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Explaining Parkinsonian postural sway variabilities using intermittent control theory

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ABSTRACT

Postural impairment due to neuro-degenerative disorders such as Parkinson's Disease (PD) leads to restricted gait patterns, fall-related injuries, decreased mobility, and loss of functional independence. Though several clinical and posturographic studies have attempted to reveal the complex pathophysiology involved in PD, the diversity of Parkinsonian population makes them unclear and sometimes even contradictory. For instance, studies related to the Center of Pressure (CoP) sway during quiet stance in PD patients highlight both increase and reduction of magnitude in contrast to age-matched healthy individuals. A possible explanation for this contradiction is presented in this article. While the presence of intermittent control has been observed in postural control in human quiet stance, we hypothesize that one of the factors that affects postural instability in PD might be the increase in intermittency in active feedback control. Using a simulation model representing the Anterior-Posterior dynamics of human quiet standing, the intermittent control strategy is first contrasted against continuous control strategy in terms of stability, energy efficiency and settling time, thus establishing the inherent advantages of an intermittent control strategy. Further, the ability of the intermittent control strategy to explain several clinical observations in PD is demonstrated. An experimental pilot study is also conducted to support the simulation study, and several body sway parameters derived from recordings of CoP are presented. The presented results are in close agreement with reported clinical observations and may also prove useful for the assessment of disease progression and future fall risk.

1. Introduction

Postural stability in humans is vital in maintaining a vertically upright orientation while performing voluntary day to day activities. During human upright stance, the neuro-muscular, vestibular, visual and proprioceptive feedback simultaneously operate to counteract gravity and disturbances (Park et al., 2015). Improper balance control results in restricted gait patterns, fall-related injuries and decreased mobility. In Parkinson's Disease (PD) one observes such instabilities in which the motor disabilities are highly pronounced (Park et al., 2015; Kim et al., 2013; Alcock et al., 2018).

Direct clinical measurements confirm the obvious fact that the intrinsic ankle stiffness is not independently sufficient to ensure postural stability (Loram and Lakie, 2002; Asai et al., 2009; Gawthrop et al., 2011). Computational works (Gawthrop et al., 2011; Loram et al., 2011) suggest that, although active continuous feedback controller can stabilize the upright posture, it results in a narrow stability margin in presence of finite neural delay. Moreover, Bottaro et al. (2005), Bottaro et al. (2008), Gawthrop et al. (2011) suggest that human quiet stance exhibits bounded behaviour but not asymptotic convergence. The possibility that this may be due to neural noise sources (Maurer and Peterka, 2005) seems unlikely as the experimentally recorded noise magnitudes are not significant enough to produce the observed sway (Bottaro et al., 2005; Bottaro et al., 2008; Yamamoto et al., 2011). On the other hand, evidence of intermittency in human control actions were reported in the early 1950's (Craik, 1947), has been extensively demonstrated by experiments on human servo mechanisms (Bottaro et al., 2005; Fitzpatrick and McCloskey, 1994).

Intermittent control provides appropriate framework for formulating postural sway in human quiet stance as a consequence of natural human physiology (Craik, 1947; Asai et al., 2009; Loram and Lakie, 2002; Loram et al., 2012; Gawthrop et al., 2011; Loram et al., 2011; Tanabe et al., 2017). Intermittent control is a control approach in which the observation is continuous, but

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the actions are intermittent based on certain threshold criteria. In other words, the control actions are only switched on when the observed variables cross certain threshold. In contrast to continuous control, this feedback control approach is shown to be less sensitive to parameter variations (Bottaro et al., 2005; Bottaro et al., 2008; Asai et al., 2009; Loram et al., 2011). While the presence of intermittency in human postural control is established, why an intermittent strategy is advantageous over a continuous strategy has not been studied.

Several clinical and posturographic studies attempt to reveal the complex pathophysiology involved in PD, but the diversity of Parkinsonian population makes them unclear and sometimes contradictory (Park et al., 2015; Kim et al., 2013; Lauk et al., 1999). According to Park et al. (2015), Kim et al. (2013), Matsuda et al. (2016), Yamamoto et al. (2011), Rocchi et al. (2002), Termoz et al. (2008), PD patients during quiet standing exhibit reduced Center of Pressure (CoP) fluctuation as compared to healthy individuals due to increased muscle stiffness. In contrast, increased amplitude of spontaneous CoP sway are reported in Kim et al. (2013), Yamamoto et al. (2011), Blaszczyk et al. (2007), Raymakers et al. (2005). It has also been reported that the sway characteristics in PD patients may not be significantly different from age-matched healthy individuals (Kim et al., 2013; Lauk et al., 1999). Further, there are evidences of forward shift in mean CoP (stooped posture) (Park et al., 2015; Blaszczyk et al., 2007; Termoz et al., 2008). These inferences indicate that there might be several types of postural impairments involved in PD or the same impairment might have been perceived differently in different research studies. Hence the quantification of postural instability still remains a challenge (Lauk et al., 1999; Rocchi et al., 2006).

