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Evaluating feedback time delay during perturbed and unperturbed balance in handstand

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ABSTRACT

Feedback delays in balance are often assessed using muscle activity onset latencies in response to discrete perturbations. The purpose of the study was to calculate EMG latencies in perturbed handstand, and determine if delays are different to unperturbed handstand. Twelve national level gymnasts completed 12 perturbed and 10 unperturbed (five eyes open and five closed) handstands. Forearm EMG latencies during perturbed handstands were assessed against delay estimates calculated via: cross correlations of wrist torque and COM displacement, a proportional and derivative model of wrist torque and COM displacement and velocity (PD model), and a PD model incorporating a passive stiffness component (PS-PD model). Delays from the PD model (161 ± 14 ms) and PS-PD model (188 ± 14 ms) were in agreement with EMG latencies (165 ± 14 ms). Cross correlations of COM displacement and wrist torque provided unrealistically low estimates (5 ± 9 ms). Delays were significantly lower during perturbed (188 ± 14 ms) compared to unperturbed handstand (eyes open: 207 ± 12 ms; eyes closed: 220 ± 19 ms). Significant differences in delays and model parameters between perturbed and unperturbed handstand support the view that balance measures in perturbed testing should not be generalised to unperturbed balance.

1. Introduction

The control of posture involves a feedback system which processes visual, vestibular, and somatosensory inputs and executes neuromuscular actions to maintain equilibrium (Horak, 2006; Wade & Jones, 1997; Winter, 1995). Feedback time delays complicate control tasks, placing constraints on the strategies employed and limiting the stability of the system (Bottaro, Casadio, Morasso, & Sanguineti, 2005; Peterka, 2000; Suzuki, Nomura, Casadio, & Morasso, 2012; Vette, Masani, Nakazawa, & Popovic, 2010). Likewise, the mechanical stability of the system places constraints on the strategies employed and limits the range of delays possible for stable control to be established (Suzuki et al., 2012). Control models of human balance must include the appropriate level of mechanical instability and a realistic time delay to accurately simulate postural dynamics. Models that incorporate large amounts of noise to replicate human sway typically use delays of 80–100 ms (van der Kooij, van Asseldonk, & van der Helm, 2005; Vette et al., 2010), whereas models using intermittent control to replicate human sway typically use delays of 180–200 ms (Bottaro, Yasutake, Nomura, Casadio, & Morasso, 2008; Suzuki et al., 2012). The different conclusions drawn from these models may be due to the different feedback time delays used.

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There are several possible sources of instability in human posture. The majority of previous literature has focused on the examination of postural control during upright stance, however, this is not the only form of human balance (Slobounov & Newell, 1996). Balancing in handstand is mechanically less stable than balancing in normal upright stance, with a reduced base of support, a higher centre of mass, and a reduced maximal strength in the controlling joints. Regardless, from a control perspective, the two tasks are equivalent, requiring estimation of the spatial arrangement and motion of the body, and implementing a multi-joint strategy to preserve orientation and configuration within certain constraints (Yeadon & Trewartha, 2003). Handstand balance performed by experienced gymnasts provides an alternative perspective to normal upright stance for understanding this complex system.

During quiet stance an ankle strategy is dominant, where the motion of the centre of mass (COM) is controlled by torque about the ankle joint with synergistic torques about superior joints to maintain a fixed body configuration (Nashner & McCollum, 1985; Runge, Shupert, Horak, & Zajac, 1999). The ankle strategy is analogous to a wrist strategy whilst balancing in handstand, where the system can be modelled as a single segment inverted pendulum controlled by wrist joint torque (Yeadon & Trewartha, 2003). The feedback time delay from an initial perturbing motion to muscle force production comprises sensory delay, neurological delay, and electromechanical delay (Fig. 1). Sensory delay arises as it takes time for receptors to reach a sensory threshold; it is expected to be related to speed of motion and sensory acuity. Neurological delay represents the time for sensory transduction, neural processing, and motor signal transmission (Peterka, 2002), and is expected to be relatively constant within an individual. Electromechanical delay is the time from muscle activation until the onset of muscle force production, which is approximately 13–55 ms (Cavanagh & Komi, 1979; Tillin, Jimenez-Reyes, Pain, & Folland, 2010; Zhou, Lawson, Morrison, & Fairweather, 1995). This wide variation in electromechanical delay can be due to differences in musculotendon tension and slack length (Muraoka, Muramatsu, Fukunaga, & Kanehisa, 2004).

