FMCH: a new model for human-like postural control in walking

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***Abstract*— Spring loaded inverted pendulum (SLIP) model used simple spring mass mechanism to explain leg function and ground reaction force in legged locomotion. Balancing the upper body can be addressed by addition of a rigid trunk to this template model. The resulting model is not conservative and needs hip torque to keep the trunk upright during locomotion, like humans. Leg force modulated compliant hip (FMCH) is our new model for postural control in walking which employs the leg force feedback to adjust the hip compliance. Such an application of positive force feedback presents a new template for neuromuscular model. This method provides stable and robust walking in simulations and also mimics human-like kinetic behavior. Analyzing human walking experiment shows that FMCH can explain the hip torque-angle relation for different walking speeds. Finally, this approach may physically implement the virtual pendulum (VP) concept, observed in human/animal locomotion.**

I. INTRODUCTION

Simple, conceptual models are very useful tools in describing and analyzing human/animal locomotion. Such models which are called "templates" [1] benefit from high level of abstraction in explaining locomotion features. Additionally, many successful legged robots are developed [2][3] based on template models. They are also utilized as explicit templates for control [4]. Spring-loaded inverted pendulum (SLIP) is one of the most popular templates [5][6]. In SLIP, whole body mass is concentrated in one point (center of mass (CoM)) and the leg behavior is modeled by a massless spring. This model and its extension to have the second spring during double support (BSLIP for Bipedal SLIP) can describe human gaits, such as hopping/running [5] and walking [7], respectively.

In spite of all advantages of (B)SLIP model, since the upper body is represented by a point mass, it cannot address postural control whereas vertical body alignment plays a key role in stabilization of human locomotion [8]. For that purpose, the SLIP must be extended to include a model of the upper body. An extension of the SLIP with a rigid trunk was introduced as TSLIP (for Trunk-SLIP) [9] or ASLIP, for "Asymmetric SLIP" [10][1]. The model that we use in this paper is based on BTSLIP (Bipedal TSLIP), shown in Fig. 2.

In contrary to most of posture control approaches which are based on control of the trunk orientation with respect to an absolute referential frame [2][4][11], Maus et al. [12] proposed a postural controller which uses the angle between leg and trunk. This controller was based on an innovative concept for posture stabilization, coined Virtual Pendulum (VP), based on observations in a variety of animals including humans [8]. One passive alternative for balance control using the same angle for postural control is having rotational springs at hip joint. Stabilizing the gait and implementing the VP concept were already accomplished in walking [13], running and hopping [14] with a passive hip spring. Looking at humans hip torque-angle behavior shows that linear spring cannot explain human-like postural control in walking. In addition, nonzero force at moment of touchdown which results in discrete actuation makes the control approach impractical.

From another point of view, humans neuromuscular system can be implemented in mechanical models considering different relations between the generated force, muscle length and velocity (e.g., Hill-type muscle model [15]) besides some sensory feedback signals (as muscle reflexes) which control the actuator parameters like model presented in [16]. In [17], stable hopping was achieved using positive force feedback to stimulate the muscle. Later, Günther et al., introduced a new muscle model in which damping effect of the muscle is tuned based on muscle force [18]. Using spring (damper) mechanism and tuning the parameters may transfer part of the knowledge of neuromuscular models to the templates. Inspired from the muscle models and reflex system, we propose FMCH (force modulated compliant hip) for postural control. In this model the hip stiffness is adjusted based on the leg force feedback signal. With FMCH we (i) achieve stable walking with upright trunk needless to measure absolute leg (or body) angle with respect to ground (ii) present an acceptable explanation of human postural control (mimicking human hip torque pattern (iii) suggest a mechanical representation of postural control method based on template models (iv) introduce a new concept in muscle reflex system which can be used to realize human locomotion.

II. METHODS

A. Simulation model

The simulation model which is used in this study is based on BTSLIP model, shown in Fig. 2. In BTSLIP model, legs are modeled by massless springs and a rigid trunk represents the upper body with mass m and moment of inertia J. Walking dynamics (gait cycle) has two phases: *single support (SS)* and *double support (DS)*.

SS starts at takeoff moment of a leg and ends at touchdown of the same leg. Touchdown (TD) is defined as the moment

This research was supported in part by the EU project BALANCE under Grant Agreement No. 601003.