The present study aims to gain insights into the advantages of intermittency on balance control strategies involved in quiet standing of healthy and PD individuals. The two key contributions of this paper are as follows. The first contribution is to provide an analysis of intermittent control strategy and contrast it against continuous control strategy for human quiet standing in terms of stability, energy efficiency and settling time. This provides an understanding of why intermittency is advantageous in postural control. The second contribution is to establish that intermittent control strategy is able to explain several clinical observations in PD including some of the seemingly contradictory observations, and thus may prove useful for the clinical posturography. The analysis provided in this paper is based on the CoP in AP (anterior-posterior) direction, however, the insights generated are not limited only to AP direction.

2. Dynamics of human upright posture and intermittent control

In this section, we approximate the dynamics of human upright standing through an appropriately simple model that allows us to qualitatively analyse the key stability parameters and performance of sensorimotor control. We note that the objective is not to develop a model for accurate quantitative predictions and therefore avoid adopting an unnecessarily detailed model.

As observed in Asai et al. (2009), the human upright posture is characterized by saddle-type instability having both stable and unstable manifolds in its phase portrait. Such dynamics may naturally be represented as a single inverted pendulum approximation (Bottaro et al., 2005; Bottaro et al., 2008; Asai et al., 2009; Loram et al., 2011; Chagdes et al., 2016a; Yamamoto et al., 2011) with passive and active torques applied at the hinge. This replicates the control of AP sway of the upright human body through passive and active torques at the ankle. Given the objective of this study, focusing on only the ankle joint suffices for our purposes. The equations of motion for the inverted pendulum model may be derived as

$$(I + mh^2)\ddot{\theta} - mgh\sin\theta = \eta + M_{\text{ankle}}, \tag{1}$$

where $M_{\text{ankle}} = M_{\text{passive}} + M_{\text{active}}, M_{\text{passive}} = k\theta + c\dot{\theta}$ and $M_{\text{active}} = K_P\theta + K_D\dot{\theta}$. The sway angle θ represents the angular deviation of the center of mass from the upright position and is assumed to be continuously estimated by the sensory organs. The body is perturbed by noise η that comprises of both the internal hemodynamic noise and external disturbances, and regulated by corrective ankle moment M_{ankle} that consists of the active and passive components. The passive moment M_{passive} includes the torque due to passive muscle stiffness and damping and acts without any intervention from the nervous system (Loram and Lakie, 2002). The active moment M_{active} arises due to the time-delayed feedback that corresponds to muscle action commanded by the sensorimotor system in response to proprioceptive, vestibular and visual sensory information. This active corrective action at an instant t is dependent on the sway angle and sway velocity at an earlier time $(t - \tau)$. Here, τ accounts for the finite neural transmission and processing delay necessary for a postural correction to occur after sensing (Chagdes et al., 2016a; Yamamoto et al., 2011; Peterka, 2002). Here τ for a healthy individual is considered to be 0.2 s (Asai et al., 2009; Loram et al., 2012; Maurer and Peterka, 2005).

Fig. 1(a) is a block-diagrammatic representation of the feedback control of the human upright posture. In case of intermittent control, the threshold criteria controls the switch determining when the active moment M_{active} is on or off. In the case of continuous control, the active torque M_{active} operates all the time. The choice of model parameters considered for subsequent simulations (Table 1) are based on standard human data (Chaakrabarti, 1997) consistent with Asai et al. (2009), Chagdes et al. (2016a), Chagdes et al. (2016b), Peterka (2002), Maurer and Peterka (2005).

The CoP sway in human quiet stance may be characterized by parameters such as the sway area, length and range in AP direction, all of which provide excellent measures for postural instability (Lauk et al., 1999; Bottaro et al., 2008; Chagdes et al., 2016a; Chagdes et al., 2016b; Matsuda et al., 2016; Raymakers et al., 2005; Maurer and Peterka, 2005). The CoP information can be evaluated from sway angle θ as (Chagdes et al., 2016b)

$$CoP = \frac{(-I\ddot{\theta} + mgh\sin\theta + m_f g d_f - h_f mh(\ddot{\theta}\cos\theta - \dot{\theta}^2\sin\theta))}{(-mh\ddot{\theta}\sin\theta - mh\dot{\theta}^2\cos\theta + mg + m_f g)}. \tag{2}$$

3. Advantages of intermittent control

In this section, we examine stability and energy efficiency of intermittent control in comparison to a continuous control strategy, thus helping understand advantages of intermittent control.

3.1. Stability and intermittent control

For analysing stability, the root locus for the inverted pendulum model (1) with a small angle approximation is plotted in Fig. 1(b) with varying active proportional gain K_P (assuming a simple proportional controller). Here, the delay τ is modelled with a first-order Pade approximation. It is observed that the stability of the system is limited to the range $165 < K_P < 692$. Since, the reported value of K_P for healthy individuals is $\simeq 527.936$ Nm $(\sim 0.8 \times mgh)$ (Asai et al., 2009; Chagdes et al., 2016a; Peterka, 2002; Maurer and Peterka, 2005), it confirms that the human quiet standing with continuous (feedback) control would be stable.

