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Phase resetting behavior in human gait is influenced by treadmill walking speed

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ABSTRACT

Gait is often modeled as a limit cycle oscillator. When perturbed, this type of system will reset its output in a stereotypical manner, which may be shifted in time with respect to its original trajectory. In contrast to other biological oscillators, relatively little is known regarding the phase resetting properties for human gait. Because humans must often reset their gait in response to perturbation, an improved understanding of this behavior may have implications for reducing the risk of fall. The purpose of this study was to further evaluate phase resetting behaviors in human gait with particular emphasis on (1) variance of the phase resetting response among healthy individuals and (2) the sensitivity of this response to walking speed. Seventeen healthy subjects walked on a treadmill at 2.0 mph, 2.5 mph, and 3.0 mph while their right limb was perturbed randomly every 12–20 strides. Discrete, mechanical perturbations were applied by a rope that was attached to each subject's ankle and actuated by a motorized arm. Perturbations were applied once during a select stride, always at a different point in the swing phase, and the amount of phase shift that occurred on the subsequent stride was recorded. A subset of 8 subjects also walked at their preferred walking speed for 3 additional trials on a separate day in order to provide an estimate of within-subjects variability. The results suggested that phase resetting behavior is relatively consistent among subjects, but that minor variations in phase resetting behavior are attributable to walking at different treadmill speeds.

1. Introduction

Unintentional falls are a significant public health problem among aging and diseased populations. Unexpected perturbations are an important cause of falls that occur during walking [1,2]. Effective solutions to fall prevention may therefore be related to an improved understanding of the human locomotor response to perturbation.

A promising approach to understanding gait stability involves modeling gait as a limit cycle oscillator [3,4]. During human gait, oscillator dynamics arise from a variety of mechanisms, including the biomechanical properties of limbs (i.e. passive dynamic walking [3,4]) and the output of the central pattern generator (CPG) [5]. This approach is convenient because a limit cycle oscillator responds to perturbations in a stereotypical manner and this behavior can help to explain how unanticipated events might contribute to falling. For example, when a healthy person stumbles, they quickly reset their limit cycle in a predictable manner and continue walking [6–9]. While there is some debate as to whether gait can be considered a true limit cycle oscillator, several previous investigations have demonstrated the utility of this approach for understanding locomotor control (e.g. [3,4,6,9,10]).

The response of a limit cycle oscillator to perturbation can be characterized by examining its phase resetting properties. These properties have been studied extensively in several biological systems, including neurons [11–13], circadian rhythms [14], and cardiac cells [15]. This behavior is succinctly described in terms of a phase resetting curve (PRC), which represents the transient change in the cycle period of an oscillator induced by a perturbation that is a function of the phase in which it is received [16,17]. This curve is often established by perturbing a system at different phases and

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measuring the amount of phase shift that occurs when the cycle is reset. Once established, the PRC for a biological oscillator can be a valuable and rich source of information.

There are good reasons to suspect that the measurement of the PRC for human gait holds implications for understanding the risk of falling. For example, Nomura et al. utilized a walking simulation to demonstrate that the probability of falling in response to a small perturbation is related to the phase reset [6,7,9,18]. These simulations were designed using a PRC that had been recorded previously in walking humans by introducing a brief stimulus that resisted motion of the swing leg [9]. The results suggested that a phase resetting response that deviated from the observed PRC in healthy humans is likely to place an individual at greater risk for fall. Other observations suggest that phase resetting in response to perturbation is abnormal in older adults. These include video motion capture of falls [1] and responses to lateral perturbations (i.e. slip) of the walking surface [19–21]. Moreover, additional data suggest that a phase-dependent modulation of the limit cycle is typical of human walking, and effective use of positive (lengthened) or negative (shortened) resetting dynamics can reduce a person's risk for fall [8,22–25].

In comparison to other biological phenomena, the PRC for human gait is currently not well defined. Several studies have demonstrated that, in general, perturbations applied early during the swing phase of gait result in a flexor response, which prolongs swing and results in a delay of the onset of the next step. In contrast, perturbations that occur later during swing result in an extensor response, which shortens swing and results in an advance of the next step [8,22,23]. However, this behavior has only recently been studied under the paradigm of the PRC [13]. In addition, this behavior may not be consistent for individuals with impairment or those walking under less than ideal conditions.

