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Short communication

Frequency-dependent contributions of sagittal-plane foot force to upright human standing

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ABSTRACT

Quiet standing is a mechanically unstable postural objective that humans typically perform with ease. Control of upright posture requires stabilization of both translational and rotational degrees-of-freedom that is accomplished by neuro-muscular coordination. This coordination produces a force at the ground-foot interface (F) that is quantified by magnitude, direction (θ_F), and point of application (center-of-pressure, CP). Previous research has shown that the nervous system controls muscle activation such that CP motion occurs at both slow and fast time scales. However, it is unknown how θ_F varies with respect to CP and how that relationship varies across time scales. We present a novel method for assessing the frequency-dependent relative variation in θ_F and CP. The center-of-pressure (CP) and direction of the ground-on-foot force (F) in the sagittal-plane during quiet standing were decomposed into 0.2 Hz-width frequency bands within 0.4–8.0 Hz. The relation between the direction and CP was approximately linear with a slope positively related to frequency. These frequency-dependent features of F have critical implications for understanding balance strategy because the translational and rotational acceleration effects of F were coupled, but with opposite phasing at high versus low frequencies. Such results suggest a system tuned for one stability mode at low frequencies and another mode at higher frequencies. This frequency-wise approach to examining the translational and rotational effects of humans' preferred F may be useful for establishing balance rehabilitation metrics, directing study of the underlying neural mechanisms responsible for the observed coordination, and for setting a biometric standard to inform biomimetic prosthetics and robotics.

1. Introduction

Despite the apparent ease with which typical individuals stand, the control strategy is not fully understood (Emery, 2003; Geurts et al., 2005; Winter et al., 2003). Translation of correlational studies between clinical balance metrics and center-of-mass (CM) and center-of-pressure (CP) excursion, velocity, and variability to clinical utility and understanding of control have been limited (Palmieri et al., 2002). This may partially result from the failure of CM and CP location measures to quantify whole body angular motion which is a critical component of the task.

Consideration of the direction (θ_F) and CP of the ground-on-foot force (F) relative to the CM through time is a clear means of portraying the simultaneous control strategy of both degrees of freedom, because the direction of translational and angular acceleration of the whole body induced by F is readily evident (Fig. 1). This approach has been used to examine steady-state walking (Gruben & Boehm, 2012; Maus et al., 2010), providing insight on the regulation of sagittal-plane angular momentum and inspiring robotic control (Sharbafi et al., 2013). However, to the authors' knowledge, that approach has not been applied to quiet standing. Therefore, study objectives were to determine how humans direct sagittal-plane F relative to the CM during quiet standing and present that information in a format that can readily be interpreted with respect to whole body translational and rotational acceleration. The purpose is to facilitate intuitive understanding of how the physical requirements of standing are met

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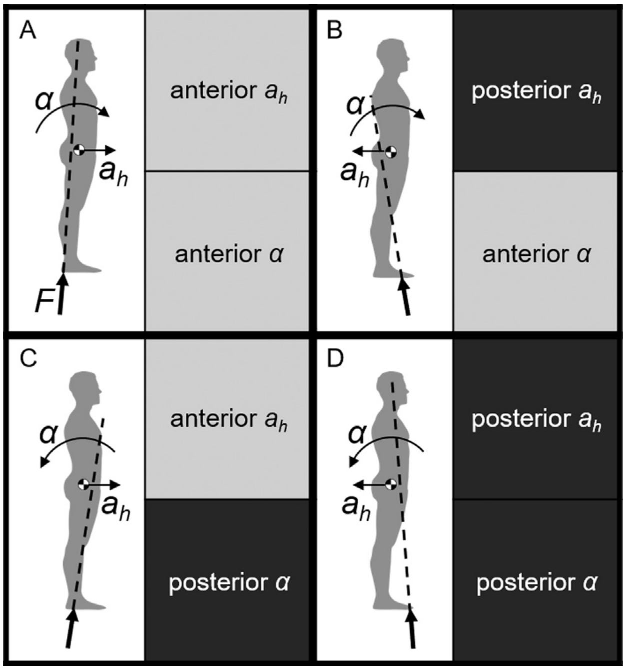