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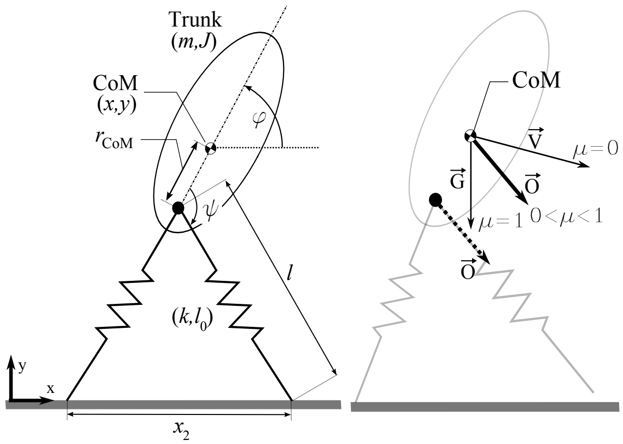
[1]Here we will use TSLIP because Asymmetric SLIP can also apply to a SLIP model with asymmetric leg properties.



(a) (b)

Fig. 1: (a) TSLIP model with a rigid trunk and a leg modeled as a massless prismatic spring. (b) Velocity-based leg adjustment (VBLA) during flight phase.

that the distal end of the leg hits the ground and takeoff is when the leg leaves the ground. In SS, one leg is in contact with the ground, called stance leg and the swing leg moves virtually (no change in dynamics when the leg is massless) to finish the SS with hitting the ground with desired angle (angle of attack). Here, the hip torque exerted between trunk and stance leg and the swing leg angle are the two control parameters.

TO is detected when the ground reaction force ($GRF = [GRF_x \ GRF_y]$) has no vertical component ($GRF_y = 0$). In this phase, $F_s = K_s \ (l_0 - l)$ gives the spring force along the leg axis, where l, l_0 and K_s are respectively the current leg length, leg rest length and the spring stiffness. Defining the states x, y and φ as the CoM horizontal and vertical positions and the trunk orientation, respectively; the hip point ($X_h = [x_h, \ y_h]$) which is positioned below CoM with distance r_h is obtained as follows

$$\begin{aligned} x_h &= x - r_h \cos\varphi \\ y_h &= y - r_h \sin\varphi \end{aligned} \tag{1}$$

The hip torque τ is determined by the controller (compliant hip) for stabilizing the posture of the trunk. The hip torque and the leg spring force produce the ground reaction force in interaction with the ground by

$$\begin{aligned} GRF_x &= F_s \tfrac{x_h}{l} + \tfrac{\tau y_h}{l^2} \\ GRF_y &= F_s \tfrac{y_h}{l} - \tfrac{\tau x_h}{l^2} \end{aligned} \tag{2}$$

Considering g as the gravity acceleration, the motion dynamic in the SS is described by

$$\begin{cases} m\ddot{x} &= GRF_x \\ m\ddot{y} &= GRF_y - g \\ J\ddot{\varphi} &= \tau + r_h(GRF_x \sin\varphi - GRF_y \cos\varphi) \end{cases} \tag{3}$$

When the swing leg in SS hits the ground the second stance leg appears (hereafter we show the parameters related to this leg by subindex $_2$), meaning DS starts and it ends with TO of first stance leg leg (shown by $_1$). In this phase, the controller produces torques (τ_1 and τ_2) between legs and trunk to keep the system stable. Defining the position of the second stance leg by $[x_2, \ 0]$, the dynamic model of DS will be as follows:

$$\begin{cases} m\ddot{x} &= GRF_{x1} + GRF_{x2} \\ m\ddot{y} &= GRF_{y1} + GRF_{y1} - g \\ J\ddot{\varphi} &= \tau_1 + \tau_2 + r_h(GRF_{x1} + GRF_{x2}) \sin\varphi \\ & \quad - r_h(GRF_{y1} + GRF_{y2}) \cos\varphi \end{cases} \tag{4}$$

where

$$\begin{cases} GRF_{x1} &= F_{s1} \tfrac{x_h}{l} + \tfrac{\tau_1 y_h}{l_1^2} \\ GRF_{y1} &= F_{s1} \tfrac{y_h}{l} - \tfrac{\tau_1 x_h}{l_1^2} \\ GRF_{x2} &= F_{s2} \tfrac{x_h - x_2}{l} + \tfrac{\tau_2 y_h}{l_2^2} \\ GRF_{y2} &= F_{s2} \tfrac{y_h}{l} - \tfrac{\tau_2 x_h}{l_2^2}. \end{cases} \tag{5}$$