Several questions remain regarding the nature of phase resetting behavior in human walking. For example, one previous study in neurons reported that different cells exhibit different PRCs, and the shape of these PRCs was sensitive to changes in the baseline firing rate of the neuron [13]. These results imply that the PRC for human gait may also vary among healthy individuals and these curves may be affected by an individual's walking speed (analogous to baseline firing frequency in a neuron). Therefore, the purpose of this experiment was to examine the phase resetting curve among young healthy walkers to determine if (1) differences exist between individuals and (2) phase resetting responses are altered by changes in treadmill walking speed. In addition, an analysis of the number of cycles (i.e. strides) required for an individual to return to their normal cycle length following perturbation is provided.

2. Methods

2.1. Subjects

A convenience sample of 17 healthy male and female subjects was recruited from the local student population (age 25.6 ± 4.9 years, mass 66.0 ± 14.8 kg, height 1.7 ± 0.1 m). All subjects were free of musculoskeletal or neurological conditions that may have affected gait. Approval for these procedures was obtained through the Institutional Review Board at California State University, San Marcos, and all subjects gave their informed consent before participation.

2.2. Apparatus

Subjects walked on a treadmill with both side rails and control panel removed. Both ends of a lightweight rope were attached to their right ankle (Fig. 1). This rope was routed through a loop via low friction pulleys, beginning from behind the subject's ankle,

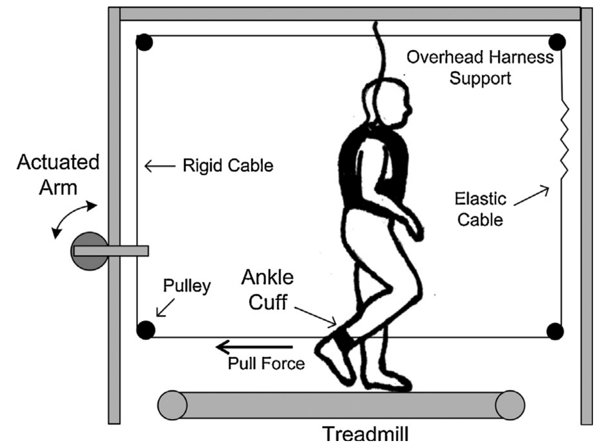

Fig. 1. Experimental setup.

through a mechanical arm mounted behind the treadmill, overhead, and then to the front of the subject where it attached again on the anterior aspect of the same ankle. A small section of this loop in front of the subject was elastic to allow the ankle to move in three dimensions, as occurs normally during swing. This loop was under a small amount of tension to ensure that it remained taut. With each step, the rope was pulled forward and backward along its loop by the action of the stepping subject. This configuration provided very little resistance to normal movement of the ankle in the sagittal plane.

Perturbations were applied directly to a section of the rope that was highly inelastic. A ¼ Hp DC electric motor (Bison Gear & Engineering Corp.) was used to actuate a mechanical arm, which performed discrete pulls on the rope from behind the subject with a peak force up to 180 N and an average rise time of 0.125 s. An overhead harness system was used to ensure safety but did not provide any support of body weight. Specialized harnesses were used to avoid any interference with lower extremity motion (Maine Anti-Gravity Systems, Inc.).

Control of the apparatus was provided by a custom program written in Matlab's xPC Target (Natick, MA). Each walking trial was completely automated using feedback received from specialized insoles worn by subjects in their right shoe (B&L Engineering). These insoles were instrumented with foot switches at the heel, base of the 1st and 5th metatarsals, and the great toe, and these sensors were used to determine the onset and end of swing. Once a trial was initiated, the system was programmed to apply a single, discrete perturbation in a pseudorandom fashion every 12–20 strides. The within-stride timing of perturbations followed a simple algorithm in which each pull advanced in time by 5% of the subject's average swing duration (determined during warm-up). The trial ended after 19 pulls, ensuring that perturbations were applied throughout the entire swing phase. This yielded an average of 300 strides per trial, and each trial was approximately 5–6 min in duration. All pulls occurred during the swing phase of the perturbed limb. No perturbations were applied during the stance phase as it was previously demonstrated that phase shift is minimal during stance for this type of perturbation [7].

2.3. Data collection and analysis

Prior to data collection, subjects were asked to walk on the treadmill for a period of 2 min in order to become familiar with the harness, treadmill, insole, and ankle cuff. Each subjects' preferred walking speed (average: 2.56 ± 0.23 mph) was determined during this 2 min warm-up. Subjects then performed 3 different trials at 3 different treadmill speeds: 2.0 mph, 2.5 mph, and 3.0 mph. A subset