The time from a discrete platform perturbation until the onset of muscle activation, detected via electromyography (EMG), will describe the time from the initiation of movement until muscle activation ($t_2 - t_0$). In perturbed standing EMG latencies are approximately 65–130 ms, with corresponding changes in joint torques occurring around 30 ms later (Horak, Diener, & Nashner, 1989; Nashner, Woollacott, & Tuma, 1979), representing the electromechanical delay ($t_3 - t_2$). However, these values may not be typical of unperturbed balance as static and dynamic posturography techniques address different aspects of the postural control system (Baratto, Morasso, Re, & Spada, 2002).

Feedback time delay estimates in quiet stance have been determined from cross correlations of centre of mass (COM) and centre of pressure (COP) trajectories, which describe the time from peak COP displacement to peak COM displacement ($t_6 - t_5$) or peak COM velocity ($t_6 - t_4$). The delay between COP and COM displacements is normally zero and has been suggested as evidence of a passive control system (Winter, Patla, Ischac, & Gage, 2003; Winter, Patla, Prince, Ishac, & Gielo-Perczak, 1998). However, it has also been suggested this is evidence of an active anticipatory feedforward control process (Gatev, Thomas, Kepple, & Hallett, 1999). These studies assume that the postural control strategy relies primarily on segment or whole body displacement information from sensory inputs, however, the postural control system adopts a strategy that relies heavily on velocity information during quiet stance (Kiemel, Oie, & Jeka, 2002; Masani, Popovic, Nakazawa, Kouzaki, & Nozaki, 2003). It has also been shown that sensory thresholds for displacement at the ankle (Clark, Burgess, Chapin, & Lipscomb, 1985; Fitzpatrick & McCloskey, 1994) and metacarpophalangeal joint (Clark et al., 1985) are reduced when the

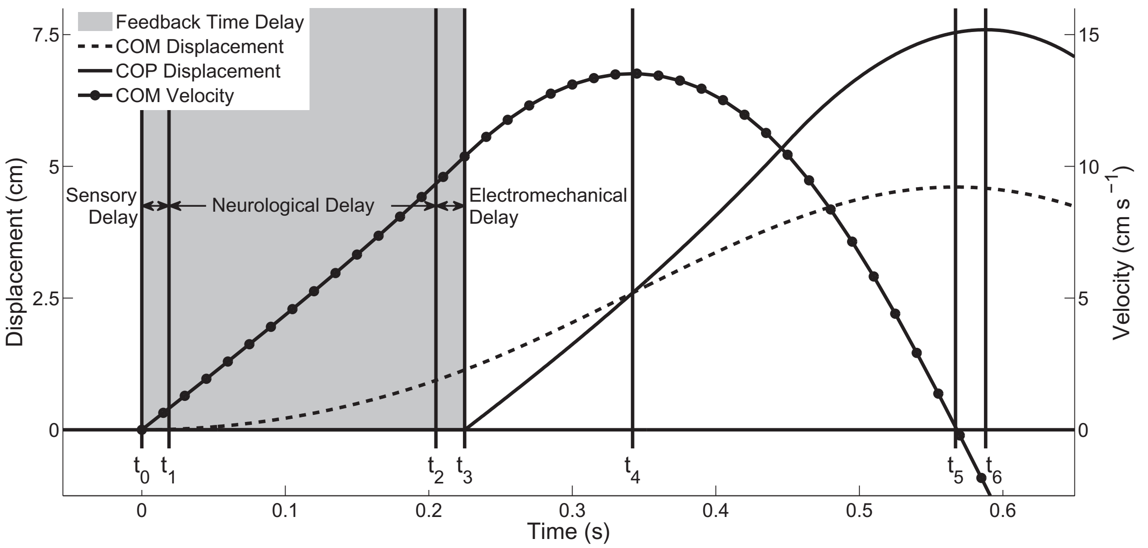

Fig. 1. A theoretical description of COP and COM motion of a single inverted pendulum model of balance controlled by a reactive strategy [t_0 = *initial motion;* t_1 = *sensory threshold is reached and motion is detected;* t_2 = *muscle activated;* t_3 = *muscle force is produced;* t_4 = *COP crosses COM, joint torque is higher than torque due to gravity, and peak COM velocity is reached;* t_5 = *peak COM displacement;* t_6 = *peak COP displacement*].