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Strategy for Flexible Walking

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Abstract. An important feature of human locomotor control is the instant adaptability to unpredictable changes of conditions surrounding the locomotion. Humans, for example, can seamlessly adapt flexibly their walking gait following various perturbations to walking speed such as external forces (e.g., induced by a collision with humans) or body changes (e.g., sudden ankle impairment as a result of an injury). In this study, we hypothesize that such various perturbations can be overcome by flexible changes of walking pattern induced through modulating the body form (posture) at the beginning of the stance phase—*the initial state*. Using a walking model, we validate our hypothesis through computer simulations.

1. Introduction

An important feature of human locomotor control is the instant adaptability to unpredictable changes of conditions surrounding the locomotion. Indeed, human walk is robust both to environmental changes (e.g., an external force occurrences induced by a collision with humans) and to physical changes (e.g., impairment from injury). If anything, these perturbations result in a modified walking pattern, although such change can sometimes be dramatic. The mechanisms underlying such flexible control are of interest to scholars in neurophysiology, biological cybernetics, and biomechanics. This study is directed towards gaining a theoretical understanding of the mechanisms of flexible locomotor control in the presence of various changes such as environmental or physical (body) changes.

Based on neurophysiological evidence, theoretical studies on bipedal locomotion (TAGA *et al.*, 1991; TAGA, 1994) have shown that a walking movement emerges from the interaction of oscillations of a central pattern generator (CPG) (GRILLNER, 1985) and a body, i.e., *the integrated system composed of CPG and body*. In the phase space, a walking pattern is formed as a limit cycle and the robustness of a walking pattern can be attributed to the stability of the limit cycle. Therefore, when a given perturbation is over the stability of the limit cycle, the walking system requires a flexible change of the walking pattern.

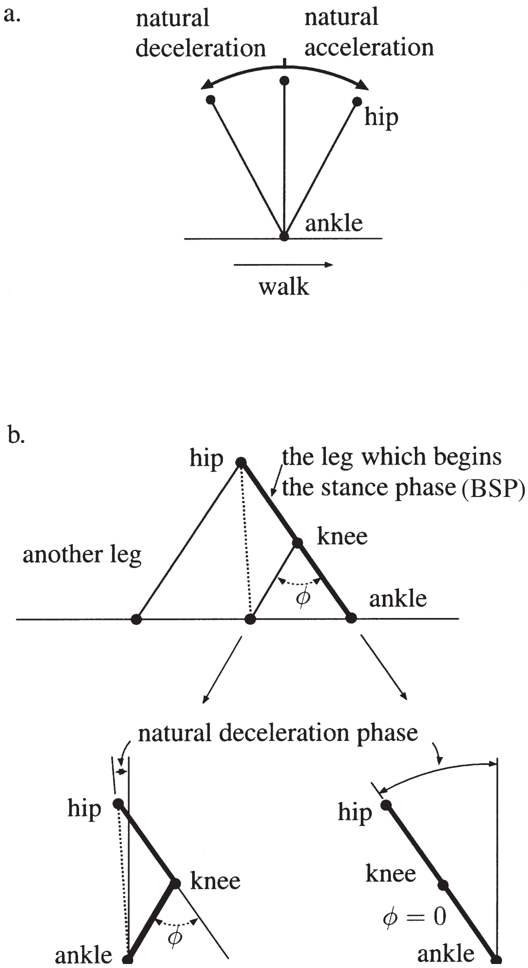

Fig. 1. A biomechanical view of the bipedal walking. a. During walking, the whole body takes a step forward with the inverted pendulum motion of the leg in the stance phase. The forward motion of the body is naturally decelerated or accelerated in the first half or the latter half of the stance phase, respectively. b. Depending on the knee joint angle ϕ of the leg which begins the stance phase, the length of the natural deceleration changes. The thick line and ϕ denote the leg which begins the stance phase and the knee joint angle, respectively.

From a biomechanical point of view, the motion of each leg during bipedal locomotion can be described as being either in the swing phase, or in the stance phase (PERSON, 1976). In the swing phase, the foot is off the ground and the leg moves forward as a pendulum. In the stance phase, the foot is on the ground and the leg moves as an inverted pendulum. The top of this inverted pendulum (the hip joint) works as an axle of the pendulum motion of the other leg (in the swing phase). The leg in stance phase develops the propulsive force needed to forward the whole body, through interaction with the ground. Thus, the motion of the whole body is left entirely to the inverted pendulum motion of the leg in the stance phase. In the normal walking, as shown in Fig. 1a, the forward motion of the body is naturally decelerated or accelerated in the first half or the latter half of the stance phase, respectively. As shown in Fig. 1b, when the knee joint of the leg which begins the stance phase is flexed and extended, the natural deceleration phase is shortened and lengthened, respectively. Thus, the knee joint angle ϕ of the leg which begins the stance phase significantly affects the body speed. Therefore, the modulation of ϕ will result in control of the body speed. This will mean that perturbations to the speed of the walking body, which can cause the body to fall, can be compensated by modulating the body form (posture; ϕ) in the beginning of the stance phase (BSP) so that walking can be sustained.

