation dynamics and the first order contraction dynamics of each muscle [12] with muscle properties extracted from [11]. The complete musculoskeletal model has a total of 50 states in $\mathbf{x}$: 9 generalized coordinates, 9 generalized velocities, 16 muscle contractile element lengths (lce), and 16 muscle activations. The dynamics of the musculoskeletal system reads as

$$\dot{\mathbf{x}}(t) = \mathbf{f}(\mathbf{x},\mathbf{u}), \quad (2)$$

where $\mathbf{u}$ are neural excitations to the muscles. The interaction between feet and ground is modeled by means of 10 nonlinear spring-damper elements uniformly distributed along each foot sole [7].

C. Reference Gait

The reference normal gait patterns for the model were obtained by solving an optimal neuromuscular control problem, see [7] for details. This problem consists of searching for time histories of controls $\mathbf{u}$(t) and states $\mathbf{x}$(t) that minimize a cost function J, and satisfy the musculoskeletal dynamics Eq. (2) and constraints that guarantee periodicity of gait and physiological muscle forces ($0 < \mathbf{u} < 1$). The cost function utilized was composed by two terms, one quantifying the deviation of model kinematics and ground contact forces from experimental data available in [13], and the other penalizing muscle activations squared. The average walking speed was prescribed as 1.1 m/s. The resulting optimal control problem was transformed into a large-scale Nonlinear Programming problem using direct collocation ([7], [14]) and solved using the SNOPT package, a large-scale, sequential quadratic programming optimization code for Matlab (Tomlab Optimization Inc., Pullman, WA).

D. Modeling of the trip

There are different procedures to model the trip. An instantaneous, frictionless collision of the swinging foot is, perhaps, the most straightforward method to model the trip. The collision yields a discontinuity in the generalized coordinates when the normal relative velocity of the contacting points is different than zero [15], [16]. In this case, according to [16], the normal impact force F can be added to Eq. (1) as

$$\mathbf{M}\ddot{\mathbf{y}} + \mathbf{k} = \mathbf{q} + \mathbf{w}_N F, \quad (3)$$

where $\mathbf{w}_N$ projects the generalized velocities on the normal direction of impact. Thus, the impact results in a jump in the velocities while the position remains unchanged. This can be expressed as in Eq. (4):

$$\lim_{te \to ts} \int_{ts}^{te} (\mathbf{M}\,\ddot{\mathbf{y}} + \mathbf{k} - \mathbf{q} - \mathbf{w}_N F)dt = \mathbf{M}(\dot{\mathbf{y}}_e - \dot{\mathbf{y}}_s) - \mathbf{w}_N \Delta P = 0, \quad (4)$$

where the indices s and e refer, respectively, to the start and the end of the impact. The finite force impulse ΔP reads as:

$$\Delta P = \lim_{te \to ts} \int_{ts}^{te} (F)dt \quad (5)$$

Considering that the impact duration is infinitesimal, $\mathbf{M}$ and $\mathbf{w}_N$ are constant in Eq. (4) and all the other generalized forces vanish when compared to the magnitude of the impact force F. This equation can be used directly to compute the generalized coordinates just after the impact when an approximation of ΔP is available from the time course of the impact force. This is the case, for instance, in the experiments described in [4].

When ΔP is unknown or not readily available, an alternative is using the kinetic coefficient of restitution e [15], which is defined as the ratio of the impulses in the compression and in the restitution phases of the impact [16]. An impact with $e = 1$ indicates no energy loss and is called elastic, whereas an impact with $e = 0$ indicates maximal energy loss and is called inelastic, see [16] for details.

E. Simulations

A series of trips were simulated at instants corresponding to multiples of 10% from 10% to 90% of the swing phase of the reference normal gait, where the swing phase is the period between toe off (0%) and the next heel contact (100%).

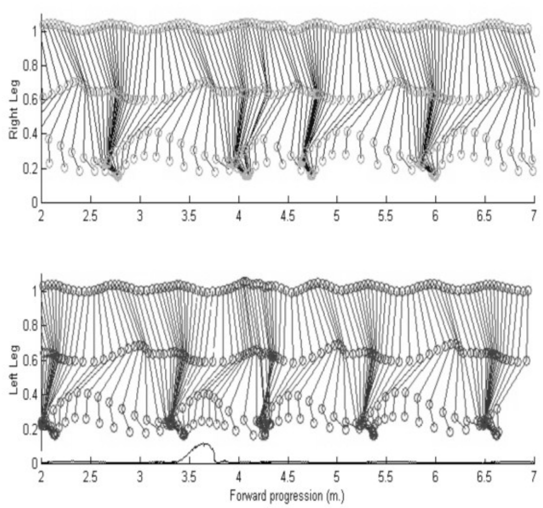

Fig.2. Stick diagram of right (above) and left (below) legs. Perturbation force was applied at early left swing (solid line below).

In order to investigate the influence of the collision nature on the results, collisions were simulated at the ankle joint and at the toes, with the coefficient of restitution assuming the boundary values $e=0$ (inelastic) or $e=1$ (elastic). For each one of these conditions, trip was modeled as a frictionless, instantaneous collision against a fixed obstacle, with generalized velocities just after the collision. Initially, body motion after the collision was predicted by applying the unchanged, optimal neural excitations computed for the reference gait pattern.

In this work the forces measured experimentally were used to simulate a trip on the model and the results compared to other two simulations using different contact formulations: 1) impulse with same magnitude as integral of applied force over time acting at the instant of initial perturbation; 2) collision at the same instant modeled as inelastic with coefficient of restitution $e = 0$.

