APPLICATION OF HARMONIC TRANSFER FUNCTIONS METHOD OF ANALYSIS TO UNDERSTAND RESPONSES TO MECHANICAL PERTURBATIONS IN HUMAN WALKING

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INTRODUCTION

Human walking has been studied extensively under approximately steady-state conditions, for example, when subjects walk on a treadmill whose belt moves at a fixed speed. In everyday life, acceleration and deceleration, obstacle negotiation, disturbance rejection, and turning are all examples of behaviors that require modulation of walking speed, which is driven by transient changes in muscle activations [1], [2], [3]. However, even during steady-state walking, the nervous system is continually modulating muscle activations based on sensory information to ensure stability, as evidenced by small continual changes in walking speed and other kinematic variables.

Weak continuous sensory and mechanical perturbations can be used to identify properties of this neural feedback control of steady-state walking [4], [5], based on a local limit-cycle approximation of human gait. Responses to perturbations are characterized in time domain using phase-dependent impulse response functions (φIRFs). A φIRF describes the response to a small brief perturbation (an impulse) applied at any phase of the gait cycle and, by extension, the response to any small perturbation. In this study, we applied this approach to characterize the effects of continually varying treadmill speed on kinematic variables.

METHODS

The experiment design has been previously described in detail [6]. Twelve subjects walked on an instrumented treadmill (Forcelink, Culemborg, Netherlands) at three different speeds of 0.8, 1.2 and 1.6 ms^{-1}. Forty seven reflective markers were attached to different anatomical landmarks and the kinematics were captured using a ten camera motion capture system (Motion Analysis, Santa Rosa, CA, USA). Each trial was ten minutes starting with one minute of normal walking, followed by eight minutes of perturbation, and ending with one minute of normal walking. The perturbation was designed by generating filtered white noise as an acceleration signal in MATLAB (The MathWorks, Inc., Natick, MA, USA) which was then integrated to derive a velocity signal. This signal was commanded to the treadmill using D-Flow software (Motek Medical, Amsterdam, Netherlands).

To approximate walking as a linear time periodic system [7], we replace the time with estimated phase of the gait cycle as the independent variable. Heel-strike times were used to derive a piecewise-linear causal estimate of phase, which was then filtered to obtain a continuous approximation. We then characterized kinematic responses to the perturbation signal (i.e., speed of the belt) in the frequency domain using harmonic transfer functions [4]. An HTF has a “mode” for each integer k with its own gain and phase functions. The k-th mode describes the mapping from frequency f to frequency $f + kf_0$ where f_0 is the gait frequency. To make it easier to interpret kinematic responses, we converted HTFs to the φIRFs described above. A correction was made for the method used to estimate phase, so that to first order in perturbation size, the resulting φIRFs did not depend on the estimation method [5].

RESULTS AND DISCUSSION

Figure 1 shows the φIRF between the belt speed and the velocity of right ankle. As we can see in the figure, there is an increase in the velocity immediately after the onset of perturbation (the red

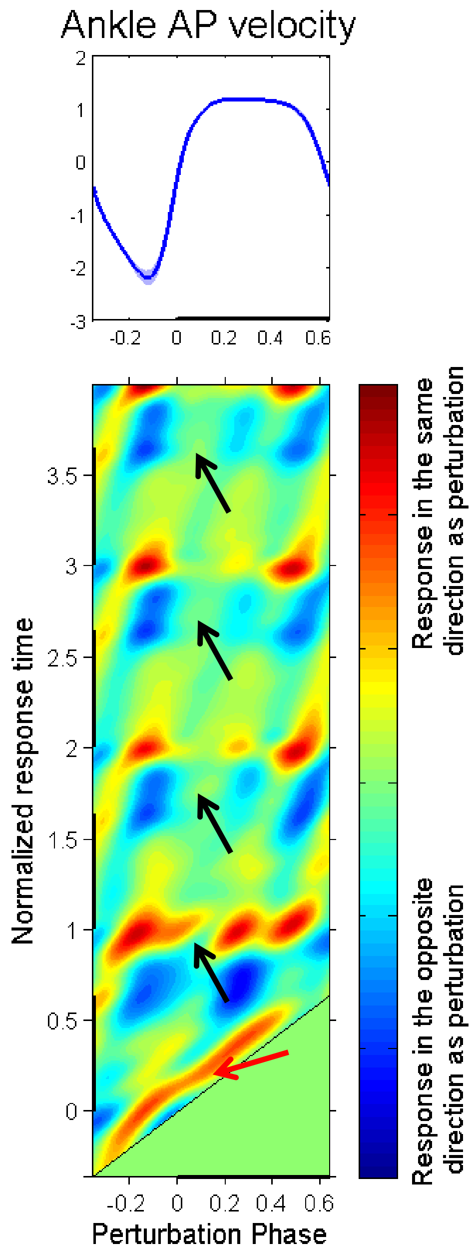

Figure 1: *Top:* Mean waveform of the right ankle anteroposterior (AP) velocity across all cycles during the trial. The shaded area shows 95% confidence intervals and the black bar on the horizontal axis marks the stance phase of the cycle. *Bottom:* the φIRF between the belt speed and the ankle velocity for four cycles after perturbation onset calculated from modes k with $|k| \leq 3$. Black bars on each axis mark the stance phase. The color red in the spectrum represents a change in the response in the same direction as the change in perturbation and blue represents changes in the opposite direction. The diagonal line is the onset of perturbation.

area along diagonal marked by the red arrow). Since the input is the speed of the treadmill, we expect to see an impulse in the foot velocity as a response to an impulse in the belt speed. This transient change in velocity is only observed in the first cycle after perturbation and fades away in the following cycles. Another important feature of the observed response is the phase resetting [8], an initial increase in the speed, followed by an immediate decrease, repeated throughout the subsequent cycles (indicated by black arrows in the figure).

The use of φIRFs provides a complete picture of kinematic responses to small perturbations. However, as the size of perturbations increases, φIRFs can be affected by second-order effects, which we are currently investigating by varying perturbation size. We are also measuring electromyographic (EMG) responses, to determine the role of neural feedback in responding to changes in treadmill speed. Longer term, future studies can apply this method to investigate transient responses in geriatric or neurological populations, using kinematic and EMG measures to better understand the neural control mechanisms used to ensure stable walking.

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