B. Control approaches

In single support phase of walking with the BTSLIP model, the controller is combined of leg adjustment for the swing leg and hip torque control between stance leg and trunk (τ). The double support does not have freely swing leg movement and two hip torques τ_1 and τ_2 should be produced by the motion controller. VBLA (Velocity based leg adjustment) is our control strategy for swing leg and FMCH is the approach for hip torque control.

1) Leg adjustment during the swing phase: The easiest leg adjustment approach is setting the leg angle to a fixed value. Although using a fixed angle of attack with respect to the ground can stabilize running [19] and walking [7], the region of attraction for the stable gait is quite small. This drawback which equals to low robustness and high sensitivity to gait speed changes and control parameters exist in other common leg adjustment methods (mostly based on Raibert approach [2]). In order to concentrate on balancing of the trunk, we need to have a robust leg adjustment method. In most of the leg adjustment strategies, the foot landing position is adjusted based on the horizontal velocity [4] [20]. In this paper, VBLA (Velocity Based Leg Adjustment) presented in [21], is used as a robust method. This method can mimic human leg adjustment strategies for perturbed hopping [22] and achieve a large range of running velocities by a fixed controller [23]. Here, we use this method for walking.

In VBLA, the leg direction is given by vector $\vec{O}$ as a weighted average of the CoM velocity vector $\vec{V}$ and the gravity vector $\vec{G} = [0, -g]^T$ (Fig. 1b).

$$\vec{O} \quad = \quad (1-\mu)\vec{V} + \mu\vec{G} \tag{6}$$

where weighting constant μ accepts values between 0 and 1.

2) FMCH for hip torque control: We consider a bi-directional rotational spring between trunk and each leg. With the configuration showed in Fig. 2(a) for double support phase, the hip torques of leg i is determined by

$$\tau_i = k_i(\psi_i - \psi_i^0) \tag{7}$$

in which k_i and ψ_i^0 are the hip stiffness and rest angle for leg i, respectively, and ψ_i is the angle between trunk and leg

i as shown in Fig. 2(a). In FMCH control approach we use the leg force for modulating hip stiffness.

$$k_i = k_i^0 \frac{F_s^i}{F_s^n}, \quad i = 1,2 \tag{8}$$

where k_i^0, F_s^i ans F_s^n are the default values for hip spring stiffness, leg force and normalization value for leg force, respectively. In [24], we showed that for a single leg in contact with ground (with length l), if k_i^0 is computed by the following equation and $\psi_i^0 = 0$, then a the GRF goes through a point on trunk axis whose distance to hip is equal to r.

$$k_i^0 = \frac{lr}{(l+r)} F_s^n \tag{9}$$

Having an intersection point for GRFs during whole gait cycle, placed above CoM, is found in human walking, called VPP (virtual pivot point) [8]. For the TSLIP model shown in Fig. 2(b), the required torque to redirect the GRF toward VPP, is

$$\tau_{VPP} = F_s \; l \; \frac{r_h \; sin\psi + r_{\text{VPP}} \; sin(\psi - \gamma)}{l + r_h \; cos\psi + r_{\text{VPP}} \; cos(\psi - \gamma)} \tag{10}$$

in which r_{VPP} and γ are the VPP distance to CoM and deviation angle from trunk axis, respectively, as shown in Fig. 2(b). In Appendix. A, it is shown that FMCH can approximate VPP out of trunk axis if the hip spring rest angle is computed as follows

$$\psi_0 = \frac{r_{\text{VPP}} \gamma}{r}. \tag{11}$$

Therefore, if the gain for adapting hip stiffness is adaptively adjusted based on leg length (see Eq. 8), leg force feedback can be employed to precisely control VPP. Since the stance leg length changes are minor in walking, l can be replaced by its average value $\bar{l}$. Therefore, from Eqs (7) to (9), based on the following equation, FMCH controller only needs to measure the leg force to adjust hip stiffness

