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RATE-DEPENDENT GAIT DYNAMIC STABILITY ANALYSIS FOR MOTOR CONTROL ESTIMATION

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This work presents the gait dynamic stability study for different walking speeds adopted by user intentions. Experimental data were collected from four healthy subjects while walking on a force platform at slow, normal and fast speed. The rate-dependent variations in the center of pressure (COP) and ground reaction forces (GRF) were modelled as motor output and input responses. Finite difference and non-linear regression algorithms were implemented to model gait transitions. Dynamic stability estimation for level ground walking was performed by analysis in time and frequency domains. Study of the COP velocity in loading phase showed that, the overdamped motor output response acts as a compensator for instabilities and oscillations in unloading phase and initial contact. Normal walking was predicted, from gait analysis in frequency domain, as the most stable gait for healthy subjects.

1. Introduction

The dynamic stability of gait and posture depends on the center of pressure (COP) and its rate dependant variations [1]. The COP is a measure of neuromuscular motor response to achieve balance during transitions also defined as the path on the foot plantar surface, where all the external forces act [1]. While important advances have been achieved on gait static stability in the last decades, only a few research works have focused on the analysis of gait dynamic stability. Some of these works have proposed to use the rate of change in the center of pressure (COP) or center of mass (COM) acceleration 'a_{COM}', as neuromuscular motor responses to achieve dynamic stability. However, the correlation in COP and COM along with motor adaptation to gait transients at different walking speed, which is unknown, is investigated in this work.

Motor disorders, ageing and pathologies are major factors for gait impairments and fall risk [2]. Two third of our body weight lies at two third of body height and successful negotiation between body and environment reduces the chances of injuries due to fall risk. In relation to this, the ankle-foot, seen as the end-effector, plays a key role in the negotiation of balance during walking. Measurements from the COP have been used to distinguish gait abnormalities, establishing a rehabilitation index for evaluation of foot orthoses [3]. Under various pathological conditions, the rate of change in COP stands as impairment assessment tool clinically [1, 5]. The time derivative of COP describes the motor response to achieve stability and stands as an important measure [4].

Only a few studies predict gait transitions in relation to gait dynamic stability. That includes the first order negative exponential model widely applied for gait stability analysis in loading phase. The time constant and gains are estimated from time domain a_{COM} response [2]. In a study, the sample entropy algorithm has been applied to predict COP, GRF rate dependant variations in loading phase [6]. Another study applied Savitzky Golay filter to compute COP velocity and analysed dynamic stabilities for four-foot conditions [5]. In this work, we extend our study of dynamic stability in [8], including gait unloading phase and vibrations effect. The stability correlation is also investigated towards gait transitions at different walking speeds. Furthermore, motor transient responses are modeled to estimate instability thresholds both in time and frequency domains.

2. Methods

The experimental protocol consisted of an array of 12 cameras connected to Qualisys motion capture system with AMTI force platforms (BP 400600-2000) shown in Fig.1. The operating frequencies were 400fps and 1200Hz respectively. The force plate data was captured using 64 channel analogue output board (PCI-DAC6703) and synchronized with motion markers using Qualysis software (QTM) and Oqus cameras. The experiments were performed with four healthy subjects (ages 20±1; foot length 27±1cm; weight 72±7kg, heights 175±7.5cm). Four trials were conducted for each subject over 8m long walking track at user self-selected level ground walking speeds i.e. slow, normal, fast. The experiments were performed with the consent of ethical approval from the University of Leeds.

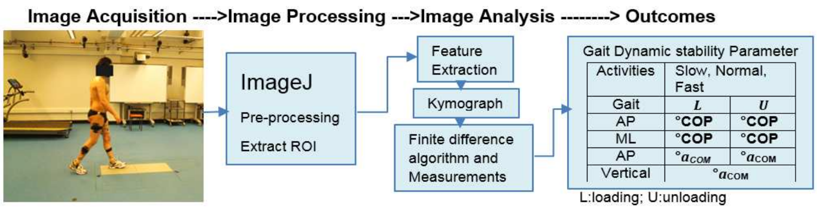

Figure 1 Experimental protocol for gait dynamic stability analysis in anterior-posterior (AP), medial-lateral (ML) and vertical directions.

