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Does knee motion contribute to feet-in-place balance recovery?

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ABSTRACT

Although knee motions have been observed at loss of balance, the ankle and hip strategies have remained the focus of past research. The present study aimed to investigate whether knee motions contribute to feet-in-place balance recovery. This was achieved by experimentally monitoring knee motions during recovery from forward falling, and by simulating balance recovery movements with and without knee joint as the main focus of the study. Twelve participants initially held a straight body configuration and were released from different forward leaning positions. Considerable knee motions were observed especially at greater leaning angles. Simulations were performed using 3-segment (feet, shanks+thighs, and head+arms+trunk) and 4-segment (with separate shanks and thighs segments) planar models. Movements were driven by joint torque generators depending on joint angle, angular velocity, and activation level. Optimal joint motions moved the mass center projection to be within the base of support without excessive joint motion. The 3-segment model (without knee motions) generated greater backward linear momentum and had better balance performance, which confirmed the advantage of having only ankle/hip strategies. Knee motions were accompanied with less body angular momentum and a lower body posture, which could be beneficial for posture control and reducing falling impact, respectively.

1. Introduction

The ability to maintain balance is important. Body segment coordination for recovery from a perturbed to a balanced state is a primary objective in many human movements (Honarvar and Nakashima, 2014). For static standing balance, ground projection of body center of mass (COM) should be within the base of support (BOS) (Rietdyk et al., 1999). To explain the mechanism of maintaining standing balancing, the inverted-pendulum model has been proposed (Winter, 1995). Based on different movement patterns, either a single or double inverted pendulum model was employed, leading to the famous ankle or hip strategy for regaining feet-in-place balance (Rietdyk et al., 1999; Runge et al., 1999). The ankle strategy was characterized by moving the entire body as a single inverted pendulum with mainly ankle joint torque, while the hip strategy resembled a double inverted pendulum with anti-phase motions at the ankle and hip (Runge et al., 1999). Although the ankle strategy could suffice under moderate perturbations, inclusion of the hip strategy was often observed when balancing on a narrow or moving surface (Horak and Nashner, 1986; Runge et al., 1999).

Being one of the major lower limb joints, whether the knee joint also plays a key role in balance as the ankle/hip joints is an important question. However, how knee motions are involved in balance recovery were commonly overlooked (Oude Nijhuis et al., 2007). For example, most studies focused on the ankle/hip strategies (Horak and Nashner, 1986; Runge et al., 1999) and modeling with only the two joints (Colobert et al., 2006; Honarvar and Nakashima, 2014). In studying balance control with a 3-joint (ankle, knee and hip) sagittal model (Alexandrov et al., 2005), ignorance of knee motion was due to the passive mechanism of knee eigenmovement (movement along eigenvectors of the motion equation). Although the same 3-joint model for investigating trunk bending while maintaining balance revealed possible usage of the "knee strategy" under special conditions requiring rapid balance restoration (Alexandrov et al., 2001a), the effect of knee motion was deemed negligible compared to ankle/hip movements in the subsequent experiment (Alexandrov et al., 2001b). It appears from literature that knee motion is either negligible during moderate balance disruptions or only present during larger disturbances (Alexandrov 2001a, b; Di Giulio 2013). However, focusing on the ankle/hip strategies is also likely due to the early model (without knee joint) proposed for balance

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recovery (Winter, 1995). For instance, based on the single inverted pendulum model, participants were instructed to recover balance using the ankle strategy only (Robinovitch et al., 2002).

In fact, knee motions have been reported in response to support surface movement (Creath et al., 2005; Runge et al., 1999) and in control of standing (Hsu et al., 2007). Researchers have referred the balancing strategy of having knee motions as the “squatting strategy” (Hemami et al., 2006) because of flexion in all the lower limb joints and the resultant squatting. Moreover, since loss of knee proprioception has been shown to result in delayed balancing responses (Bloem et al., 2002) and the ability to incorporate voluntary knee flexion into automatic balance corrections has been revealed (Oude Nijhuis et al., 2007), it is reasonable to argue that knee flexion might have a functional role in balance recovery.

Although knee motion has been observed, mechanisms of recruiting this joint motion and whether it affects balance recovery performance are still unclear. Therefore, the purpose of this study was to experimentally monitor the amount of knee motion during balance recovery, and to explain its role by simulating balance recovery with and without locking the knee joint. In the pilot testing of four participants, although intra-subject performance was generally consistent, great inter-subject variability in movement patterns (including knee flexion magnitude) was found. Simply comparing movements with large/small knee motion could not exclude factors due to individual differences (e.g. training history or psychological issues). Experimentally comparing balance recovery with and without constraining knee motion was also considered. However, influences from being unfamiliar with or even fearing of losing the freedom of knee joint motion were inevitable. Thus, model simulations were used to clearly identify how knee motion contributes to balance recovery and to avoid subjective factors in human testing. Furthermore, these simulations were not intended for reproducing experimental results but for exploring alternative (and possibly better) movements within human limits.

2. Methods

2.1. Experimental testing

An experimental testing examined whether considerable knee motions could be observed in balance recovery. Twelve healthy male students aged 23.9 ± 1.9 years volunteered to participate in the experiment. The mean mass and height were 66.5 ± 9.8 kg and 1.72 ± 0.04 m, respectively. The University Research Ethics Committee for Human Behavioral Sciences approved the research objectives and experimental procedures. The written informed consent was obtained before data collection.