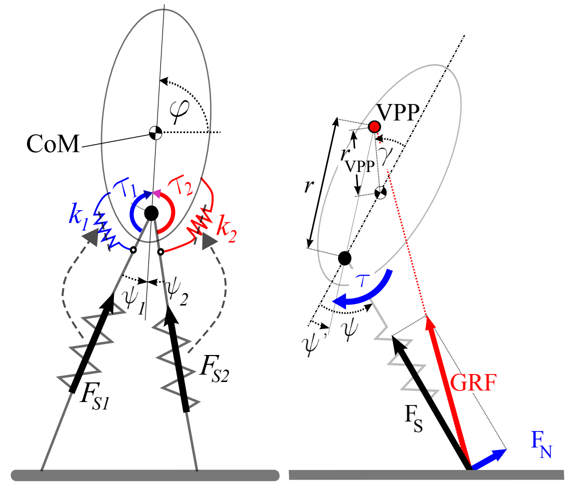
$$\tau_i = c F_s^i (\psi_i - \psi_i^0) \tag{12}$$

and it can also properly approximate VPP if the constant gain (c) is computed as follows

$$c = \frac{\bar{l} r}{(\bar{l} + r)} \tag{13}$$

C. Walking experiment

We investigated the ability of FMCH in replicating human virtual hip torque in walking. The virtual hip torque is the torque between upper body and the virtual leg (the line between hip and COP (center of pressure)). Since the single support is the major part of walking cycle (about 80%) [25], we look at this phase of walking in the experimental data. Another reason is that with just one leg in contact with the ground the error can be characterized better because there is just one controller for balancing. For such analysis, we compute the ratio between hip torque and leg force



(a) (b)

Fig. 2: (a) FMCH for double support. (b) Virtual pendulum-based posture control (VPPC) during stance phase.

($r_F^\tau = \frac{\tau_h}{F_S}$) and draw r_F^τ versus hip angle ψ (between stance leg and trunk). The more linear behavior, the more fitting with FMCH concept.

The data was collected in walking experiments on a treadmill (type ADAL-WR, Hef Tecmachine, Andrezieux Boutheon, France) at different speeds. Motion capture data (Qualisys, Gothenburg, Sweden) from 11 markers and ground reaction force data (12 piezo-electric force transducers within the treadmill) were collected. Twenty one subjects (11 female, 10 male) were asked to walk at different percentages of their preferred transition speeds (PTS)[2]. The treadmill speed which equals the average velocity during strides was employed as the walking speed. The subjects were between 22 to 28 years old with $1.73 \pm 0.09 m$ height and $70.9 \pm 11.7 kg$ weight.

D. FMCH for finding VPP location

As mentioned before, VPP is a concept which was observed in human/animal upper body balancing. For every control approach, existence of the VP concept can be investigated. VPP is defined [8] as "the single point at which the total transferred angular momentum remains constant and the sum-of-squares difference to the original angular momentum over time is minimal, if the GRF is applied at exactly this point". In [24] the mathematical details to find VPP based on this definition was presented.

Based on FMCH concept, we propose a new method to compute VPP from experimental data. The first step is fitting a line to $r_F^\tau - \psi$ curve (e.g., with least square approach). Then, the rest angle ψ^0 and coefficient c in Eq. (12) are found. Using the average length of the virtual leg $\bar{l}$, Eqs. (11) and (13), r, γ and finally r_{VPP} are calculated. For more details see Appendix. A.

[2] PTS is the preferred speed for transition between running and walking which is typically about $1.9 - 2.1 \; m/s$ for humans [25] .

III. RESULTS

In this section, first, the results of stable walking using VBLA for leg adjustment and FMCH for posture control are shown in simulation model. Then the experimental data analyzed based on FMCH for hip torque control. Explaining the human balance control and also examining the accuracy of FMCH in finding VPP are shown.

A. Simulation results

BTSLIP for walking, explained in Sec. II-A, is simulated in MATLAB/SIMULINK 2013b using ode45 solver. The system initiates in single support when the stance leg is vertical (mid-stance). At this moment the stance leg is compressed and leg length (l_{in}) is less than the spring rest length. The model parameters are set to match the characteristics of a human with 80 kg weight and 1.89 m height (Table I). For different walking speeds $0.5-1.3[m/s]$, different combinations of VBLA coefficient ($0.2 \leq \mu \leq 0.4$), leg spring stiffness ($10 \leq k_N \leq 40$), hip stiffness ($0.1 \leq c \leq 0.5$) and rest angle ($0 \leq \psi \leq 0.1$) result in stable motion[3].