Fig. 1. By changing the ratio of torque between the hip, knee, and ankle joints, humans can produce F with different combinations of angular and translational effects on the whole body (Panels A–D). Panels A & D show cases of the CP for which F with a line of action passing posteriorly and anteriorly to the CM, respectively, couples translational and angular accelerations (a_h and α, respectively) with similar senses. Panels B & C show cases of the CP for which F with a line of action passing posteriorly and anteriorly to the CM, respectively, couples a_h and α accelerations with opposite senses. Note that F can also be directed through the CM for any CP, resulting in translational acceleration of the CM without whole-body angular acceleration.

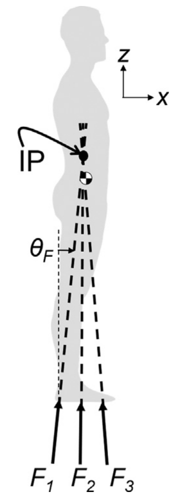

Fig. 2. When F lines-of-action from multiple time instances (F_{1-3}) are compiled and expressed relative to a reference point on the body (such as the CM), an intersection point (IP) may emerge.

and introduce a metric of neuromuscular control with potential for assessment of impaired individuals.

Over time, quiet standing requires translational and angular acceleration of the whole body to each average zero. Instantaneously, there will be non-zero whole-body and/or segmental translational and angular accelerations, however, the body coordinates these dynamics such that the CM remains near or above the base of support and the rotation of the long axes of the legs and torso do not deviate more than approximately 10 degrees from vertical. Despite anatomical and physiological constraints on the feasible possibilities for segmental accelerations (Kuo & Zajac, 1993), there are infinitely many ways to control F to meet these requirements. Humans likely select a pattern of F tuned to the inertial properties of the body (Creath, et al., 2005) with considerations for costs such as stability (Hof, Gazendam, & Sinke, 2005) and energy expenditure (Kuo, 1995).

Comparable physical requirements apply to steady state walking, where the pattern of F has been concisely characterized and is considered favorable for maintaining upright posture (Gruben & Boehm, 2012; Maus et al., 2010). In the sagittal-plane during the single-leg stance phase of steady-state walking, the F line-of-action relative to the CM approximately intersects a point on the body (IP, Fig. 2) located at a height greater than the CM. The intersection point behavior is an equivalent representation of the nearly linear variation between θ_F and the CP (for small $\theta_F \approx F_x/F_z$, Fig. 2). This IP F pattern, when used with appropriate CP shift, produces whole-body angular acceleration toward upright with increasing magnitude for increasing whole-body angular deviations from vertical (Gruben & Boehm, 2012; Maus et al., 2010). Additionally, translational acceleration of the CM towards the center of the base of support increases with lateral deviation. Falls typically occur with similarly-coupled whole-body forward or backward angular and translational velocities (Runge, et al., 1999), and thus, the IP behavior in walking has been considered useful for remaining upright (Gruben & Boehm, 2012; Maus et al., 2010). The presence of IP behavior during standing could similarly explain resistance to small translational and angular perturbations.

The relation between the unfiltered θ_F and CP during quiet standing is not linear (Fig. 3). Further inspection reveals brief, approximately linear, relations between θ_F and CP. This character suggested that a frequency-wise analysis of the relation between the θ_F and CP is warranted. Additionally, previous research (Zatsiorsky & Duarte, 1999, 2000) quantified the horizontal component of F (F_{xy}) during quiet standing and reported a correlation between F_{xy} and CP for frequencies above 0.4 Hz. Therefore, we hypothesized that the >0.4 Hz portion of CP and θ_F would vary linearly such that the F lines-of-action are well-characterized by an intersection point (IP).