Participants first stood barefoot with both feet flat on a force plate (BP400600-2000, Advanced Mechanical Technology Inc., Massachusetts, USA) while maintaining a forward leaning initial posture. Leaning at three forward inclination angles (7.5°, 10°, and 12.5° relative to the vertical line) was achieved by a tether-release system. During tethering, participants were instructed to avoid excessive muscle force/joint torque production at the toes and ankle joint. The 7.5° initial leaning had the corresponding COM projection (COMP) within the BOS but near its frontal border, and the COMP positions were slightly in front of the BOS for 10° and 12.5° initial leaning. Detailed experimental settings were described elsewhere (Cheng et al., 2015). Since this study focused on balancing with lower limbs, arm motions were constrained by fixating crossed arms in front of the chest with elastic bandages. To have natural responsive movements, participants were not instructed on how to move the trunk/limbs (Corbeil et al., 2013) but only instructed to remain ground contact with both forefeet. Lifting the heels was allowed because this movement was observed in all pilot trials. After being held at each initial lean angle, the tether was unexpectedly released with a random delay of 1–5 s. Postural adjustments due to anticipation of the upcoming loss of balance were avoided by visually monitoring the ground reaction forces (GRF). Six trials were performed at each lean angle. Randomized lean angle conditions were employed to reduce the effect of learning or habituation.

After releasing the tether, a successful balance recovery generally included two stages: stopping forward falling with joint motions, and returning to upright standing, which have been termed the reflex and recovery stages, respectively (Abdallah and Goswami, 2005). Two Visualeyez motion tracking systems (VZ4000, Phoenix Technologies Inc., Canada) with sampling rate 100 Hz recorded positions of active markers located bilaterally at the 5th metatarsal phalangeal joints, ankles, knees (lateral femoral epicondyles), femur great trochanters, sacrum, and acromions. The built-in VZAnalyzer software calculated joint angles varying with time. The GRF data were sampled at 1000 Hz. Having considerable knee motion was defined by maximum knee flexion (from straight-knee configuration) of $>30°$ because flexion over this range could not be performed comfortably and effortlessly (Oude Nijhuis et al., 2007). Rather than suggesting knee flexion of $<30°$ being functionally insignificant, this choice of flexion angle of $>30°$ signified obvious inclusion of knee motion. This amount of knee flexion has also been suggested in clinical examination for ligament injury (Duffy and Miyamoto, 2010), indicating major distinction in knee joint force/torque production beyond this range of motion. Existence of considerable knee motion was determined by the statistical significance level of $p<0.05$.

2.2. Computer simulation

Two multi-segment rigid body models were used to investigate the effect of knee motion on balance recovery. Since no mediolateral movements were observed experimentally, only sagittal plane motions were assumed. The 3-segment (3S) model included the feet (without toes), shanks+thighs (ST), and head+arms+trunk (HAT). Knee motion was allowed in the 4-segment (4S) model by having both the shanks and thighs segments (Fig. 1). Segments were connected by frictionless hinge joints and movements were driven by joint torque generators. Each joint torque was the product of three variables: maximum isometric torque (depending on instantaneous joint angle), angular velocity dependence, and activation level. These variables were equivalent to the force–length curve, force–velocity curve and effective muscle activities, respectively, during muscle force generation (Pandy et al., 1990). Details (including segment length/mass/moment of inertia parameters and joint torque related variables) and validation for the present models have been provided elsewhere (Cheng, 2008; Cheng et al., 2008) and summarized in Appendix.

Because the observed actual movements generally included two stages (stopping forward falling followed by returning to upright stance) and success in the first stage guaranteed success in the second one, simulations were focused on the first stage. Starting from static forward leaning with a straight body posture, the goal was to let the COMP be inside the BOS. That is, to reposition the COMP behind the toe joint (the 5th metatarsal phalangeal joint) by adjusting the joint activation level (and consequently producing appropriate joint motions). This was achieved by defining the objective function as the sum of joint effort (calculated by integrating joint activation with time), final COMP position and joint angular velocity magnitude. Minimizing the objective function equivalently searched for an efficient way to move COMP backward and ended at a stable configuration (which guaranteed success in the second movement stage). To avoid falling backward, a lower bound of the COMP was set at 0.1 m behind the toe joint because the COMP rarely fell behind that position in experiments. Since excessive hip flexion causing the head to have a lower position than the hip was not observed in the experiment, the trunk angle (relative to the horizontal line) was constrained to be positive throughout simulations. In addition to the constraints for preventing joint hyperflexion/hyperextension, forces at the toe joint induced by segment motions were also constrained. Since the feet could not pull up the floor, vertical forces acting on the toe joint were constrained to be positive (but horizontal forces could be either positive or negative). At ground contact point the static friction coefficient was assumed to be 1 (O'Meara and Smith, 2001), and force magnitude in the horizontal direction was constrained to be less than that in the vertical direction.

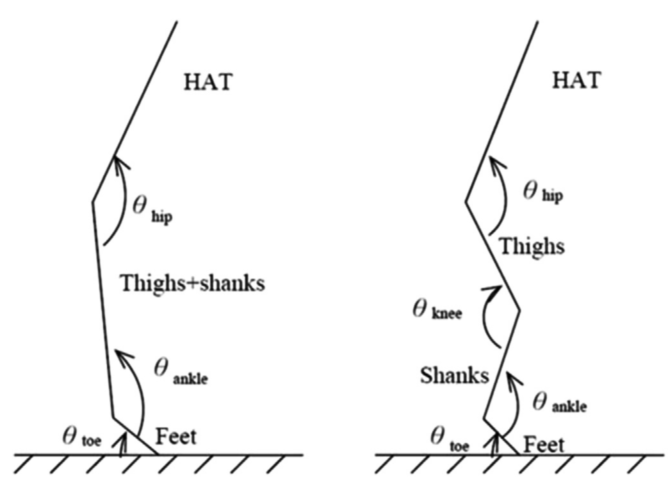

Fig. 1. The 3-segment (3S) and 4-segment (4S) models.