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An active balance board system with real-time control of stiffness and time-delay to assess mechanisms of postural stability

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ABSTRACT

Increased time-delay in the neuromuscular system caused by neurological disorders, concussions, or advancing age is an important factor contributing to balance loss (Chagdes et al., 2013, 2016a,b). We present the design and fabrication of an active balance board system that allows for a systematic study of stiffness and time-delay induced instabilities in standing posture. Although current commercial balance boards allow for variable stiffness, they do not allow for manipulation of time-delay. Having two controllable parameters can more accurately determine the cause of balance deficiencies, and allows us to induce instabilities even in healthy populations. An inverted pendulum model of human posture on such an active balance board predicts that reduced board rotational stiffness destabilizes upright posture through board tipping, and limit cycle oscillations about the upright position emerge as feedback time-delay is increased. We validate these two mechanisms of instability on the designed balance board, showing that rotational stiffness and board time-delay induced the predicted postural instabilities in healthy, young adults. Although current commercial balance boards utilize control of rotational stiffness, real-time control of both stiffness and time-delay on an active balance board is a novel and innovative manipulation to reveal balance deficiencies and potentially improve individualized balance training by targeting multiple dimensions contributing to standing balance.

1. Introduction

Rotational/translational balance boards are frequently used to assess and improve standing upright balance (Balogun et al., 1992; Distefano et al., 2009). They can be used to study the sensory contributions (e.g., visual, vestibular, somatosensory) to the maintenance of stance (Cohen et al., 1996; Mergner et al., 2003; Maurer et al., 2006) and can differentiate healthy and balance-compromised cohorts (Liston and Brouwer, 1996; Oliveira et al., 2008; Ray et al., 2008). In addition, balance training on a rotational balance board prevented injury in young athletes (Paterno et al., 2004; Aaltonen et al., 2007; Emery and Meeuwisse, 2010) and improved stability in balance-compromised populations (de Bruin et al., 2009; Godard et al., 2004; Hinman, 2002; Nordt et al., 1999; Chen et al., 2002; Hue et al., 2004; Zech et al., 2010).

Commercial systems such as the Biodex System SD® and the NeuroCom SMART Balance Master® manipulate board stiffness and force/torque to alter the difficulty of standing. Both provide metrics which correlate well with measures of balance performance such as the Balance Error Scoring System (Riemann et al., 1999) and the Berg Balance score (Liston and Brouwer, 1996). The adjustable rotational stiffness of the platforms allows individualized balance training.

However, to our knowledge, no commercial balance board is capable of manipulating feedback time-delay. Feedback time-delay is an important parameter given the inherent time-delays of neuromuscular system due to afferent and efferent nerve conduction, processes of the central nervous system, and electromechanical delays of skeletal muscles. Neuromuscular time-delay often increases with disease or injury (Cameron et al., 2008; Bloem, 1992; Eckner et al., 2012). Longer time-delays can lead to sustained, large amplitude, harmonic postural fluctuations

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called limit cycle oscillations (LCOs). LCOs have been observed in patient populations including athletes with concussion, persons with Parkinson's disease, and persons with multiple sclerosis (Chagdes et al., 2016a,b). Neuromuscular deficits can also result in decreased sensitivity to postural fluctuations, or the inability to produce contextually appropriate corrective muscle torque (Bisdorff et al., 1996; Gutierrez et al., 2001; Carpenter et al., 2004). Thus, a comprehensive balance board system for populations with neuromuscular deficits should allow for the manipulation of sensory inputs, stiffness/torque, as well as time-delay which can be used to identify the specific mechanism contributing to postural instability.

In this paper, we describe the design and construction of a 1-DOF active balance board and a mathematical model of bipedal stance the board to explore the relationship between upright stability and the controllable board parameters: rotational stiffness and time-delay. Subsequently, we present an experimental study testing the active balance board in a population of healthy, college-aged adults. Our results provide evidence of two forms of postural instabilities, (a) as stiffness decreases, and becomes insufficient to maintain upright stance, the board can rapidly "tip" through a pitchfork bifurcation, and (b) LCOs about the upright position arise when active board time-delay is increased. The ability to induce unique forms of instability by controlling two independent parameters may allow for better characterization of balance deficiencies, and the ability to create personalized balance training regimens.

2. Construction of 1-DOF active balance board

A prototype of a balance board with controllable stiffness and feedback time-delay to assess human posture was constructed (Fig. 1). The maximum angular deviation of the board was ±20° about the mediolateral (ML) axis. The board geometry was designed to mimic an inverted pendulum, with dynamics similar to the Biodex Balance System SD®. This design ensured that as rotational stiffness was decreased, the upright position would eventually become unstable.

The desired rotational torque, M_{board}, was generated by adjusting the pressure in the actuating component, two pneumatic cylinders, and taking into account the board angle, which was measured with linear position sensors in line with the cylinders. The rotational torque was controlled using LabVIEW and the cRIO system from National Instruments. The hardware had a small response time as the cylinders filled with air, but observations indicated that this delay did not affect the predicted board operation. To implement a board feedback time-delay, the desired M_{board} was calculated, but the rotational stiffness gain was multiplied by the angle which occurred τ_{board} prior. More details about the design and fabrication of the balance board can be found in Supplementary Information.