TABLE I: Model parameters

Parameter	symbol	value [units]
trunk mass	m	80 [kg]
trunk moment of inertia	J	4.6 [kg m^2]
distance hip-CoM	r_{CoM}	0.1 [m]
Normalized leg stiffness	k	40
leg rest length	l_0	1 [m]
Gravitational acceleration	g	9.81 [m/s^2]

Fig. 3 shows the hip torque of each leg (τ_1 and τ_2) and the total torque ($\tau = \tau_1 + \tau_2$) for a sample set of control parameters and initial conditions (see Table .II). The results are shown for one gait cycle starting with double support (touchdown) and ending with single support (before the next touchdown). The hip torque pattern of each leg are similar to what found in human walking [8]. Similar patterns for different leg in opposite directions with a phase shift results in smaller net torque on trunk. The main difference is having double support longer than what is observed in human walking at this speed. The torque between each leg and trunk is smooth (with no jump like in passive hip spring [13]), resulting from modulating hip stiffness by leg force.

TABLE II: Initial conditions and control parameters

Parameter	symbol	value [units]
speed	V_0	1 [m/s]
initial leg length	l_{in}	0.99 [m]
normalized hip spring stiffness	c	0.26
hip spring rest length	ψ_0	0[°]
VBLA coefficient	μ	0.34 [m]
Normalized leg stiffness	K_N	40

[3]With SLIP-based models with fixed leg spring stiffness, fast walking (speed more than $1.3[m/s]$) is not achievable

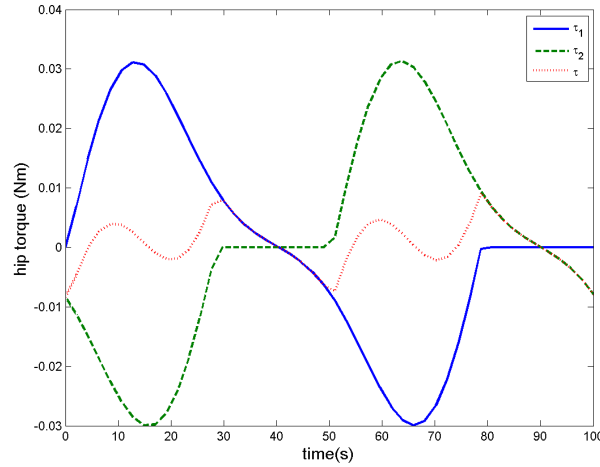


Fig. 3: Hip torques of different leg (τ_1 and τ_2) and total hip torque τ for a complete gait cycle of walking at $1[m/s]$ with FMCH model.

B. FMCH approximation of hip torque in human walking

In this section we use the FMCH model to explain human posture control. If Eq. (12) holds for human hip torque control, relation of $r_F^\tau = \frac{\tau_h}{F_S}$ with ψ will be linear. In Fig. 4, these relations are shown for different walking speeds (from $25\% - 125\%PTS$). It is observed that the curves can be approximated by straight lines. The hip stiffness (c) and rest angle (ψ_0), found from the line slope and its intersection with horizontal axis, are shown in Fig. 5. The hip stiffness decreases with increasing the motion speed except from $100\%PTS$ to $125\%PTS$. It means that for faster movement more oscillations are allowed for the upper body, except for very fast walking. However, at $125\%PTS$ running is preferred to walking, requiring stiffer hip. The trend is the same for rest angle in the opposite direction.

Based on the parameters found for the FMCH model, the normalized hip torque (to body weight and length) found in experiment and the approximation of FMCH are drawn in Fig. 6. It is shown that the model can predict the hip torque using a modulating hip compliance by simple reflex from leg force. Some deviations from the prediction are observed at the the beginning of the cycle for fast walking. It might be the effect of large push off at high speeds which will be handled after passing 20% of swing phase (single support). It is observed that the FMCH model can properly explain the hip torque control strategy in humans.