The motion data captured from each subject was exported as AVI file at 400fps. The rate dependant variations were measured using open source image processing software ImageJ (http://imagej.nih.gov/ij/). After selecting desired region of interest and pre-processing steps, the GRF vector edges were detected from kymograph plot. Kymograph is an image analysis technique for single plane time-displacement motion capture. Here, this algorithm was applied for macroscale motion analysis as shown in Fig.2. The finite difference algorithm (Table 1) was applied to COP and GRF vector paths extracted by kymograph. The rate of change in GRF was normalized with body mass to obtain vibrations signal i.e. $a_{C\dot{O}M}(= GRF/mass)$ for respective subjects.

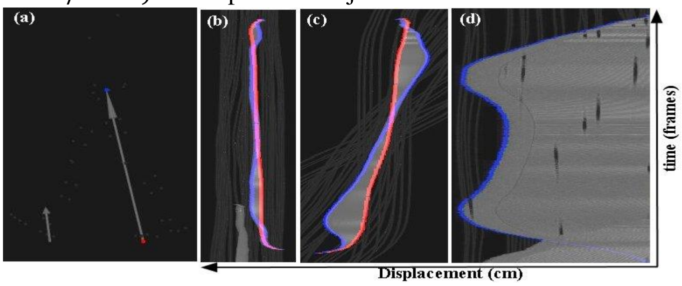

Figure 2 (a) GRF vector position and magnitude in stance phase (blue – GRF position, red – COP position), Kymograph plot in (b) AP, (c) ML, (d) vertical directions.

Table1 Finite difference algorithm.

Parameter	Actuate Value	Average Value
Rate of change in COP	$V_{COP} = \frac{d_{xi}}{d_{ti}} = \frac{\lvert X_{i+1} - X_i \rvert}{\lvert t_{i+1} - t_i \rvert}$	$V_{COP} = \frac{d_{xi} + d_{x\_sum}}{d_{ti} + d_{t\_sum}}$
Rate of change in GRF	$'GRF = \frac{d_{zj}}{d_{tj}} = \frac{\lvert Z_{j+1} - Z_j \rvert}{\lvert t_{j+1} - t_j \rvert}$	$'GRF = \frac{d_{zj} + d_{z\_sum}}{d_{tj} + d_{t\_sum}}$

Initial conditions $d_{x\_sum}=0; d_{z\_sum}=0$.

Finite difference algorithm was implemented to determine average rate of change. The averaging method can smooth the noise, however, the signal diminishes at gait unloading phase. That filtration allege is overwhelmed by windowing concepts. After due analysis, window sizes of 100 frames and 50 frames showed to be the optimal for loading and unloading phases respectively. Exception was made for vertical $a_{C\dot{O}M}$ signal as it takes whole stance to get stable as shown in Fig.4 (b). The modelling assumptions included: 1) the analysis made in force plate coordinate system i.e. X-axis for AP, Y-axis for ML and Z-axis stands for vertical, 2) In finite difference algorithm, the initial values were assumed in range to mean for all 16 trials.