For simplicity, the threshold criteria for intermittent control is imposed on the sway angle θ, that is, the active controller is turned on when θ exceeds a certain threshold and remains off otherwise. Based on the experimental observations in Fitzpatrick and McCloskey (1994) we choose this threshold value to be

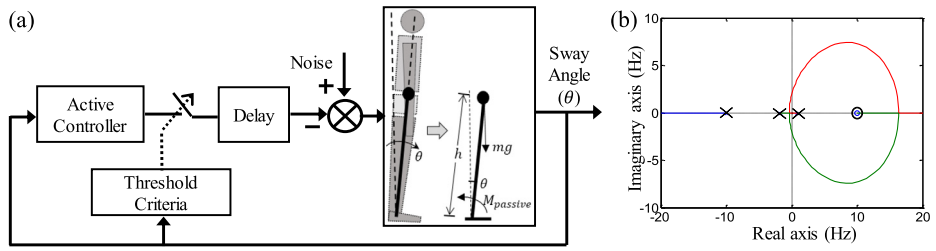

Fig. 1. (a) Model of human quiet stance control as an inverted pendulum with intermittent active feedback controller and delay and (b) root locus plot of human quiet stance model with varying active feedback controller gain with out intermittency.

Table 1
Human body parameters used for modelling human quiet stance as a simple inverted pendulum.

Parameters	Symbol	Values	Units
Body mass	m	67.27	kg
Body inertia	I	22.42333	kg – m^2
Body height	h	0.89	m
Foot mass	m_f	2	Kg
Foot height	h_f	0.085	m
Foot width	d_f	0.05	m
Passive stiffness	k	527.93	Nm
Passive damping	c	98.99	Nm – s
Active proportional gain	K_P	527.93	Nm
Active derivative gain	K_D	32.99	Nm – s
Delay	τ	0.2	s
Noise	η	zero mean (0.4 STD)	Nm
Acceleration due to gravity	g	9.81	m/s^2

$\theta_{th} = \pm 0.002$ rad. The CoP sways (in AP directions), phase portraits, the active controller effort and corresponding CoP histograms are presented in Fig. 2. Note that with increase in active controller gain or delay, continuous control results in unbounded sway. While on the other hand, stable CoP trend due to intermittent control matches well with experimental observations (Bottaro et al., 2005; Loram and Lakie, 2002; Gawthrop et al., 2011; Loram et al., 2011; Tanabe et al., 2017). This reflects the inability of continuous control to provide robustness to variations that are likely to occur both in healthy and diseased conditions.

It is also worth noting that in case of intermittent control, a bimodal distribution is observed in the CoP histogram which is consistent with Bottaro et al. (2005), Bottaro et al. (2008), Gawthrop et al. (2011), but in case of continuous control, the distribution appears to be unimodal. Due to unboundedness, histograms are omitted for the unstable responses.

3.2. Energy efficiency and settling time

The effort by the active controller may be used for estimating the energy expenditure involved during control action. Since, 60 – 120 s of time period is considered in standard practices of clinical posturography (Craik, 1947; Yamamoto et al., 2011), the energy expended by the active controller (E_a) is computed as the incremental sway angle multiplied by the active controller effort integrated over a time period of 100 s. E_a for intermittent as well as continuous control are evaluated (Fig. 2). It is observed that the net effort by the active controller in case of intermittent control is lower in comparison to continuous control. For example, in case of healthy individuals ($K_P = 527.9$ Nm, and $\tau = 0.2$ s) the intermittent controller expends a total E_a of 0.638 mJ as opposed to 3.4 mJ by continuous control. To study the effect of intermittency on settling time, the posture model is tested against an impulse force. The effort by active controller and the corresponding CoP sways are plotted in Fig. 3. It is apparent that with intermittent control (in blue) the CoP settles significantly faster than continuous control (in black).

4. Intermittent Control in Parkinsonian quiet stance

In this section, we first highlight key clinical observations in PD related to control of upright posture, then examine how intermittent control is able to explain these observations. We primarily focus on the following 6 clinical observations related to the dynamics of underlying postural sway characteristics that appear most relevant to our analysis.

4.1. Clinical observation

4.1.1. Increase in postural sway

It is reported in Kim et al. (2013), Yamamoto et al. (2011), Blaszczyk et al. (2007), Raymakers et al. (2005) that an increased sway range (~5–10 mm) and area (~ 150 – 400 mm^2) in comparison to healthy individuals are the characteristics of parkinsonian postural instability. According to Blaszczyk et al. (2007), even PD patients not complaining of balance disorder have higher sway dimensions in comparison to healthy individuals.