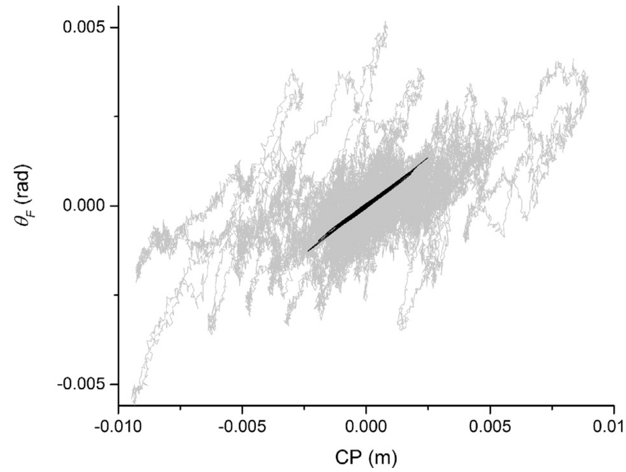

Fig. 3. Representative data for a single participant during 50 s of quiet standing shows the relation between sagittal-plane CP and θ_F. Unfiltered data (light gray) appear as a cloud. As an example, bandpass filtering each of the CP and θ_F signals with a 0.5–0.7 Hz band reveals an approximately linear relation (dark trace).

2. Methods

2.1. Experiment

Participants stood quietly with their arms at their sides, feet together, viewing an 'X' at head height 1 m anterior. One 50 s trial was recorded per participant on a 6-axis force-plate (1000 Hz, 16-bit A/D converter, National Instruments, Austin, TX, USA). The 50 s trial length provided reliable frequency content down to 0.4 Hz and was motivated by preliminary 10 min trials yielding no correlation between θ_F and CP at frequencies below 0.4 Hz. Statements regarding signal content below 0.4 Hz are not addressed in this manuscript.

2.2. Participants

Ten non-impaired adults participated in this experiment (4 female and 6 male, aged 24.2 ± 10.3 years). Participant exclusion criteria consisted of neuromuscular diseases or current musculoskeletal injuries. Participation was voluntary, written consent was obtained, and the protocol was approved by the University of Wisconsin Institutional Review Board. CM height was estimated as 56% of body height with 1% standard deviation (Croskey, et al., 1922). CM height was assumed constant across time, because vertical excursion due to body sway was on the order of sub-millimeter (assuming less than 0.5° ankle sway (Loram, Maganaris, & Lakie, 2005)) which is more than an order-of-magnitude smaller than the uncertainty in the CM height estimate.

2.3. Analysis

The anterior-posterior CP and θ_F signals were bandpass-filtered (zero-lag, 2nd-order Butterworth) in bands of 0.2 Hz width centered on frequencies from 0.5 to 7.9 Hz at 0.2 Hz increments (38 non-overlapping bands). Multiple bands were analyzed to resolve IP height variation with frequency. The power spectrum was calculated for CP and θ_F (Hann window) to report distribution of signal energy. Student's t-tests were used on each frequency bin to test for significant IP height difference from the height of the CM (Bonferroni correction: alpha-level of 0.05 adjusted for 38 tests, $p < 0.0013$).

The first principal component (1PC) of the relation between θ_F and CP (each standardized by dividing by the respective standard deviation) in each frequency band was calculated. The height of the IP was calculated as the inverse slope of the 1PC. Note that any frequency-dependent *horizontal* IP offset from the CM cannot be resolved due to the bandpass filtering. The *variance accounted for* (VAF) by the 1PC was calculated to assess the ability of the intersection point to describe the relationship between CP and θ_F.

3. Results

In each frequency band, *F* was well-described through time as having lines-of-action intersecting near an IP (Figs. 3–5). The height of the IP varied systematically with frequency, with height above the CM at low frequencies and below the CM at high frequencies (Fig. 6). The IP height was not significantly different from CM height between 1.2 and 2.6 Hz.