3. Results

The posturographic test was conducted over 16 times to model dynamic stability. Motor I/O signals i.e. $a_{C\dot{O}M}$ and COP velocity presented a non-normal distribution. The Spearman's correlation was applied using SPSS (IBM SPSS Statistics 22) for 16 trials in each case. The inter-trial average correlation was found between 0.9-1. The dynamic models were estimated using MATLAB curve fitting tool. The non-linear least square regression (least absolute residual) method was applied to estimate motor output response and sum of sinusoids used to model vibrations input signal. The gait transient responses were modelled maximizing the coefficient of determinant (R^2) and 95% of confidence bounds as shown in Fig.3 and Fig. 4. Our modelling approach achieved better accuracy than parameter estimation for defined inverted pendulum or negative exponential models [2]. The gait cycle was categorized in loading and unloading phases, both in AP and ML directions for COP signal, and, AP and vertical directions for a_{COM} signal. The one way ANOVA test performed to check inter-trial walking speed variance (p=.68 for slow, p=.93 for normal, and p=0.98 for fast), implies that the speed variation is insignificant (p>0.05). Hence, average time of trials was used for respective windows. Eq. (1) represents motor input signal as transient and steady state components. The average R^2 lies in 99.7±0.06% for loading and 92.3±6.2% for unloading phases including AP and ML for the 16 trials.

$$C\dot{O}P = ae^{bt} + ce^{dt} \tag{1}$$

where a and c are gains ($K = a + c$) and b, d (±) represents reciprocal of time constants ($\tau_{net} = b \times d/b + d$).

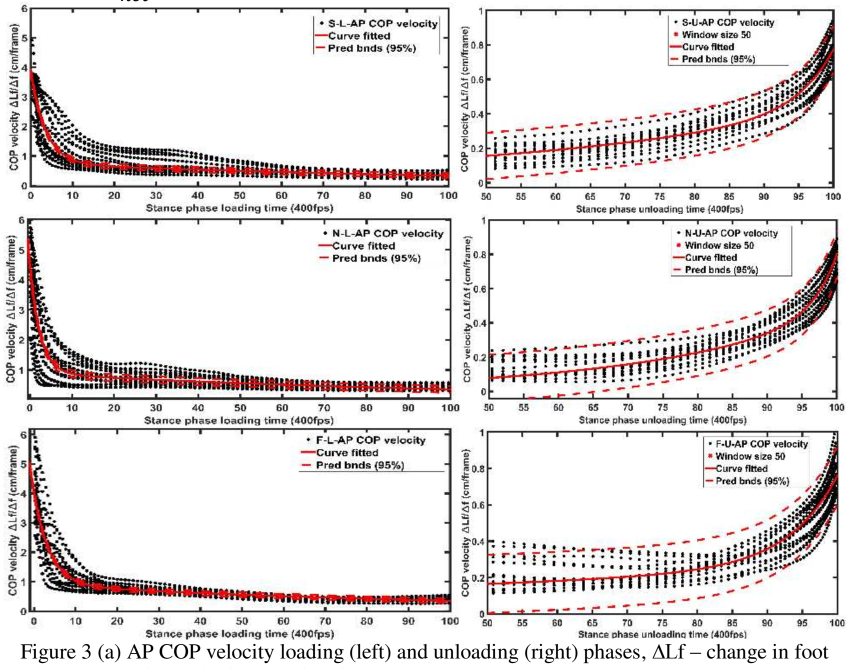

Figure 3 (a) AP COP velocity loading (left) and unloading (right) phases, ΔLf – change in foot length, ΔWf – change in foot width, Δf – change in frames, S-slow, N-normal, F-Fast, L-loading, U-unloading, AP-anterior-posterior, ML-medial-lateral