4.1.2. Decrease in postural sway

Often subjects with PD stand with a narrow CoP sway (Park et al., 2015; Kim et al., 2013; Yamamoto et al., 2011; Termoz et al., 2008) and exhibit increased stiffness in the leg and pelvis complex (Park et al., 2015; Lauk et al., 1999; Chagdes et al., 2016a; Chagdes et al., 2016b; Rocchi et al., 2002; Wright et al., 2007) though the exact amount of decrease in sway differs within the studies. Note that the observation of decreased postural sway is in contradiction to the earlier set of studies observing increased postural sway.

4.1.3. Forward shift of mean CoP

In parkinsonian quiet stance, a forward shift of mean CoP (stooped posture) is observed (Park et al., 2015; Kim et al., 2013; Blaszczyk et al., 2007; Termoz et al., 2008; Wright et al., 2007). In Blaszczyk et al. (2007) the mean CoP position in the AP direction is found to be 102.5 ± 12.0 mm in the PD group, which is significantly higher than the control subjects, that is 98.8 ± 11.2 mm. Here the shift magnitude varies with the disease severity, but on the average is seen to be of the order of 4 mm.

4.1.4. Increase in muscle activity

It is extensively documented that subjects with PD exhibit increased muscle activity while standing upright (Tanabe et al., 2017; Termoz et al., 2008; Wright et al., 2007). According to Jacobs et al. (2005), PD individuals tend to activate (direction specific) muscles 50% more than healthy individuals. Further, patients with mild level of PD exhibit lower muscle activity with intermittent activation patterns whereas severe PD patients exhi-

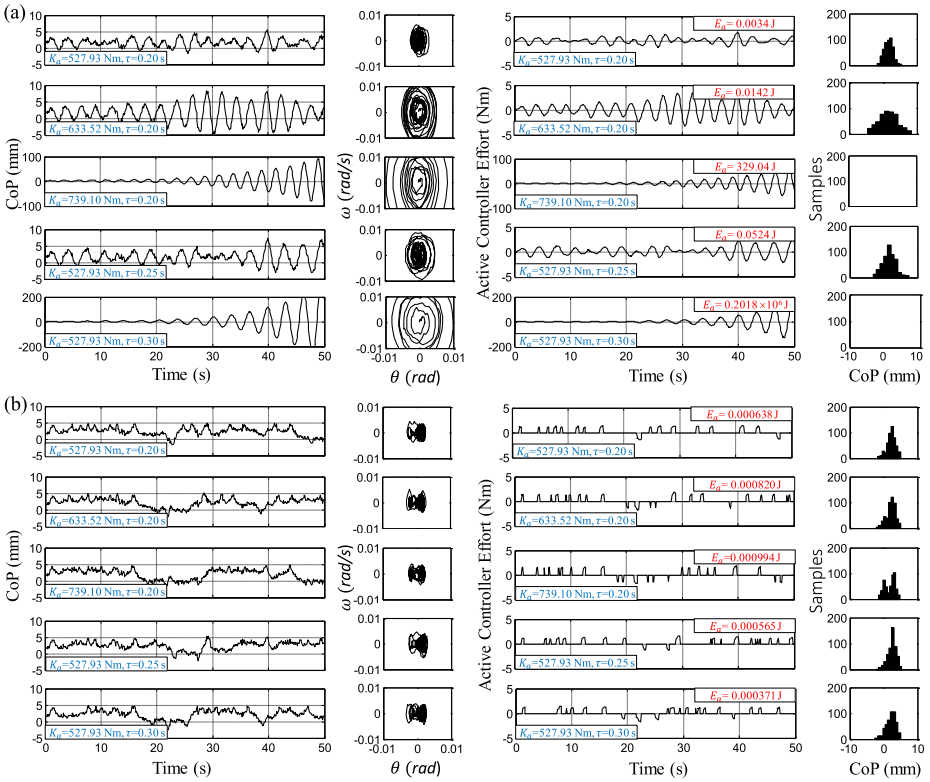

Fig. 2. Anterior-posterior CoP sways, phase portraits, active controller efforts and CoP histograms of the inverted pendulum model representing healthy individual ($K_P = 527.93$ Nm and $\tau = 0.2$ s) with varying active controller gain and delay with (a) continuous and (b) intermittent ($\theta_{th} = \pm 0.002$ rad) active controller.