The frequency-specific relation between CP and θ_F was well-described as approximately linear (high VAF, Fig. 5) indicating that an intersection point was a good predictor of θ_F from CP. Limitations on the reliability of the frequency-based analysis could arise due to the finite measurement duration. Reliability is expected to be better for higher frequencies due to the larger number of cycles included. However, the data showed that a line describes the θ_F versus CP behavior best at low frequencies and less accurately at higher frequencies (Fig. 5). Possible explanations include decreased linearity in the behavior and lower signal power at high frequencies (Fig. 8). Longer pilot trials of 600 s provided IP heights similar to those reported here but incurred larger risks of fatigue and boredom that could interfere with the desired task objective. Those considerations lead to the chosen 50 s duration.

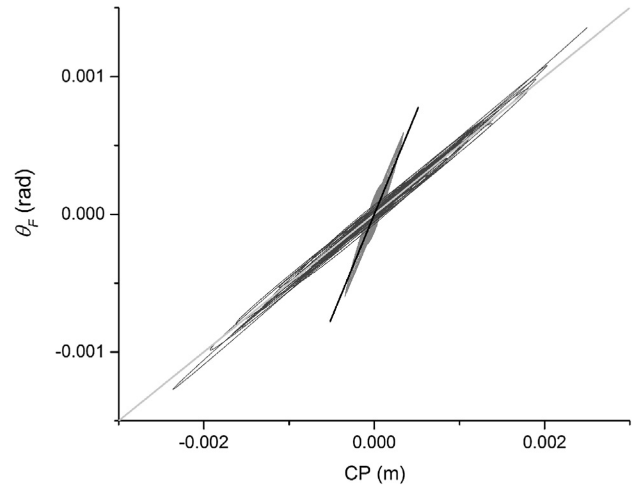

Fig. 4. The slope of the θ_F versus CP relation varies by frequency. For example, in a representative trial (Fig. 3), the slope of the relation in the 2.6–2.8 Hz band (light gray) is steeper than in the 0.5–0.7 Hz band (dark gray). The high variance accounted for by a 1PC for each of these bands (Fig. 5) indicates that the relation was well-described by an IP. The steeper slope of the 1PC of the higher frequency component (dark line) is indicative of a lower IP height compared to the 1PC of the lower frequency component (light gray line).

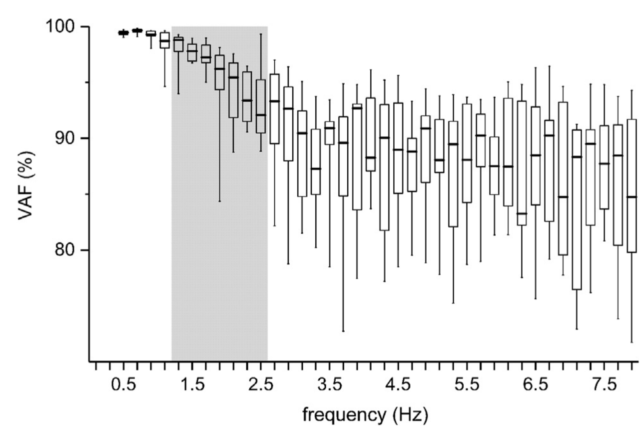

Fig. 5. The relation between θ_F and CP is well-described as linear, as quantified by the high mean percent *variance accounted for* (VAF) for all participants (median, boxed quartiles, whiskers range). A single IP outlier was removed from further analysis due an intersection point being a poor descriptor of the F behavior (VAF < 0.7) for one participant in the 3.4–3.6 Hz frequency band.

