MODELLING THE NEUROMECHANICAL EVENTS OF LOCOMOTION AT VARYING GRAVITATIONAL LEVELS

M. Kathleen Day[1,2], Ryan J. Monti[1], Kevin Vallance[2], Shawn McGuan[3], Roland R. Roy[2], V. Reggie Edgerton[1,2]

[1]Department of Physiological Science and [2]Brain Research Institute
University of California, Los Angeles,
Los Angeles, CA 90095

[3]Mechanical Dynamics, Inc.
Ann Arbor, MI 48105

INTRODUCTION

A computer model that allows accurate simulations of bipedal and quadrapedal locomotion at varying gravitational levels and speeds can be an effective research tool in efforts to identify the critical control signals of a given locomotor system.

The most common definition of bipedal walking is a gait pattern in which at least one foot is always in contact with the ground and which has a period when both feet are simultaneously in contact with the ground. The latter condition is known as the double support phase. In running, the double support phase is replaced by a flight phase during which neither foot is in contact with the ground. In addition to foot-to-ground contact patterns, the mechanics of walking and running are different (5). During walking the body's center of mass passes over each leg in an inverted pendulum motion as the feet contact the ground. During running, the mechanics change to a spring-mass pattern with the legs acting as springs on which the center of mass bounces with each stride.

Animals exhibit a preferred walking speed above which they normally switch to a running gait. The change from a walking to a running gait occurs abruptly rather than as a gradual transition (8). Individual humans switch from walking to running at different absolute speeds, but at mechanically equivalent speeds (3). The transition occurs when a value known as the Froude number (see equation below) is approximately 0.5.

$$\text{Froude \#} = \frac{\text{Velocity}^2}{\text{Gravity x Leg Length}}$$

Based on this equation, the transition from a walking to a running gait should occur at slower speeds as gravitational levels decrease.

The purpose of the present study was to determine the feasibility of using a neuromechanical model of human locomotion based on a model previously published by Taga et al. (7) to simulate gait at various speeds and gravitational levels. The results indicate that this model may be appropriate for studying walking at 1G but not for higher speed or lower G locomotion.

METHODS

See Taga et al. (7) for a detailed description of the model including neural oscillator equations and initial conditions. Generally, the musculoskeletal model consisted of: a) five segments including a pelvis, 2 thighs and 2 shanks; b) six joints, two each at the hips, knees and ankles; and c) flexor and extensor "muscles" that generate active torques around the joints in proportion to the output of the neural system. Leg length was 1.1 m. Muscle forces were modeled as net torques about each joint. The model also included frictional torques at each joint and torques exerted by passive joint structures at the knees that limited the range of motion. The musculoskeletal system and the environment with which it interacted were created as a 3-D model in Adams® prototype simulation software (Mechanical Dynamics, Inc., Ann Arbor, MI). Since the Taga model is only capable of 2-D locomotion, the Adams® model was constrained in the saggital plane. The software allowed easy manipulation of environmental conditions such as ground stiffness, surface grade and gravitational level.

The neural control system was modeled in MatLab® (The MathWorks, Inc., Natick, MA) as a higher center and 12 oscillator units. There were two oscillators per joint, one controlling flexor torque and the other controlling extensor torque. Changing the magnitude of the higher center output constant resulted in changes in the speed of locomotion.

The Matlab® control system software was interfaced with the Adams® simulation software. Simulations were run with the higher center output constant set at values ranging from 2.5 to 8.5 and the gravitational level set at 1.0, 0.38 or 0.17G.

RESULTS

At 1G, the model achieved stable bipedal locomotion at speeds ranging from 1.0 m/s with the higher center constant set at 4.0 to 2.0 m/s with the higher center constant set at 7.5. At speeds below 2.0 m/s the model exhibited a walking gait with a double support phase and inverted pendulum mechanics (Figs. 1a and 2a). At 2.0 m/s there was an abrupt change to a gait resembling running. The running gait had a flight phase, however, it maintained the inverted pendulum mechanics seen in walking rather than transitioning to the expected spring-mass mechanics (Figs. 1b and 2b). The Froude number equaled 0.37 at the transition point.

At gravitational levels simulating those found on the moon, 0.17G and Mars, 0.38G, the model was not able to achieve stable locomotion at a speed higher than 0.8 m/s. The gait pattern had a double support phase and inverted pendulum mechanics (Fig. 2c) as the Froude number would have predicted.

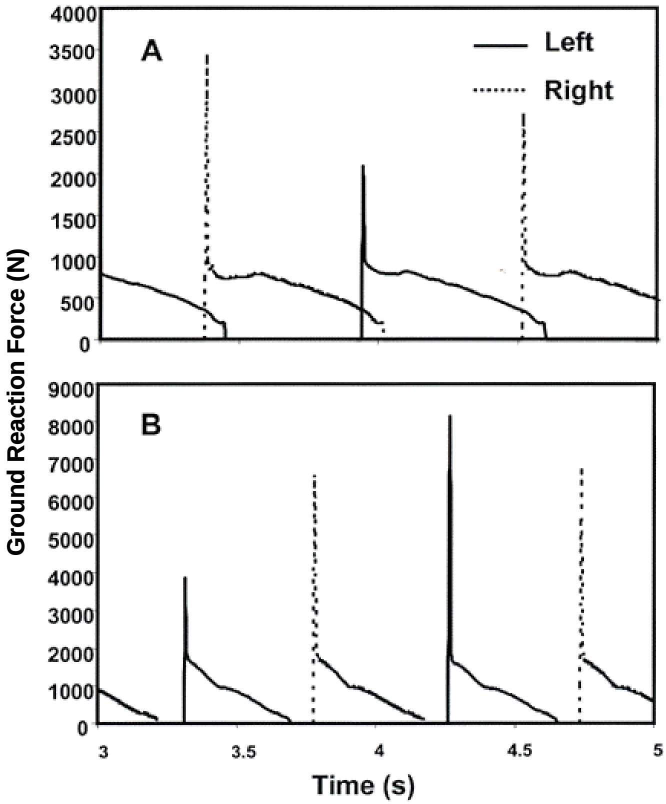

Figure 1. *Ground reaction forces (y component) at locomotion speeds of: a) 1.2 m/s and b) 2.0 m/s. At the slower speed one foot is always in contact with the ground and there is a double support phase indicating that the model is using a walking gait. At the higher speed there is a flight phase indicating a running gait.*