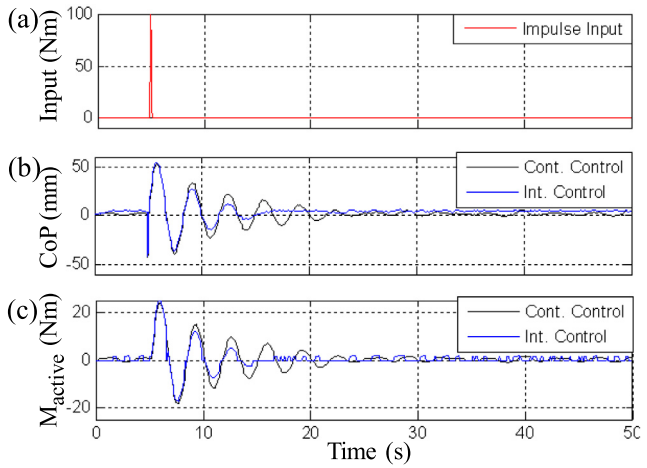

Fig. 3. (a) The impulse as an input torque, (b) CoP sway response and (c) the active controller effort of the inverted pendulum model representing healthy individual ($K_P = 527.93$ Nm and $\tau = 0.2$ s) for both continuous and intermittent ($\theta_{th} = \pm 0.002$ rad) active controller.

bit tonic muscle activity (Yamamoto et al., 2011; Tanabe et al., 2017).

4.1.5. Appearance of limit cycle oscillations (LCOs)

Limit cycles are self-sustained periodic oscillations that often arise from instability in non-linear time delayed systems. It is observed that the postural instabilities associated with mild PD results in LCOs (Chagdes et al., 2016a). Using wavelet analysis, Chagdes et al. (2016b) showed that ~44% of individuals with PD exhibits LCOs in their postural sway.

4.1.6. Trends in PSD slopes and Diffusion-Stabilogram parameters

Based on a random-walk approach, the Diffusion-Stabilogram directly relates the steady-state behaviour and underlying functional interaction of the neuromuscular mechanisms (Raymakers et al., 2005; Mitchell et al., 1995). This represents the squared distance between pairs of CoP's plotted against the corresponding time intervals. In Raymakers et al. (2005), it has been shown that individuals with PD exhibit lower critical time interval (~0.5 s), higher mean squared critical displacement (~70 mm^2) and lower diffusion constant (~15 mm^2/s) in contrast to healthy individuals. Further, PSD analysis is extensively used to discriminate between PD and healthy individuals. Matsuda et al. (2016), Yamamoto et al. (2011) suggest that the spectral power at a high-frequency band (1 – 5 Hz) and the low-frequency band (0.01 – 0.7 Hz) differ in case of PD patients from healthy individuals.

4.2. Model of PD Upright Posture

Before proceeding further, it is to be noted that the inconsistent observations in the CoP sway in PD may be due to several reasons. Parameters like time delay, muscle stiffness, medications, attention, sensory information may affect the postural stability in PD. In Kim et al. (2013), it is suggested that though sensory information affect the body orientation, there is no significant contribution from sensory information and attentional strategies while maintaining balance in PD. Hence these effects are not included in our analysis. PD patients exhibit higher stiffness at the ankle joint in comparison to healthy age-matched individuals (Park et al., 2015; Lauk et al., 1999; Chagdes et al., 2016a; Chagdes et al., 2016b; Rocchi et al., 2002). Through clinical studies (Zhang et al., 2009), it is also shown that the response time and therefore τ is larger (~ 0.3 – 0.4 s) in PD patients. To incorporate these impairments, the active controller gain and delay are set at 739.10 Nm

and 0.35 s respectively in our posture model for PD. The effect of medication on postural sway is not included in the current study. Further, we hypothesize that the increase in intermittency in active feedback control might be responsible for both higher and lower sway area. Therefore we vary the threshold limits of intermittent control in our PD posture model ($\theta_{\text{th}} = 0, \pm 0.001, \pm 0.002, \pm 0.003$ and ± 0.004 rad) based on the experimental observations in Fitzpatrick and McCloskey (1994). The resulting AP CoP sways, phase portraits, active controller efforts and CoP histograms are presented in Fig. 4. This controller effort may be considered equivalent to the active muscle activity in PD quiet stance. Other than controller gain, response time (time delay) and threshold values, the rest of the parameters of the model are kept same as of Section 2.

4.3. Comparative analysis of quiet stance in PD and healthy individuals

A comparative analysis is carried out between PD and healthy individuals by clustering them into three groups. Based on intermittent threshold limits, the simulations for PD population is divided into two groups PD-1 and PD-2. The threshold limits considered for healthy, PD-1 and PD-2 are ± 0.002 rad, ± 0.003 rad and ± 0.006 rad respectively. For each case, 100 Simulations are performed for 100 s time interval. The PSD analysis is performed by evaluating the Fast Fourier Transform (FFT) of steady state CoP data followed by a piecewise linear least square fit. The glossary of CoP sway measures are listed in Table 2 and the simulated results are summarized in Table 3.

Sway measures like mean CoP (m), CoP range (R), total distance traveled by the CoP (D) and parameters related to Diffusion-Stabilogram analysis are computed. V_m is the mean speed (or equivalently mean of absolute value of velocity). *SDT* is a measure of the time that a subject spend near the boundary of stability. The *CT* is a representation of the interval during which the posture is governed by an open loop control strategy. The *MSCD* corresponds to a measure of CoP shift that is tolerated before a restraining action is taken.