4. Discussion

During quiet standing, humans must use hip, knee, and ankle torque coordination to produce *F* appropriate for maintaining rotational and translational time-averaged equilibrium. This study determined the structure of that coordination above 0.4 Hz and revealed a consistent variation with frequency across participants. The IP behavior is an equivalent representation of the nearly linear variation between θ_F and the CP (for small $\theta_F \approx F_x/F_z$, Fig. 2). The IP behavior characterizes a controller with three principal features: (1) systematic covariation between θ_F and CP, (2) interpersonal consistency of the functional relationship between IP height and frequency, and (3) the ability to stabilize the range of combined

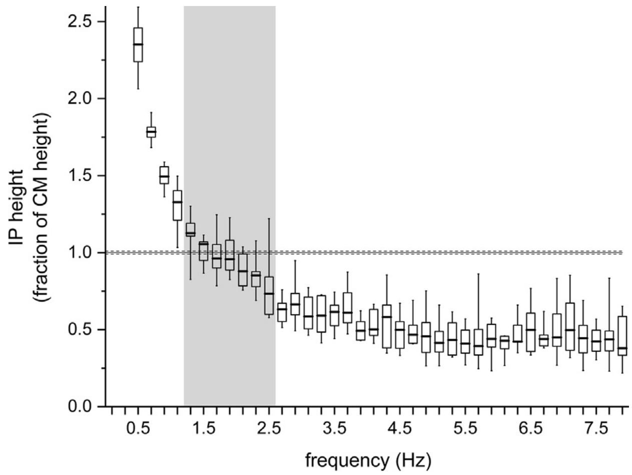

Fig. 6. The IP height (median, boxed quartiles, whiskers range) decreased as frequency increased. CM height is indicated as a line at 1.0 with dashed lines indicating ±1% standard deviation. The region from 1.2 to 2.6 Hz is shaded to highlight the region where mean IP height was not significantly different from CM height.

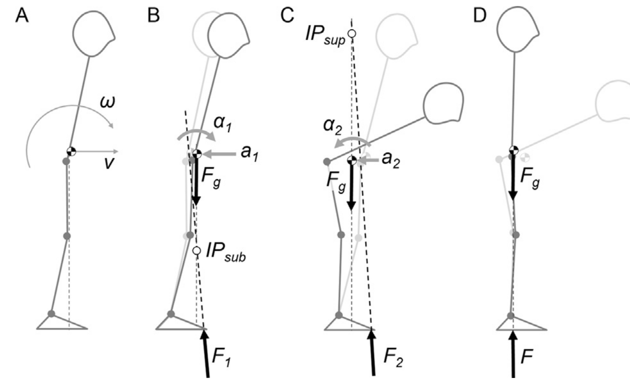

Fig. 7. If the body has translational and angular velocity (Panel A: v, ω) that will lead to a fall if not arrested (forward to the right in this example), the body must respond with appropriate translational (to the left) and rotational (counter-clockwise) acceleration using the force of the ground on the feet (F). Finite foot-length limits the ability to accomplish both simultaneously with larger perturbations, which necessitates prioritization of balance strategies based on the following physical conditions: Without stepping or pronounced rotation of some body segment (e.g. arm wind-milling), it is not possible to remain standing if the gravity line is outside the base-of-support. This places a fairly rigid constraint on the anterior-posterior location of the CM. On the contrary, humans can remain standing with whole body orientation significantly deviated from anatomical position (e.g. standing on one leg with the other leg, arms, and torso all oriented horizontally or even nearly inverted) at the cost of increased joint torque. The body can respond quickly (high-frequency) by using a relatively large translational acceleration consistent with a sub-CM IP (panel B: a_1, IP_{sub}) to attend to the more urgent problem of translational motion at the cost of angular acceleration in the opposite direction of upright (α_1). That angular motion result in postural change (panel C). A slower acting (low-frequency) response consistent with a supra-CM IP (panel C: IP_{sup}) provides relatively slower translational correction ($a_2 < a_1$) but provides the angular acceleration (α_2) appropriate for resuming quasi-static upright posture (panel D).

translational and angular balancing demands present in quiet standing.