C. VPP estimation by FMCH

In [8], the method mentioned in Sec.II-D is presented to find VPP (for mathematical details see [24]). Here, we use the new method to find the VPP based on FMCH model as explained in Appendix. B. The ground reaction forces are plotted from CoP by dashed lines in Fig. 7 where the coordinate system centered at CoM and aligned with upper body orientation are shown. The CoM and the estimated VPP

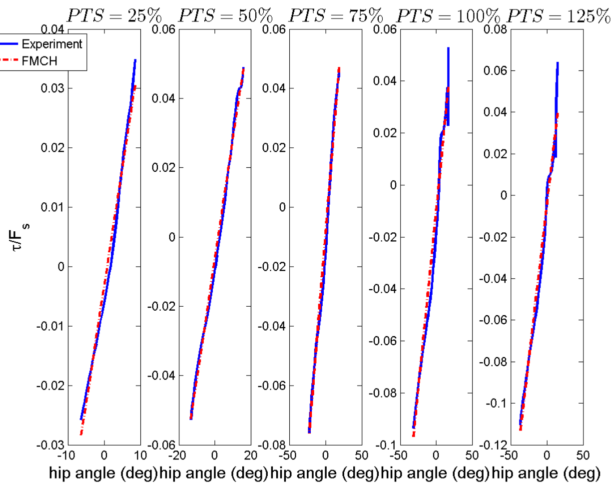


Fig. 4: Hip torque τ_h to leg force ratio (r_F^τ) versus hip angle (ψ). Solid line is the experimental result and dashed line shows the fitted FMCH model.

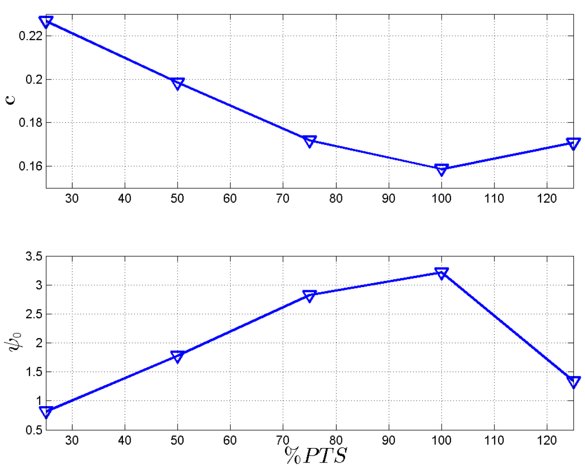


Fig. 5: Variations of normalized hip stiffness (c) and rest angle (ψ_0) according to walking speeds.

are also shown with green and red circles, respectively. It is clear that the estimation of VPP by FMCH is the focus point in these graphs. Hence, VPP can be physically implemented by FMCH model, just using leg force feedback.

IV. DISCUSSION

Using oscillatory behavior of spring mass system, SLIP as a template model can describe bouncing property of legged locomotion. Here, we presented another template model for balancing, comprised of a (rotational) spring and inverted pendulum (as oscillatory system) with feedback signal to adjust the stiffness. The new template, called *FMCH*, employs feedback signal for tuning the property of the passive mechanical system (spring) similar to reflex model in neuromuscular system [17]. Instead of force-velocity [18] or stimulation [17], we suggest to adapt stiffness as force-length

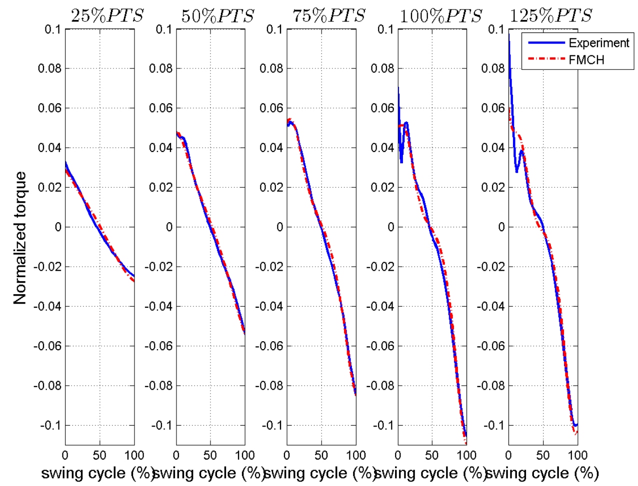


Fig. 6: Normalized hip torque to body weight and leg length during single support cycle.