4.4. Experimental pilot study

A pilot study was conducted to provide further evidence to substantiate the results obtained from the above model. Three PD patients (mean age of 76 years with SD of 3 years, mean weight 65 kg with SD of 15 kg) and five healthy elderly age matched subjects (mean age of 74 years with SD of 8 years, mean weight of 75 kg with SD of 10 kg) were participated in the test trail. Three PD patients had Hoehn and Yahr scale scores 1.5, 1 and 2 respectively indicating mild motor impairment. Each subject was asked to stand barefoot quietly on a force platform (Wii Balance board) with their eyes open and focused to a fixed target 3 m away in front of their eyes and arms hanging along the sides of the body (Fig. 5(a)). The standard test procedure with precautions were followed similar to Bottaro et al. (2005), Yamamoto et al. (2011), Raymakers et al. (2005). Each subject performed three trials (with 2-3 min rest between trials) of quiet standing about 120s long and the CoP data from the first 10s to 60s of every trial was subjected to the subsequent analyses.

Table 2
Glossary of CoP sway measures (Raymakers et al., 2005; Yamamoto et al., 2011).

Symbol	Description	Units
m	Mean CoP	mm
V_m	Mean CoP Speed	cm/s
R	Maximum CoP Range	mm
D	Total distance travelled by the CoP	mm
S_l	Slope of logarithmic power spectral density function (PSD) of a CoP sway between 0.01 and 1 Hz	–
S_h	Slope of logarithmic power spectral density function (PSD) of a CoP sway between 1 and 5 Hz	–
PP	(Phase Plane Parameter) $\sqrt{(\sigma^2 x + \sigma^2 V)}$ where x and V are the displacement and speed in AP direction	–
SDT	(Sum of Maximal Deviation Time) $\sum_{i=1}^{16} t.i \times D_{max}$; where t is time interval unit (0.1 s), i is the number of time-units and D_{max} is the maximal deviation with a duration of $t \times i$ summed for $t = 0.1$ to 1.6 s.	mm.s
CT	(Critical time) Interval at which slope of the regression of the mean squared distance between random pairs on their time interval in a Diffusion-Stabilogram shows a significant break towards a shallow regression than initial one	s
$MSCD$	(Mean Squared Critical Displacement) Mean squared distance between random samples of CoP pairs in the Diffusion-Stabilogram with a time difference corresponding to the critical time interval	mm^2
D_s	(Short-term Diffusion coefficients) Slope of regression line through points before the critical time interval in the Diffusion-Stabilogram	mm^2/s
D_l	(Long-term Diffusion coefficients) Slope of regression line through points beyond the critical time interval in the Diffusion-Stabilogram	mm^2/s

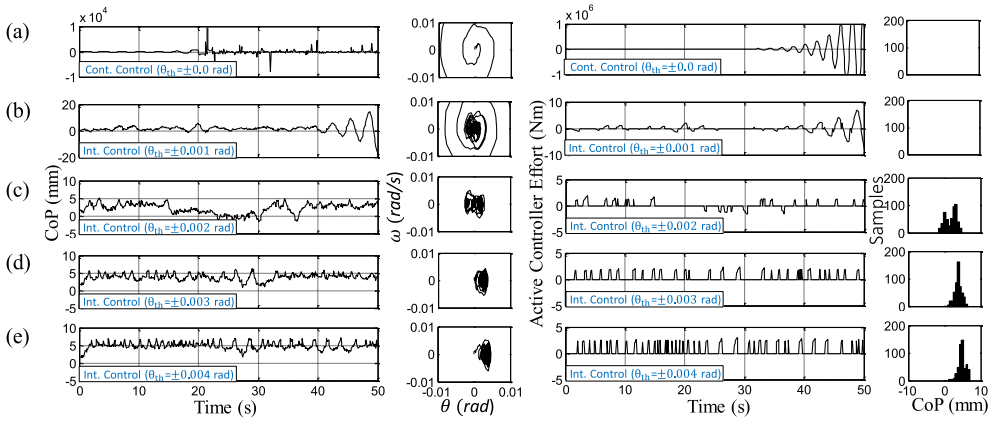

Fig. 4. Anterior-posterior CoP sways, phase portraits, active controller efforts and CoP histograms of the inverted pendulum model representing quiet standing of PD patients ($K_P = 739.10$ Nm and $\tau = 0.35$ s) with continuous and intermittent controller for varying threshold limits.

Table 3
Comparative analysis of various sway measures between PD and healthy individuals from the simulation study.