DISCUSSION

It has been demonstrated that individuals have a preferred walking speed and that the transition from walking to running is abrupt (8). These observations suggest that the neuromotor control strategy consists of a stable limit cycle for walking and that the transition to running represents a bifurcation to another stable limit cycle. The neural oscillator model described by Taga et al. (7) exhibits these behaviors and is based on flexible control, rather than on an engineering control theory of planning and execution. Also, some physiological evidence suggests that the speed of locomotion is normally modulated by higher centers as in Taga's neural control system (6).

However, while the neural control system in Taga's model mimics some of the characteristics of physiological behavior associated with locomotion, the musculoskeletal system demonstrates accurate gait mechanics only under limited conditions. In our simulations, the model was incapable of producing the spring-mass mechanics seen in bouncing gaits such as running at 1G and did not achieve stable locomotion in low G at a speed higher than 0.8 m/s.

A feature of the Taga model that may account for this deficit is that the torques generated at the joints

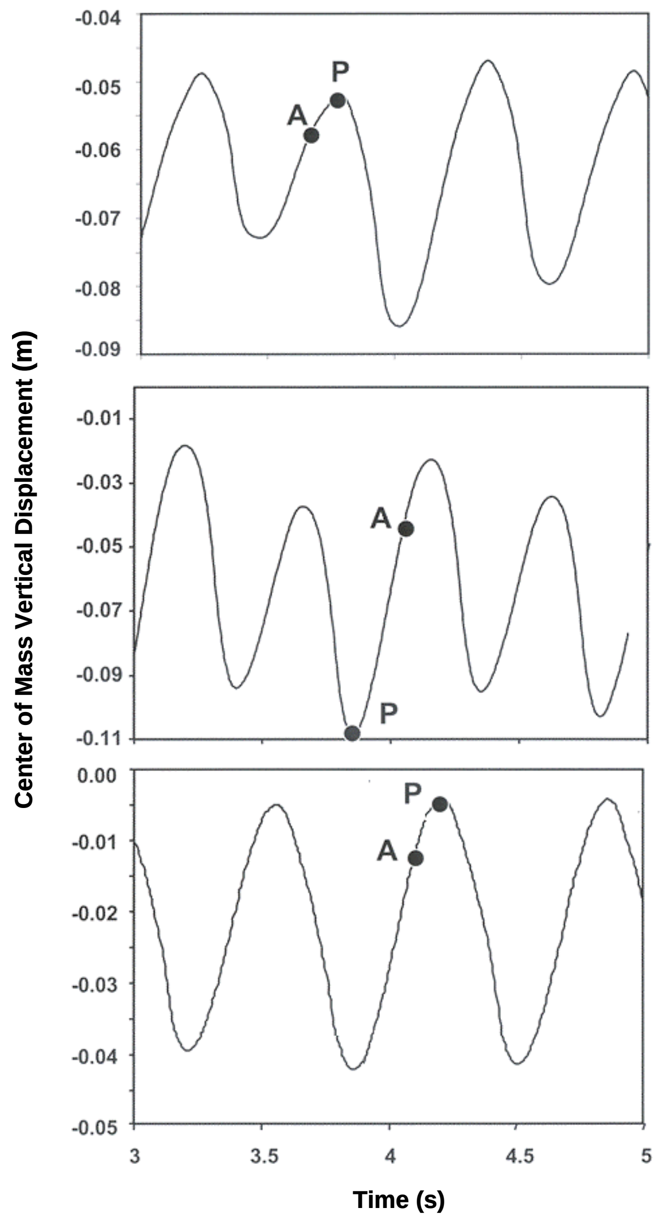

Figure 2. *Center of mass movement during locomotion at: a) 1G, 1.2 m/s, b) 1G, 2.0 m/s, c) 0.38G, 0.8 m/s. In an inverted pendulum gait the center of mass should be at its highest point at mid-stance. In a spring-mass gait the center of mass should be at its lowest point at mid-stance. The predicted (P) mid-stance point and actual (A) mid-stance point observed for each condition is indicated.*

are proportional to the output of the neural oscillators. This means that a muscle's torque production is assumed to be the same throughout the range of joint motion and the stress-strain properties of the muscle-tendon units are ignored.

As the joint angle changes, the mechanical advantage of a muscle changes (4). Therefore, the same neural output to a muscle can produce dramatically different torques at different joint angles. In human running this is an important factor to consider because increases in speed are associated with increases in leg sweep (2) which results in greater changes in joint angle.

An important feature of spring-mass gaits is that mechanical energy is stored as strain energy in the muscle-tendon units. This strain energy is recovered during each step at minimal metabolic cost to the system resulting in more efficient locomotion (1). Therefore, the Taga model may seriously underestimate net joint torques during running because it does not have the capability of conserving and returning muscle-tendon strain energy.

This study suggests the importance of coupling a neural control system model with a musculoskeletal system model that is based on accurate morphology and physiology when studying human movement dynamics. It appears that this coupling becomes even more important when factors related to gravitational load and speed of locomotion are to be considered.

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