The mechanics of the task of quiet standing do not prescribe that an IP exists nor that the IP has any specific height at any particular frequency. Many solutions of $F(t)$ within the feasible acceleration space for humans (Kuo, 1995) can meet the zero time-average requirements of translational and rotational acceleration required for standing. The biomechanical constraints of muscle strength, joint range of motion, and finite foot length which limits CP excursion but not F direction, impose boundaries on feasible solutions for remaining upright, but they do not constrain humans to a unique solution. The preference for IP behavior reflects a neural strategy that likely weighs costs such as metabolic expenditure, ease of control structure, deviation from upright posture, and controller stability in the context of the inertial and anatomical properties of the multi-segmented human-body linkage.

The height of the IP relative to CM is important to consider because it dictates the sign coupling between changes in translational and angular acceleration that occur with CP shift. Typically, as a person begins to tip over forward, ankle torque is used to shift the CP forward toward the toes. Backward translational and backward angular accelerations of the whole body are needed to prevent falling (the opposite holds for backward tipping). With forward CP shift, an IP with height above the CM increases the backward translational and backward angular acceleration effect of F. Thus, an IP higher than the CM may be viewed as stabilizing when the body acts approximately rigidly above the ankle as is common in the presence of minimal perturbation (referred to as a kinematic "ankle strategy" (Kuo, 1995; McCollum and Leen, 1989; Runge et al., 1999)). This argument supports the use of the observed ankle strategy for small disturbances despite the speed and neural delay robustness associated with a hip actuation strategy (Kuo, 1995; McCollum and Leen, 1989).

Finite foot length, however, limits ankle strategy (Kuo, 1995). For example, when falling forward, a person cannot simultaneously produce F with a relatively large posterior horizontal component and a large posteriorly pitching torque about the CM. The CP of that F would need to be anterior to the toes. However, a sequence of balancing strategies (Fig. 7) that utilizes fast-acting sub-CM IP behavior followed by slow-acting supra-CM IP behavior can be used for recovery. This strategy is possible without modifying the control strategy observed in this study, but simply by shifting energy to higher, then lower, frequencies (Cheng & Yeh, 2015). In

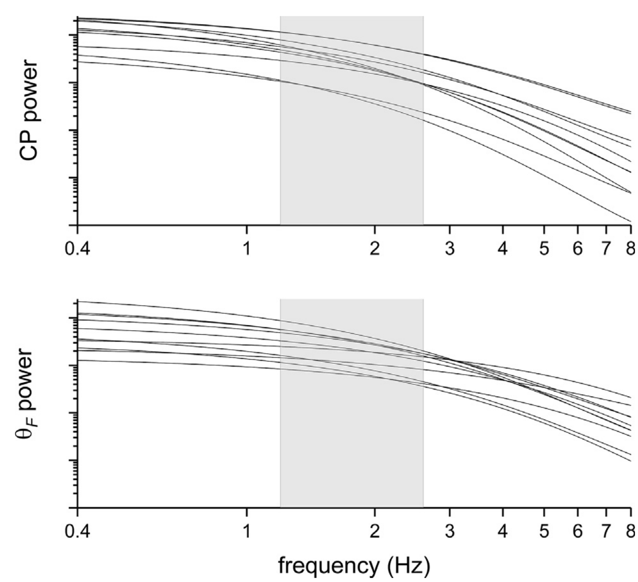

Fig. 8. Power distributions for each participant above 0.4 Hz for both CP and θ_F show decreasing signal energy with increasing frequency. The approximate frequency band that produced supra-CM IP's (between 0.4 Hz and 1.2 Hz, the lower boundary of the CM crossing band) contained 86.8% of the CP and 56.9% of the θ_F signal energy above 0.4 Hz. The frequency band with IP height statistically indistinguishable from CM height (1.2–2.6 Hz) contained 10.7% of the CP and 22.2% of θ_F signal energy above 0.4 Hz. Thus, 2.5% and 20.9% of CP and θ_F signal energies contributed to sub-CM IP locations (above 2.6 Hz). The region from 1.2 to 2.6 Hz is shaded to highlight the region where mean IP height was not significantly different from CM height. Both scales are logarithmic. To present the primary features of the power distribution, each curve drawn is a second-order polynomial fit to each participant's power spectrum.

addition, the relatively low signal power contributing to high frequency sub-CM IP behavior (2.5% of CP and 20.9% of θ_F signal energy above 0.4 Hz) is consistent with the reduced need for this strategy during quiet standing. Future studies should characterize the IP during perturbation or other conditions which challenge balance, where a sub-CM IP may be of greater necessity.