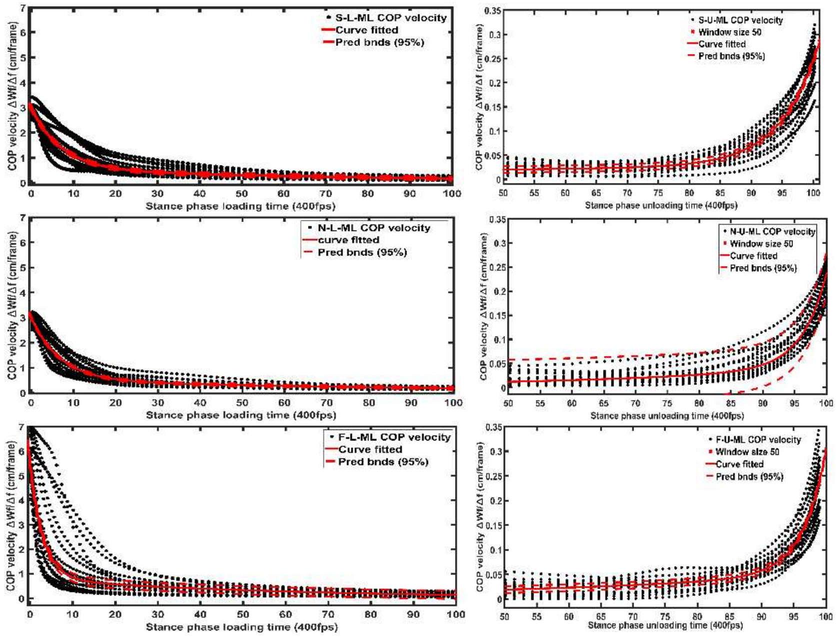

Figure 3 (b) ML COP velocity loading (left) and unloading (right) phases

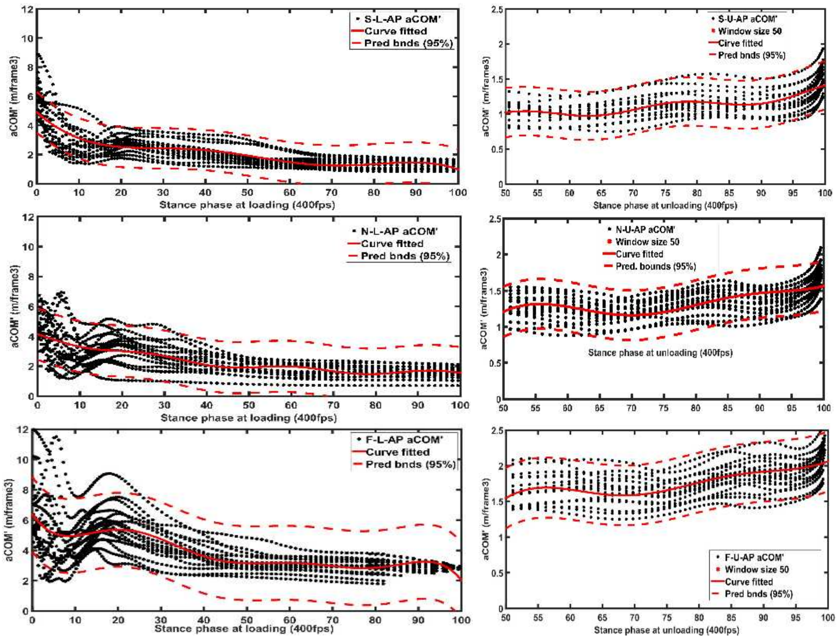

Figure 4 (a) Rate of change in a_{COM} in AP direction loading (left) and unloading (right) phases.

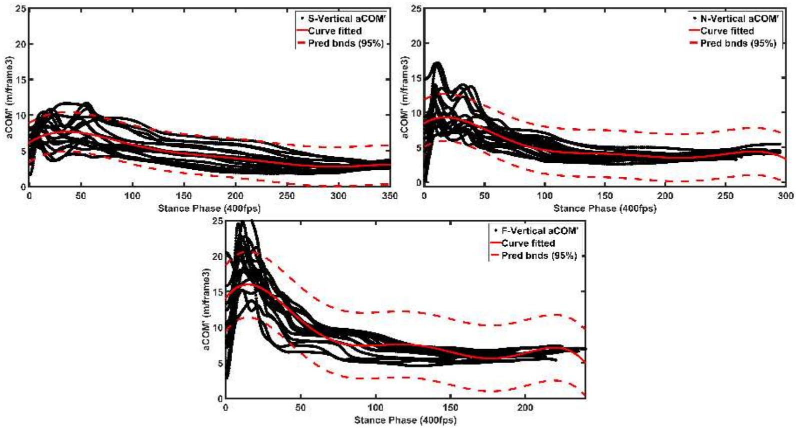

Figure 4(b) Rate of change in a_{COM} in vertical direction at slow (S), normal (N) and fast (F) speed.