Parameters	Healthy		PD-1		PD-2	
	mean	SD	mean	SD	mean	SD
m	1.548	1.617	1.525	2.322	6.966	1.062
D	1160.8	18.456	1339.3	24.535	2193.0	43.124
R	8.181	0.504	9.915	0.371	7.269	0.619
V_m	0.0002	5.250	0.0006	6.304	0.0006	11.357
S_l	1.507	0.089	1.444	0.084	1.047	0.166
S_h	0.817	0.125	1.185	0.096	1.106	0.109
PP	5.494	0.078	6.719	0.116	11.409	0.188
SDT	79.425	9.492	84.616	11.415	69.805	7.918
CT	2.099	0.762	2.844	0.404	2.993	0.026
$MSCD$	0.376	0.334	0.574	0.411	1.529	0.368
Ds	1.563	0.448	0.955	0.392	0.013	0.0524
Dl	0.201	0.116	0.507	0.160	0.012	0.061
E_a	0.0018	0.0001	0.0052	0.0004	0.025	0.0008
E_f	81.614	6.045	186.748	9.166	832.952	25.476

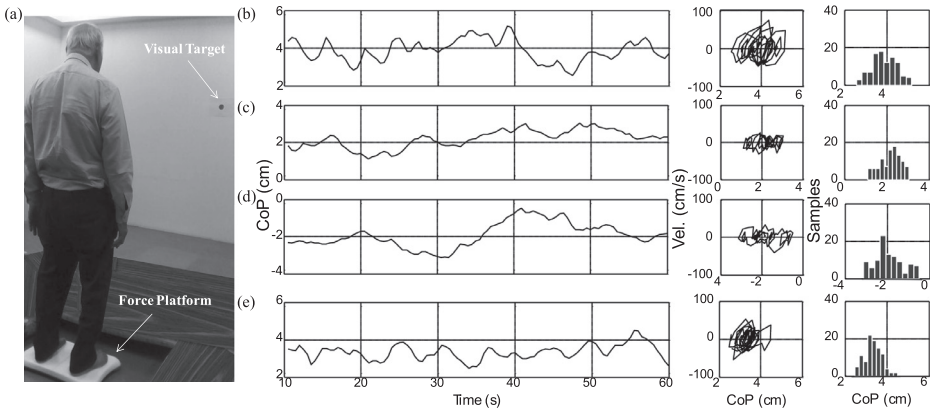

Fig. 5. (a) Experimental set-up for CoP measurement and exemplary plots for anterior-posterior CoP sways, phase portraits and CoP histograms representing quiet standing of (b) a healthy individual, (c) PD patient-1, (d) PD patient-2 and (e) PD patient-3 from the experimental pilot study.

5. Results and discussion

The analysis in Section 2 indicates that degradation in stability caused by higher gain and delay can be compensated by increase in intermittent control threshold. From the PD Simulation model, it is observed that, the upright posture is unstable for both continuous control and intermittent control with smaller thresholds (Fig. 4(a, b)). Increasing this threshold beyond a certain limit ($\theta_{\text{th}} > \pm 0.0012$ rad) results in bounded response (Fig. 4(c)). Here, three key observations can be made; an increase in intermittent control threshold is required to keep the response bounded, further, an increase in parkinsonian postural sway measures (for example D increases ~200 mm in PD-1 in Table 3) are seen in comparison to healthy individuals and finally, a behaviour similar to LCO appears in the sway. Note that, these observations are in line with the clinical observations discussed in Section 4.1.1 and 4.1.5. With further increase in the threshold limit, LCOs disappear and bimodality in stable CoP sway histogram is observed (Fig. 4(d)). When the threshold limit is increased even further, a reduction in sway range along with a shift in the mean CoP position (~5 mm) in PD-2 (seen in Fig. 4(d)) and Table 3) is observed as discussed in Section 4.1.2 and 4.1.3. It is to be noted that, this shift in mean CoP occurs in both anterior and posterior directions in our simulation as our approximation of human standing dynamics is symmetric. But in reality, asymmetries in the skeleto-muscular structure of human body may lead to asymmetric CoP shift. Again, in Fig. 4(e), the sway angle θ oscillates in vicinity of the threshold boundary and hence the duration of control activity is increased. This results in an increased energy expenditure which represents the higher muscle activity during quiet standing. The active controller effort plot in Fig. 4(e), and energy expenditure values in Table 3 suggest that the results are in close agreement with the clinical observation in Section 4.1.4.

The CoP waveform in Fig. 4(c) (equivalent to PD-1 in Section 4.3) shows a slow oscillatory trend whereas that of Fig. 4(e) (equivalent to PD-2 in Section 4.3) exhibits an almost flat trend across which the CoP oscillates with high frequency. From the comparison in Table 3, it is seen that the slowly fluctuating and less noisy CoP sway pattern in PD-1 results in large values of S_l and S_h, while the CoP sway pattern in PD-2 leads to small S_l and S_h. These observations are consistent with Yamamoto et al. (2011). It is observed that, individuals with PD exhibits higher mean CoP speed (V_m), mean squared critical displacement ($MSCD$) and phase plane parameter (PP) in comparison to healthy individuals which are also seen in Raymakers et al. (2005), Mitchell et al. (1995), Wright et al. (2007).

