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A unified approach for revealing multiple balance recovery strategies

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ABSTRACT

In human balance recovery, different strategies have been proposed with generally overlooked knee motions but extensive focus on the ankle, hip, and step strategies. It is not well understood whether maintaining balance is regulated at the lower "muscular–articular" level of coordinating segment joints or at a higher level of controlling whole body dynamics. Whether balance control is to minimize joint degrees of freedom (DOF) or utilize all the available DOF also remains unclear. This study aimed to use a realistic musculoskeletal human model to identify multiple balance recovery strategies with a single optimization criterion. Movements were driven by neural excitations (which activated muscle force generation) and were assumed to be symmetric. Balance recoveries were simulated with forward-inclined straight body postures as the initial conditions. When the position of the toes was fixed, balance was regained with virtually straight knees and mixed ankle/hip strategies. Under a severely perturbed condition, use of the forward hop strategy after releasing the fixed-toes constraint indicated spontaneous recruitment or suppression of DOF, which mimicked functions of optimally computed CNS commands in humans. The results also indicated that increase/decrease in the number of DOF depends on the imposed perturbation intensity and movement constraints.

1. Introduction

Maintaining balance is one of the fundamental yet difficult skills for human beings. This is because most daily activities require this skill, but the human body is inherently unstable due to its mass distribution (Alexandrov, Frolov, Horak, Carlson-Kuhta, & Park, 2005; Loram & Lakie, 2002; Winter, 1995) and challenging perturbations from the environment make human beings highly susceptible to falls (Bhatt, Wening, & Pai, 2006). It has been well established that to maintain static standing balance, the vertical projection of whole body center of mass (COM) should be within the base of support (BOS) (Horak & Nashner, 1986; Winter, 1995). This idea was further extended for dynamic stability by considering COM velocity in establishing the margin of stability (Hof, Gazendam, & Sinke, 2005) and probability of recovery (Honarvar & Nakashima, 2014).

Based on the number of body segments/joints used for modeling human standing, different balance recovery strategies have been proposed. The ankle strategy involves applying joint torque mainly at the ankle and treating the whole body (excluding the feet) as a single inverted pendulum. The hip strategy is characterized by modeling the human body as a double inverted pendulum and balance is regulated primarily by applying hip joint torque (Runge, Shupert, Horak, & Zajac, 1999; Winter,

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1995). The hip strategy is employed when pure ankle strategy is ineffective in restoring balance, and the two joints often exhibit counter-phase motions (Horak & Nashner, 1986; Runge et al., 1999). With the added knee joint in the three-link human model (representing the shanks, thighs, and head–trunk–arms), the squat strategy was identified as flexing both the hip and knee joints (Atkeson & Stephens, 2007; Hemami, Barin, & Pai, 2006). In addition, the step strategy is regarded as a natural and preferred response especially when balance is severely perturbed (Cheng, Huang, & Kuo, 2014; Duncan, Studenski, Chandler, Bloomfeld, & LaPointe, 1990; Maki & McIlroy, 1997). Although rarely mentioned in balance literature, the hop strategy is employed more frequently than the step strategy in gymnastics under small landing errors (Marinšek & Čuk, 2010).

How knee motions participate in standing balance have generally been overlooked in the literature (Oude Nijhuis, Bloem, Carpenter, & Allum, 2007) despite the functional roles identified in balance maintenance (Gunther, Grimmer, Siebert, & Blickhan, 2009; Iqbal & Pai, 2000). Researchers have focused primarily on the ankle and hip strategies (Horak & Nashner, 1986; Runge et al., 1999; Winter, 1995) and used only the two joint motions in simulations (Colobert, Crétual, Allard, & Delamarche, 2006; Honarvar & Nakashima, 2014). However, noticeable knee motions have been reported in reaction to support surface movements (Creath, Kiemel, Horak, Peterka, & Jeka, 2005; Runge et al., 1999). Moreover, the capability of incorporating voluntary knee flexion into non-anticipatory balance corrections were demonstrated (Oude Nijhuis et al., 2007).