Finally, one method to simulate the trip was chosen to analyze different recovery strategies. These strategies were implemented as increases in the muscle excitation of the model after the trip. An excitation of the uniarticular flexors of the perturbed leg and of the uniarticular extensors of the contralateral leg is presented to illustrate the procedure. The recovery reaction was composed of a small (25% of the maximum) increased background activity of the extensors of the stance leg (Glutei, Vasti and Soleus) and of the flexors of the swinging leg (Iliopsoas and Tibialis Anterior).

III. RESULTS

The experimental response to a trip during early swing is presented in Fig. 2. The subject tried to elevate the perturbed limb on the treadmill band, then performed a step that resulted in a slightly shorter length and, finally, prepared a compensation step with the contra-lateral leg as shown in the stick diagrams (Fig. 2). It is critical that the left foot is placed before the right one in order to avoid a non-stable configuration of the body. Fig. 3 shows the results for the comparison of different contact formulations: a) normal unperturbed gait; b) trip with applied perturbation force F(t) measured experimentally as shown in Fig. 2; c) trip with perturbation modeled as an instantaneous impulse computed as in Eq. (5); d) trip with perturbation modeled as an instantaneous, inelastic, frictionless collision with coefficient of restitution ($e=0$). The inelastic frictionless collision was chosen as the reference because it replicated the real conditions of the trip with sufficient realism and with lower computational costs. Fig. 3 also underscores that after a perturbation, if there is no change in the muscle excitation patterns, the body ended a step too short to avoid a fall.

A tentative recovery strategy is presented in Fig. 4, which shows a simple recovery reaction, composed of a small increased background activity (25% of the maximal) of the pure extensors of the stance leg (Glutei, Vasti and Soleus) and of the pure flexors of the swinging leg (Iliopsoas and Tibialis Anterior).

IV. DISCUSSION

The results showed that with a very small adaptation of the excitation patterns of the musculoskeletal model aimed at copying the recovery reaction, the configuration of the body at the following foot contact is improved. The results underscore the importance of the initial stages of the perturbation in the final outcome of the recovery as has been suggested in previous research [4], [5]

It is very important to model adequately the perturbation, as shown in Fig. 3, more specifically, when the perturbation is of short duration as typically occurs during a trip.

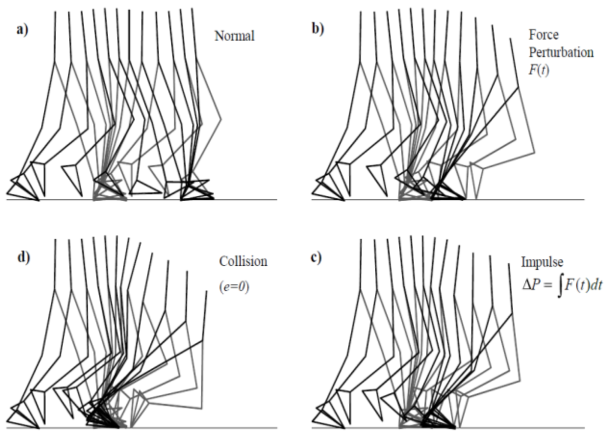

Fig. 3. Stick figure simulations with different contact formulations: a) normal, unperturbed gait; b) experimentally measured force perturbation applied to the ankle; c) horizontal impulse applied to the ankle; d) inelastic frictionless collision (e=0) at the ankle.

It appears to be more convenient, in these cases, to formulate contact as an instantaneous event with force perturbation characterized by an equivalent impulse or by an instantaneous collision characterized by a proper restitution coefficient. The appropriate choice of this coefficient is not trivial. Nevertheless, a purely inelastic collision provided acceptable results [10].

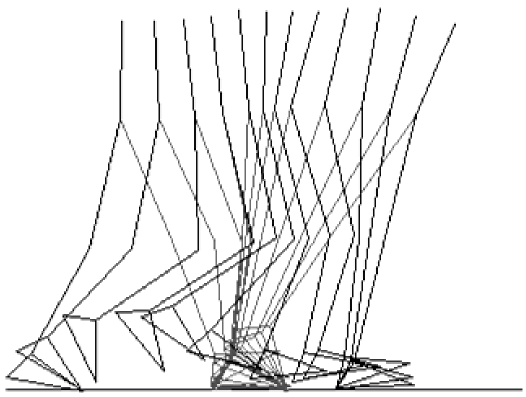

Fig. 4. Stick figure of the simulation with instantaneous inelastic collision and changes in the muscle excitation (25% of maximal) of the pure flexors of the perturbed leg and the pure extensors of the contralateral leg immediately after the trip.

During impact events significant soft tissue motion occurs weakening the rigid body assumption. Soft tissue behavior can be modeled using wobbling masses [17]. It was shown that the addition of wobbling masses to the model changes dramatically the internal forces only during the first tens of ms after the impact but has a limited influence on the kinematics of the model [17]. This observation justifies the use of a rigid multibody model for the study of recovery strategies after a stumble.

These experimental results are being used in an ongoing research in several ways: 1) the experimentally measured perturbation force was used to improve the simulation of the trip, as described in [10]; 2) in order to provide a basis for comparison of the simulation outcomes and validation of the model; 3) to guide the design of the control model of gait under development.

A final question that arises is if the addition of the coactivation on the muscle effectively results in an increase of joint impedance and how this increase could help in the rejection of the perturbation. It must also be considered that the coactivation has an increased energy cost and must be applied for a limited amount of time to avoid reaching fatigue. It is likely, that during the recovery the cost function is not based on energy but on stability requirements. These results pave the way for future analysis and modeling of the control mechanisms of gait.

Acknowledgment

The Authors thank Prof. B Koopman, AJ van den Bogert and FCT van der Helm for their inspiration in this work.

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