The $a_{\dot{COM}}$ input signal is modelled in Eq. (2). The R^2 values lies 52.6±12.4%, 33.1±3.1%, and 67±5.8% for AP loading, unloading, and vertical direction.

$$a_{\dot{COM}} = a_1 \sin(b_1 t + c_1) + a_2 \sin(b_2 t + c_2) + a_3 \sin(b_3 t + c_3) \qquad (2)$$

where $a's$ are amplitudes, $b's$ are frequency of oscillation and $c's$ are phase shifts. The time domain stability index (I), previously defined in [2], has been extended in this work as shown in Eq. (3), by correlating loading (L) and unloading (U) during double limb stance support.

$$I = \tau_L \times K_L / \tau_U \times K_U \qquad (3)$$

Table 2 Gait stability thresholds in time domain and frequency domain.

Exponential decay model $\dot{COP}$					**Sinusoids model $a_{\dot{COM}}$**		
Motor Output*	**Time Domain**		**Frequency Domain**		**Motor Input***	**Frequency Domain**	
Speed-phase-direction	Time constant τ (frames)	Net Gain K	Gain Margin M (dB)	Phase Margin Φ (deg)	**Speed-phase-direction**	Gain Margin M (dB)	Phase Margin Φ(deg)
S-L-AP	3.44±.04	3.85±.020	∞	8.218	**S-L-AP**	1.2×10^{-7}	0
N-L-AP	2.42±.03	4.47±.025	∞	11.142	**N-L-AP**	6.2×10^{-7}	0
F-L-AP	3.69±.03	4.02±.019	∞	7.722	**F-L-AP**	8.4×10^{-9}	0
S-L-ML	6.70±.08	2.96±.023	∞	4.963	**S-U-AP**	6.7×10^{-7}	0
N-L-ML	6.40±.05	2.98±.017	∞	5.174	**N-U-AP**	1.8×10^{-6}	0
F-L-ML	2.93±.04	5.46±.044	∞	8.346	**F-U-AP**	9.4×10^{-6}	0
S-U-AP	4.95±1	.058±.014	∞	-47.68	**S-Vertical**	∞	0
N-U-AP	7.55±0.7	.087±.007	∞	-25.54	**N-Vertical**	∞	0
F-U-AP	6.80±1.5	.104±.042	∞	-26.62	**F-Vertical**	∞	0
S-U-ML	5.89±.14	.015±.002	∞	-67.72	-	-	-
N-U-ML	3.51±.04	.003±.003	∞	-142	-	-	-
F-U-ML	3.35±.06	.007±.001	∞	∞	-	-	-

*S-slow, N-normal, F-fast, L-loading, U-unloading, AP-anterior-posterior, ML-medial lateral, I-input, O-output.

4. Discussion

The dynamic stability was modelled considering $a_{C\dot{O}M}$ and $C\dot{O}P$ as motor I/O signals. The transient response stability thresholds were mentioned in Table 2 analysed here in time and frequency domains.