It remains unclear whether standing balance control requires minimizing joint degrees of free (DOF) or utilizing all the available DOF. A recent study addressed this issue by examining 24 healthy participants' reactions to knee joint perturbations in order to determine whether the control strategy preprograms a rigid knee (Di Giulio, Baltzopoulos, Maganaris, & Loram, 2013). The results revealed two possible patterns: the majority showed rigid knee configuration while 2 participants had larger knee flexion. However, this kind of experimental testing is subject to numerous uncontrollable variables including familiarity of the movement examined, personal habits, physiological and even psychological factors. To exclude these subjective influences, model simulation with one optimization criterion was carried out (Atkeson & Stephens, 2007). Despite the encouraging results of different balance strategies revealed by this unified approach, the unrealistic anthropometric parameters and joint torque actuation without neuromuscular modeling, nonetheless, decreased the likelihood of practical applications. Therefore, the purpose of this present study was to use a more realistic musculoskeletal human model with neural excitation/muscle activation inputs to identify different balance recovery strategies under various levels of perturbations. Since optimal simulations of multi-segment models have reproduced realistic human activities mimicking optimally coordinated movements generated by the central nervous system (CNS) (Hicks, Uchida, Seth, Rajagopal, & Delp, 2015; Mistry, Theodorou, Schaal, & Kawato, 2013), this approach was employed in the present study. For simplicity movements were assumed to be symmetric. Although this assumption excluded employing the step strategy, it was hypothesized that different balance strategies (including utilizing ankle/knee/hip motions and hopping for changing the BOS) could be identified by a single optimization criterion. Furthermore, similar to the proposition that coordination of redundant DOF is task dependent (Ko, Challis, & Newell, 2003), it was hypothesized that the number of DOF used in balance recovery depends on the intensity of balance perturbation and imposed movement constraints.

2. Methods

A musculoskeletal model from the open-source software OpenSim (Delp et al., 2007) was modified to perform balance recovery simulations in the present study. The model with a lumped head-arms-trunk (HAT), pelvis, thighs, shanks and feet segments (http://opensim.stanford.edu/) was originally used in gait and jumping simulations. To create more natural-looking movements, the current model allowed metatarsal-phalangeal joint motions. Although the model has two legs, for simplicity movements were driven by symmetric neural excitations (which could activate muscle contraction and consequently apply forces to the skeleton). Model DOF included pelvis positioning plus tilting, and flexion/extension of the hips, knees, ankles, and the metatarsal-phalangeal joints. Muscles used in the current model are listed in Table 1. Choice of these muscles was based on pilot simulations which ensured the model's capability of creating movements involving joint flexion/extension for squatting and returning to an upright stance.

System dynamics of rigid bodies were configured by generalized coordinates of joint angular displacements. The equations of motion are:

$$M(q)\ddot{q} + C(q,\dot{q}) + G(q) = J^{\mathrm{T}}(q)F_{\mathrm{c}}(q) + R(q)F_{\mathrm{MT}}(a), \tag{1}$$

where q, $\dot{q}$, and $\ddot{q}$ are joint angular displacements, velocities, and accelerations, respectively. $M(q)$ is the mass matrix of the multi-body system; C represents the Coriolis and centripetal force vector; G is the gravitational force vector; F_{c} is the ground reaction force (GRF) vector applied to the feet and J is the corresponding contact Jacobian matrix; R is the muscle moment matrix; F_{MT} is the muscle force vector which is a function of activation $a(t) \in [0,1]$ for activating muscle force generation and driving model movement. Contact between the feet and the ground was modeled according to Hunt and Crossley (1975).

A first-order differential equation modeled the delay between the neural excitation $u(t) \in [0,1]$ and muscle activation $a(t)$ (Anderson & Pandy, 1999):

$$\dot{a} = \begin{cases} (u-a)\cdot\left[\frac{u}{\tau_{\mathrm{act}}} + \frac{(1-u)}{\tau_{\mathrm{deact}}}\right] & \text{for } u \geqslant a \\ \frac{(u-a)}{\tau_{\mathrm{deact}}} & \text{for } u < a \end{cases} \tag{2}$$