It is neurophysiologically known that animals initiate locomotion after the posture is adjusted, and the control of the posture is involved to the end of the locomotion (SHERRINGTON, 1906, 1910). That is, a locomotor control involves not only a generation of a stepping rhythm but also a posture control; clinical trials on human patients (CALANCIE *et al*., 1994; DIMITRIJEVIC *et al*., 1998) have also shown that although the activity of the spinal cord including the locomotor CPG can induce the legs to rhythmically step (on a bed), this motion is definitely different from the locomotion where a balanced body goes ahead. Neurophysiological experiments on cats showed that by receiving proprioceptive sensory signals through the mossy fibers system, Purkinje cells in the cerebellum can modulate the activity of the motor neurons in the spinal cord. When the paravermal part of lobule IV and V or the vermal part of lobule V in the cerebellum is partially cooled, an excessive flexion or extension of the legs, respectively, is induced before and after the BSP, which results in the failure to walk (UDO *et al*., 1979, 1980). The activity of Purkinje cells in the paravermal part of lobule V or in the vermal zones become significantly high around the BSP (UDO *et al*., 1981; ARMSTRONG and EDGLEY, 1984). From these, it is obvious that the Purkinje cells participate in the moderate extension of the legs in the BSP (CHAMBERS and SPRAGUE, 1955; UDO *et al*., 1976). On the other hand, from a biomechanical viewpoint, the moderate extension of the legs in the BSP is the key to accomplishing a walking, as mentioned in the above paragraph. Thus, we consider that the posture control of the leg in the BSP by Purkinje cells plays a key role in establishment of adaptive walking. In human bipedal walking, an extension of the leg depends roughly on the angle of the knee joint only. In this study, we will focus on the relationship between the modulation of the knee joint angle in the BSP and the adaptive change of a walking pattern.

OHGANE *et al*. (2004a) has proposed the walking model adaptable to a sudden level change of the ankle joint torque which play a principal role in the development of the propulsive force for locomotion. They have reported that such a sudden body change can be overcome by flexible changes of the walking pattern induced through modulating the posture at the BSP. In this study, we hypothesize that various perturbations to walking speed such as the body change or environmental changes, which can cause the system to fall, should be overcome by the consistent strategy, i.e., the modulation of the body form in the BSP. A consistent strategy, if indeed it is effective to overcome various perturbations, is very useful under the world where various conditions unpredictably change. Since perturbations to a walking speed from the environment could be almost approximated as the external force (to the body center), we consider the case where a walking is strongly disturbed by external forces (e.g., induced by a collision with humans). The effectiveness of the body form (posture) in a certain phase has never been discussed in terms of robustness of locomotion, although models of locomotor pattern's adaptive change have been proposed (TAGA, 1998; ITO *et al.*, 1998). Based on this hypothesis, we constructed a numerical model of bipedal walking, and validated the hypothesis through computer simulations.

2. The Walking Model

The model is constructed by adding a posture controller to the model of *the integrated system composed of CPG and body* (TAGA *et al.*, 1991; OHGANE *et al.*, 2004b). Our model thus consists of the body and the neural system composed of the CPG and the posture controller, as shown in Fig. 2 (Appendix A).

The body is similar to that of TAGA *et al.* (1991). The body consists of an interconnected chain of 5 rigid links in the sagittal plane, as shown in Fig. 4. The motion for the body is represented by differential equations of a ($\mathbf{x} = 6 \times 1$) vector of mass point positions of 1 link and inertial angles of 4 links. The equations are derived by the Newton-Euler method (Appendix A).

The neural system is made up of coupled Bonhoeffer-van der Pol (BVP) neurons. The outputs of the neural system are sent as motor signals to the body. The body state is considered as sensory feedback to the neural system through the feedback pathway.

The body moves according to its own dynamics, under the conditions of the environment, the body, and motor signals from the neural system. The neural system is activated by its own dynamics, under the constraints of sensory feedback which expresses the state of the body. Thus, the body and the neural system are coupled through a recurrent loop.