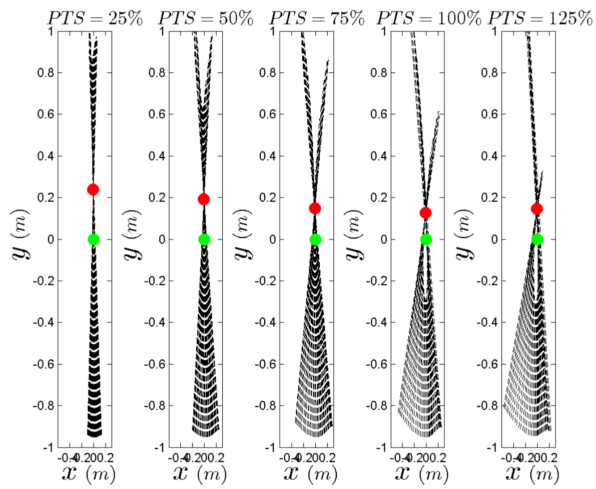


Fig. 7: Ground reaction forces from CoP. The coordinate system is centered at CoM (green circle) and aligned with trunk orientation. Red circles show VPPs, estimated by FMCH.

property, based on muscle reflex.

With FMCH model, we can explain the hip torque control in human walking. The closer linear relation between hip torque over leg force (introduced by r_F^τ) and hip angle, the better representation of human posture control by FMCH. Thus, linear curves in Fig. 4 support the idea of using leg muscle reflex for hip muscle control. In addition, existence of such kinds of sensors for measuring leg configuration in human body was already shown [26], e.g., the hip bi-articular muscles may change their properties based on vastus muscle length measuring the leg force, in human body.

Changing the stiffness and rest angle of hip spring with respect to motion speed shows that the balance control strategy may contribute to gait speed adjustment, except for very fast walking. This property besides leg angle and stiffness adjustment can precisely control the motion speed.

Finally, with mathematical relation between the hip normalized stiffness and rest length and position of VPP with respect to CoM (the distance and angle from upper body axis), we proposed a method to find VPP based on FMCH model. The main benefit of such calculations appears when the VPP changes during gait e.g., to recover from perturbations or to change the gait speed. In such cases VPP adaptation can be detected from slope changes in $r_F^\tau - \psi$ curves.

APPENDIX

A. Relation between FMCH and VPP

From Fig. 2, the distance between VPP and hip (r) is

$$r = \sqrt{r_{\text{VPP}}^2 + r_h^2 + 2 r_h r_{\text{VPP}}\ cos\gamma} \tag{14}$$

In addition, the angle between line from VPP to hip and trunk axis ψ' can be found by

$$\psi' = \arctan \frac{r_{\text{VPP}}\ sin\gamma}{r_h + r_{\text{VPP}}\ cos\gamma}. \tag{15}$$

If VPP angle $\gamma < 20^\circ$, (14) and (15) can be approximated by

$$\begin{cases} r &= r_{\text{VPP}} + r_h \\ \psi' &= \frac{r_{\text{VPP}}\gamma}{r} \end{cases} \tag{16}$$

Eq. (10) gives the required torque τ_{VPP} to have GRF going through VPP. For hip angle range during walking ($\psi < 30^\circ$), this equation can be approximated by the following equation with error less than 1.5%

$$\tau_{VPP} \approx F_s\ l\ \frac{(r_h + r_{\text{VPP}})\psi - r_{\text{VPP}}\gamma)}{l + r_h + r_{\text{VPP}}} = F_s\ l\ \frac{r(\psi - \psi')}{l + r} \tag{17}$$

Setting the hip rest angle to ψ' (meaning $\psi_0 = \psi'$) and replacing l by its average during single support results in the hip torque generated by FMCH. The approximation error for parameter ranges used in walking is less than 2%.

B. Approximating VPP with FMCH

First, we find FMCH parameters (c and ψ_0) from linear approximation of $r_F^\tau - \psi$ curve. Then, r as the approximated distance of VPP to hip is computed by the following equation

$$r = \frac{\bar{l}c}{(\bar{l} - c)} \tag{18}$$

in which $\bar{l}$ is the average leg length in single support. From (11), (12) and (18) the VPP parameters are calculated as

$$\begin{cases} r_{\text{VPP}} &= r - r_h \\ \gamma &= \frac{r\psi_0}{r_{\text{VPP}}} \end{cases} \tag{19}$$

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