3. Mathematical model and simulations

Our model of a person-board coupled dynamical system was adapted from the human-balance board model developed in Chagdes et al. (2013), a 1-DOF inverted pendulum human model coupled with a 1-DOF inverted pendulum balance board. Adaptations included the addition of controllable rotational stiffness and time-delay. The sway angle, θ, of the person is measured relative to the balance board in the sagittal plane; the rotation angle of the balance board is ϕ, also in the sagittal plane. The key inputs are the applied ankle torque, M_{ankle}, and desired rotational torque, M_{board} (Fig. 2). The system dynamics are described by the coupled equations

$$\begin{bmatrix} I_{11}(\theta) & I_{12}(\theta) \\ I_{21}(\theta) & I_{22}(\theta) \end{bmatrix} \begin{Bmatrix} \ddot{\theta} \\ \ddot{\phi} \end{Bmatrix} + \begin{Bmatrix} C_{11}(\theta,\dot{\theta},\dot{\phi}) \\ C_{21}(\theta,\dot{\theta},\dot{\phi}) \end{Bmatrix} + \begin{Bmatrix} K_{11}(\theta,\phi) \\ K_{21}(\theta,\phi) \end{Bmatrix} = -\begin{Bmatrix} M_{ankle} \\ M_{board} \end{Bmatrix}. \tag{1}$$

where

$$I_{11} = I_{body} + m_{body} h_{body}^2$$

$$I_{12} = I_{21} = I_{body} + m_{body}(h_{body}^2 + h_{ankle} h_{body} \cos\theta - h_{body} x_{ankle} \sin\theta)$$

$$I_{22} = I_{body} + m_{body}(h_{ankle}^2 + x_{ankle}^2 + h_{body}^2 + 2h_{ankle} h_{body} \cos\theta - 2h_{body} x_{ankle} \sin\theta) + m_{board} h_{board}^2 + I_{board}$$

$$C_{11} = m_{body}(h_{ankle} h_{body} \sin\theta + h_{body} x_{ankle} \cos\theta)\dot{\phi}^2$$

$$C_{21} = -2m_{body}(h_{ankle} h_{body} \sin\theta + h_{body} x_{ankle} \cos\theta)\dot{\theta}\dot{\phi} - m_{body}(h_{ankle} h_{body} \sin\theta + h_{body} x_{ankle} \cos\theta)\dot{\theta}^2$$

$$K_{11} = -m_{body} g h_{body} \sin(\theta + \phi)$$

$$K_{21} = -m_{body} g [h_{ankle} \sin\phi - x_{ankle} \cos\phi + h_{body} \sin(\theta + \phi)] - m_{board} g h_{board} \sin\phi$$

In Eq. (1), the I matrix represents the inertial terms, the C matrix represents the angular velocity-dependent acceleration terms, and the K matrix represents the destabilizing gravity terms. Excluding the feet, the human body has mass, m_{body}, with center of mass (CoM) assumed to be at a constant distance of h_{body} from the ankle joint. The ankle joint is assumed to be positioned above the balance board's axis-of-rotation such that $x_{ankle} = 0$. The balance board and foot have a lumped mass m_{board} that is centered at a constant distance of h_{board} from the balance board axis-of-rotation; g is acceleration due to gravity. The balance board has external actuators that apply a torque to the board as follows:

$$M_{board}(t) = K_{board}\phi(t) + K_{3,board}\phi^3(t) + C_{board}\dot{\phi}(t) + K_{\tau,board}\phi(t - \tau_{board}), \tag{2}$$

where K_{board} represents the passive, linear rotational stiffness gain, C_{board} represents the passive, linear rotational damping gain, $K_{3,board}$ represents the passive, cubic rotational stiffness gain, $K_{\tau,board}$ represents the delayed, linear rotational stiffness gain, and τ_{board} represents the board feedback time-delay.

The corrective ankle torque applied by the body to maintain upright position is the sum of passive (i.e., muscle stiffness and damping) and active elements, from proprioception, vestibular, and vision. We have followed the framework of the ankle torque models presented in the literature (Asai et al., 2009; Maurer and Peterka, 2005; Peterka, 2002; Vette et al., 2010), but also added a nonlinearity in the muscle stiffness term to limit the amplitude of the LCOs,

$$M_{ankle} = \underbrace{K[\theta(t) + \beta\theta^3(t)] + C\dot{\theta}(t)}_{M_{ankle,passive}} + \underbrace{K_p\theta(t-\tau) + K_v[\theta(t-\tau) + \phi(t-\tau)]}_{M_{ankle,active}}. \tag{3}$$

The passive elements are modeled as a nonlinear spring, with linear stiffness gain K, cubic stiffness gain $K\beta$, and a linear dashpot with damping gain C which acts with zero time-delay. The active elements are composed of two linear springs with delayed responses representing the delay in neuromuscular feedback, τ. The first spring has a gain of K_p and models the contributions of the proprioceptive system, which detects changes in body position relative to the board, correcting only when $\theta \neq 0$. The second spring has a gain of K_v which models the combined contributions of the vestibular and visual systems to detect changes in the body position relative to the gravitational line, correcting when $(\theta + \phi) \neq 0$. The block diagram of the postural control system can be seen in Fig. 2(b).