# Virtual Slope Control of a Forward Dynamic Bipedal Walker

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## Introduction

Active joint torques are the primary stabilizing factor in dynamic motion.  However the rate, timing, phasic behavior and amplitude of the joint torques necessary to achieve stable performance are difficult to establish.  MacMahon (1981) introduced Ballistic Gait, which is the idea that the swing phase of walking can be modeled as an unpowered multilink pendulum.  This method used a sinusoidal trajectory of the hip to represent the effect of the stance leg.  McGeer (1993) added the stance leg and conducted an analysis on dynamic stability of ballistic walking in continuous multi-stride simulations (Fig. 1).   Periodic energy


Fig.1. Compass Model

input is necessary to maintain stable locomotion.  In McGeer's study this input energy was achieved by walking downhill thereby limiting the applicability of the analyses.  Nonetheless it was demonstrated that natural step length and speed, for a given mechanical configuration (hip and leg masses and leg length), were a function of the slope, i.e. added energy rate.
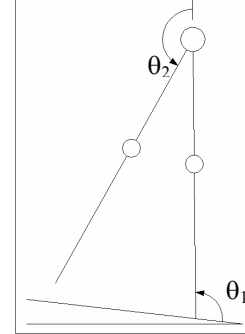
Previous studies use potential energy, i.e. downhill walking, to control passive motion thereby establishing dynamic stability.  Note that no active joint torques are required for downhill walking stability but gravitational joint torques are quantifiable.  Active joint torque is necessary to achieve uphill walking.  To guarantee stability the joint torques can be controlled in such a fashion to simulate the gravitational torques of downhill walking.  Hence, a simple proportional controller can be used to apply these active joint torques and maintain stable dynamics.  In this study a simple control system was developed to enable the forward-dynamic model to walk along any incline, e.g uphill.

## Statement of Clinical Significance

To assess pathologic constraints on human locomotion it is necessary to understand stability and control of human dynamics.  Compensation patterns in pathologic gait may be observed as adaptations necessary to achieve stability in the presence of physiologic constraints.

## Methodology

A forward dynamic model (Goswami 1988) was successfully developed and used to quantify stability and define the initial conditions associated with the region of stability.  Anthropomorphic data including segment lengths, mass, and mass distribution were used to develop more a two-segment bipedal walker with physical attributes of an average adult male.  To determine typical joint toques during downhill walking an unpowered passive model was implemented.  To walk uphill active joint torques based upon these kinetics were implemented.

$$\begin{bmatrix} T_{Ankle} \\ T_{Hip} \end{bmatrix} = g_y \begin{bmatrix} M_L(L-u_1)\tan(V_{Slope})\sin(\Theta_1) + M_H L \tan(V_{Slope})\sin(\Theta_1) + M_L L \tan(V_{Slope})\sin(\Theta_1) + M_L u_2 \tan(V_{Slope})\sin(\Theta_1+\Theta_2) \\ M_L u_2 \tan(V_{Slope})\sin(\Theta_1+\Theta_2) \end{bmatrix}$$

Since human locomotion occurs on a variety of surfaces, level, uphill, an active controller of joint torques is required for stable gait.  These joint torques can be added in any number of

ways, this system uses torque trajectories based on Virtual Slope, applied at the ankle and knee joints. The Virtual Slope controller uses the joint torques to simulate the forces a joint would experience on a pre-specified downhill slope. The result is a simple nonlinear proportional controller using an angle based torque feedback. To achieve desired stride length on any given slope one simply applies the appropriate virtual slope (Fig. 2).

## Results

Using this Virtual Slope approach for uphill walking the model is dynamically stable. Stability was validated by recording eigenvalues of the step function to be less than 1 (McGeer 1993). This demonstrates that the simple nonlinear controller is capable of producing stable gait for this system. Moreover, through the appropriate selection of virtual slope, the model can produce any reasonable desired stride length over the actual ground slope
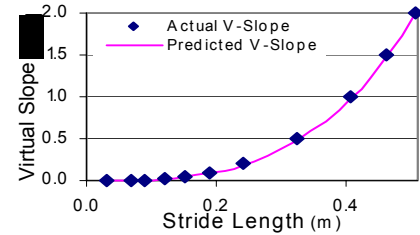
Fig. 2: Theoretical and numerical estimates of stride length as influenced by slope.

The virtual slope method of adding energy resulted in a smooth and gradual application of energy to the system throughout the gait cycle (Fig. 3-4). Previous models have assumed energy input was impulsive, i.e. short bursts of energy, despite human data that illustrates power consumption is smooth an continuous. Due to the nonlinear nature of the control algorithm the torque application curve cannot be normalized so the torque values are simply scaled for various Virtual Slopes, but the general application shape is similar.

## Discussion

Results from this model suggest that most of the torque necessary for walking is applied at the ankle joint, agreeing with human data indicating push-off is a primary energy source for walking. However, because there neither feet nor extensible leg the model predicted unrealistic dorsal flexion moments. Similarly, rigid-body dynamics necessitates instantaneous load transfer, i.e. no double stance phase. To permit improved realism and pathologic modeling future studies will expand the model to include knee and ankle joints.
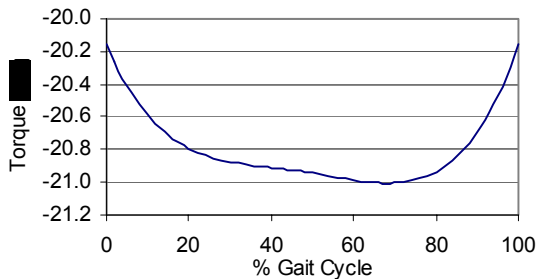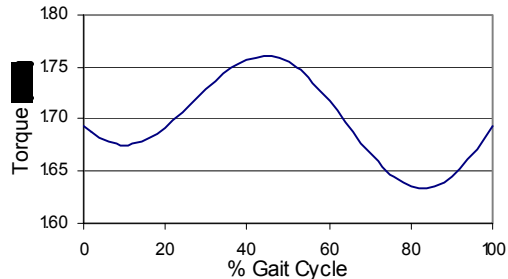
Figure 4a. Torque at stance ankle

Figure 4b. Torque at hip

## References

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