The use of ankle joint actuation for slow movements to attenuate small disturbances and hip joint actuation for fast movements to attenuate larger disturbances has been observed by (Cheng & Yeh, 2015), and is consistent with observations of (Kuo, 1995). In addition, Kuo approximated some of that human behavior with an optimal control model. The model was primarily driven by a penalty for CM excursion and deviation from upright stance, with consideration for anatomical constraints, energetic cost, and the feasible acceleration space of the lower limb joints. While that work did not identify the explicit control objectives of the central nervous system, the ability of the optimization algorithm to arrive at behavior compatible with the IP behavior observed in the current study highlights the factors that likely contribute to the presence of an IP and the shape of the IP vs. frequency curve. It is unclear how the filter properties of the body's mechanics, sensory, and processing systems constrain the IP and how much flexibility humans have in trading off stability and energetic costs. Implementation of a model similar to that of (Kuo, 1995) with the additional constraints of matching IP behavior could reveal answers to these questions, and thus, provide insight on the nervous system's control structure and priorities. The model could also be used to investigate how the interaction of neural control and mechanics influence IP behavior and the associated stability implications.

In a similarly three-segmented model, (van der Kooij et al., 1999) used optimal estimation theory to validate a model of upright sagittal stance capable of quantifying the effects of sensory uncertainty on standing stability. In their model, it was shown that inherent systemic time delays in proprioceptive transmission produce an unstable system in the absence of a predictive state controller. Optimal estimation of that necessary body state prediction was based on sensory uncertainty. The observed IP behavior (particularly the systematic preference for employing distinct coordination patterns at specific frequencies) may be a feedforward means of enhancing body state prediction. Further work should investigate the sensitivity of the IP curve to manipulation of sensory feedback reliability and environmental perturbation uncertainty to test this relationship. For example, small destabilization of the support surface could reduce the predictability of distal proprioceptive feedback and encourage a stiffening strategy to act as a one-segment inverted pendulum model (ankle strategy (Kuo, 1995; McCollum & Leen, 1989). This would help stabilize the head, where the more reliable visual and vestibular feedback modalities can be sourced. Behaviorally this would appear as reliance on ankle strategy and would be expected to correspond to accentuation of behavior with IP above the CM.

This study reveals an interesting F pattern above 0.4 Hz, but it is important to also consider the implication of content below this frequency. The coordination strategy contributing to accelerations below 0.4 Hz is beyond the scope of this study but should be characterized alongside the results of the current study to yield a more complete understanding of control.

IP characterization of standing may be a powerful, simple tool for characterizing standing balance. In comparison, traditional methods that only characterize CP dynamics (Benda et al., 2008; Collins and De Luca, 1993; McClenaghan et al., 1994; Palmieri et al., 2002) provide limited perspective on motor behavior. Motor control changes occurring with aging and disease should be examined with the current approach to provide insight on possible mechanisms of change and impairment and detect trends toward balance problems before clinically significant behaviors are observed. Characterization of IP behavior in each foot for unimpaired, perturbed, aged, and diseased populations could provide additional insight. While this study assumes F_z is relatively constant, effects due to F_z variation between limbs could also factor into IP behavior and should be investigated. Overall, IP characterization provides insight on control, ease of standing, recovery from perturbation, and may have clinical relevance in treatment and detection of disease.

Conflict of interest statement

The authors report no conflict of interest.

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