'$C\dot{O}P$' *motor output response* – The $C\dot{O}P$ models showed overdamped motor response in loading. The time constant (τ) increased for AP in slow and fast speeds, while it decreased at normal speed. The unloading phase presented opposite trends for slow, normal and fast speeds. The gains at loading achieved less variance (0.10 in AP, 2.0 in ML) w.r.t walking speed, while gains for ML in unloading were negligible. The frequency domain pole-zero map in Fig.5 (a) shows loading as the only stable phase and compensates unloading phase instability as well as undamped vibrations during double limb stance phase. The stability index (I) was computed using Eq. (3). For slow, normal and fast speeds, the mean values of stability index were 226±16, 2595±2052, 696±27 and 48±1; 16.5±0.33, 23±5 for ML and AP directions. The larger intra class variance (280) showed that the walking speed had significant effect over stability. The higher ML index indicated least instability. The AP indices were relatively in range. At slow speed the index (I) predicted the least instability in AP. To prove the index concept, the analysis was performed in frequency domain using Bode plot, assuming each model as single I/O open loop linear-time invariant system, the results are shown in Fig.5 (b).

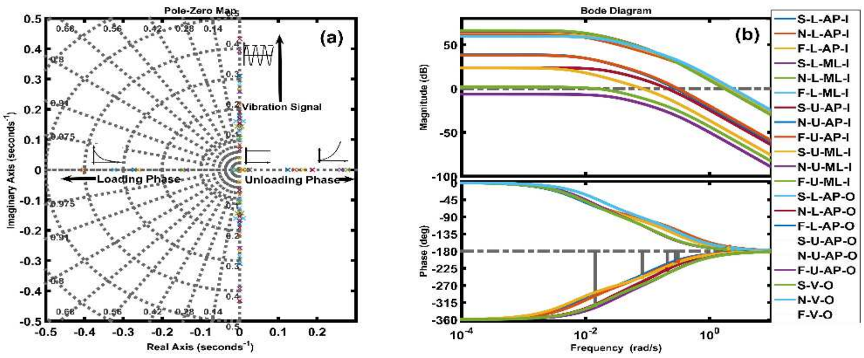

Figure 5 (a) Pole-zero plot in frequency domain, (b) Bode Plot for gait loading and unloading phases

Gain and phase margins were infinite and (9±1.6)° respectively during AP loading phase. The loading phase proved to be more stable, however, the normal speed stands for least instability consistent with [7] and inconsistent with time domain analysis. Less unloading phase margin (-25.54°) also showed least instability at normal speed. Considering sagittal motion the most significant, the fast walking speed showed the least stability margins both in AP and ML directions.

'$a_{C\dot{O}M}$' *motor input response* – The $a_{C\dot{O}M}$ models predicted undamped oscillatory response. The variability in oscillation amplitude (m/frame^3) was 55±52, 11.8±14 for AP loading, unloading, and 61.5±41 in vertical plane. The sagittal plane $a_{C\dot{O}M}$ showed maximum amplitude with reducing cycle time as speed increased as shown in Fig.4 (b). The frequency domain analysis showed that $a_{C\dot{O}M}$ in sagittal plane was the maximum disturbance to be compensated by the motor. The rhythmic and long spread vibrations caused less instability as observed at slow speed in Fig.4 (b). This study predicted the frequency domain motor I/O responses, extending previous studies in time domain [2, 5, 6]. Overall, this research work on the analysis of dynamic stability provides a methodology for ankle-foot joint stability thresholds assessment and adopting wearable soft orthotics design e.g. smart actuator response in relation to modeled I/O signals and vibrations damping characteristics. Some aspects of research carried out here can be transferred to applications like bipedal robot where stability and fall prevention are very crucial for a successful locomotion strategy.

5. Conclusion

The goal of this study was to investigate gait stability at transition phases. During loading phase, the frequency domain analysis showed the most stable results at normal walk for healthy subjects. The $C\dot{O}P$ signal generated by the motor in loading phase was maximally stable/overdamped response. The $C\dot{O}P$ in unloading was unstable and compensated by opposite limb during loading. Furthermore, the motor responded to both instabilities i.e. $C\dot{O}P$ and $a_{C\dot{O}M}$ in parallel control mode. The outcome of this research work can be used for the design and control of biped assistive or rehabilitative robotics. The future work will include motor behavioral control estimation to adopt different terrain conditions using proposed I/O signals.

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