From the experimental data collected in the pilot study, few exemplary CoP sway patterns in AP directions with phase portraits and histograms are plotted in Fig. 5. Table 4 represents various CoP parameters evaluated as the average of all three trials for each PD patient which is compared against the average of all the trials of all 5 age matched healthy participants. All the test trails for two PD patients (with H/Y Score 1 and 1.5) show larger CoP sways and show ~10 mm higher CoP range than the average of healthy age

Table 4
Various sway measures evaluated from AP CoP data collected from Healthy and PD individuals in the experimental pilot study.

Parameters	Average of Healthy Subjects		PD Subject-1 (H & Y-1.5		PD Subject-2 (H & Y-1.0)		PD Subject-3 (H & Y-2.0)	
	mean	SD	mean	SD	mean	SD	mean	SD
D	423.39	107.03	402.28	55.9	458.95	47.98	616.90	51.52
R	28.317	5.57	32.71	3.68	51.64	14.94	19.87	1.78
V_m	0.48	0.18	0.966	0.45	2.59	0.38	0.177	0.07
S_l	1.273	0.49	1.17	0.166	1.47	0.167	1.046	0.155
S_h	2.209	0.53	2.10	0.34	2.028	0.576	2.28	0.24
PP	14.52	3.58	11.27	1.45	15.56	3.19	16.66	1.32
SDT	207.79	38.27	178.98	54.0	320.148	112.58	194.16	8.18
CT	1.96	0.479	2.1	0.1	2.33	0.115	1.8	0.3
$MSCD$	33.32	8.13	19.36	8.99	79.07	28.94	20.785	2.589
Ds	15.14	7.8	4.72	2.69	3.92	1.07	2.71	1.69
Dl	19.447	1.545	20.79	2.23	11.12	13.85	10.607	0.391

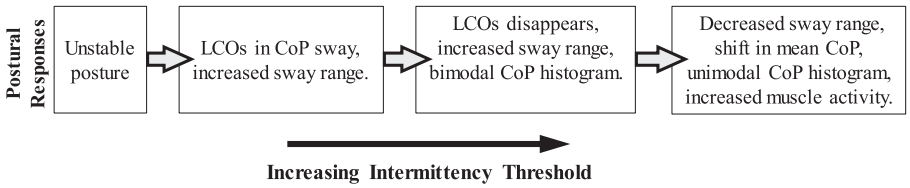

Fig. 6. Postural behaviour of PD patients with varying intermittency threshold.

matched individuals (Fig. 5(c) and (d)) whereas a smaller sway range (~8 mm) is observed in the other PD patient (with H/Y Score 2) in Fig. 5(e). Both in our simulations and experiments, we observe a similar increase in AP sway path length (D) similar to Blaszczyk et al. (2007). A slight variations in S_l and S_h in PSD for a group of PD patients with respect to healthy individuals is observed and is consistent with that of Matsuda et al. (2016), Yamamoto et al. (2011). As seen in Table 4, the CoP parameters obtained from diffusion stabiliogram analysis such as D_l, D_h, CT also follow similar trends as that of Raymakers et al. (2005) discussed in 4.1.6. Thus, the CoP patterns in AP direction and related parameters evaluated from both the simulation and experimental trails presented in this paper show close agreement with the clinical observations reported in various literature.

The pilot study helps support the simulation study in arguing that the notion of intermittent control is able to provide a conceptual underpinning for understanding the clinical observations observed in PD quiet stance. However, the above analysis also suggest that additional measurements of sway angle (measured via goniometer placed at ankle joint) and EMG activities of the leg muscles along with the CoP sway may help confirm these insights about varying threshold limits in postural control in PD. Fig. 6 represents the postural behaviours in PD patients emerging with increasing intermittency in control action. These insights may be helpful while performing clinical posturography for balance assessment. For example, if smaller CoP sway range is observed, the patient is also expected to show higher muscle activity with a shift in mean CoP position, which reflects the presence of higher intermittency in the control action.

6. Conclusion

Intermittent control is a control approach in which the observation is continuous, but the actions are intermittent based on a threshold criteria depending on the observed variable. While the presence of intermittent control has been observed in postural control in human quiet stance, we helped develop insights about intermittent control in terms of stability, energy efficiency and settling time. Furthermore, we discussed few key clinical observations related to postural sway in PD and demonstrated through a simulation study that these posturographic responses (including apparently contradictory observations) may be attributed to changes in threshold limits of the intermittent control action. Larger or smaller AP CoP sway magnitude can emerge from the same PD impairment and need not be viewed as contradictory. Further, clinical observations like the appearance of LCOs in CoP sway, stopped posture and tonic muscle activity might also appear as the consequences of varying controller intermittency. An experimental pilot study was also conducted to support these observations. The current study also suggests a follow-up in-depth study. with additional measurements of sway angle and EMG activities of the leg muscles along with the CoP to confirm these insights about varying threshold limits in postural control in PD.

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