The neural system (Fig. 2) is represented by the following differential equations:

$$\begin{cases} \tau_i \dot{u}_i(t) = u_i(t) - v_i(t) - u_i(t)^3/3 + \sum_{j=1}^{16} w_{ij} y_i + u_0 + F_i(\mathbf{x}(t)), \\ \tau'_i \dot{v}_i(t) = u_i(t) + a - b v_i(t), \\ y_i = f(u_i(t)), \quad f(u) = \max(0, u), \quad (i = 1,16), \end{cases} \tag{1}$$

where, u_i is the potential of the ith neuron; v_i is responsible for the accommodation and refractoriness of the ith neuron; w_{ij} is the connecting weight from the ith neuron to the jth neuron; τ_i and τ'_i are the time constants of the inner state and the accommodation and refractory effects, respectively; y_i is the output of the ith neuron; u_0 is a constant parameter. F_i is a sensory feedback from the body (Appendix A), and $\mathbf{x}$ is a (6×1) vector of the mass point positions of 1 link and the inertial angles of 4 links; t is time; a and b are positive constants. The CPG comprises neurons 1 to 12.

The posture controller consists of neurons 13 to 16. To modulate the knee joint angle in the beginning of the stance phase (BSP), it realizes the following two tasks; (1) the

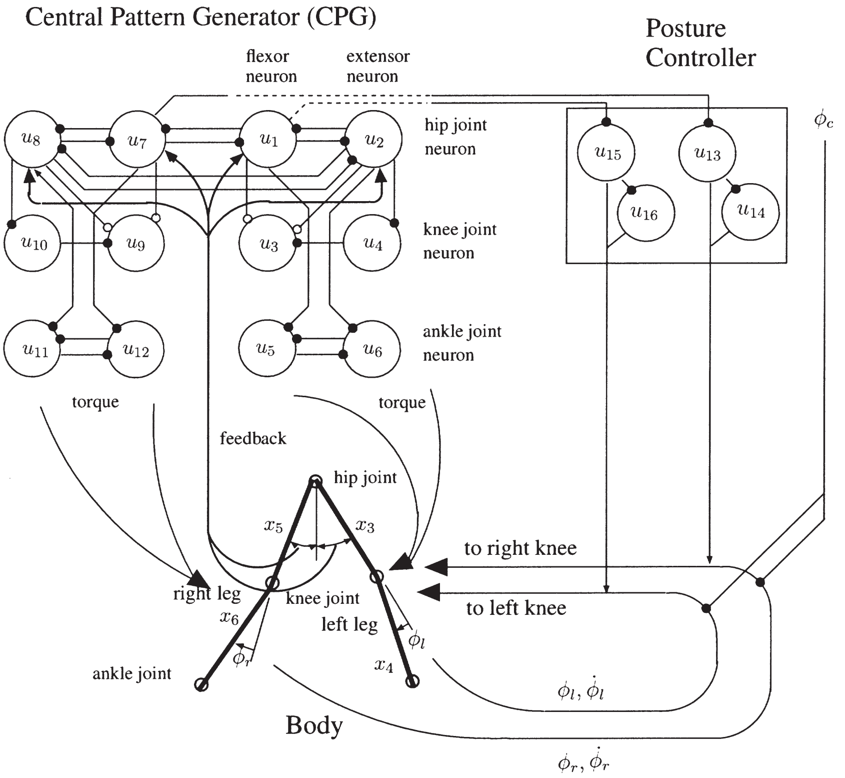

Fig. 2. The neural system composed of the central pattern generator (neurons 1 to 12) and the posture controller (neurons 13 to 16). ○ and ● denote excitatory and inhibitory connections respectively; the motion of the hip, the knee and the ankle joint in the left leg is governed by neurons 1–2, 3–4, 5–6 and 13–14 respectively. Similarly, the motion of the joints in the left leg is governed by neurons 7 to 12 and 15–16. Odd-numbered neurons and even-numbered neurons in the CPG control represent flexion and extension of the joint, respectively. The posture controller gives the equilibrium angle ϕ_c of the knee joints ϕ_l and ϕ_r in the BSP.

transformation of perturbations into the equilibrium angle of the knee joint in the BSP; (2) outputting the motor commands only in the BSP, which can be attained by receiving the CPG output. The posture modulation is executed by the motor commands (T_{bspl}, T_{bspr}) represented by the following equations:

$$\begin{cases} T_{bspl} = g\big(g(u_{13}) - g(u_{14})\big)\Big(p_{b1}(x_3 - x_4 - \phi_c) - p_{b2}(\dot{x}_3 - \dot{x}_4)^2\Big), \\ T_{bspr} = g\big(g(u_{15}) - g(u_{16})\big)\Big(p_{b1}(x_5 - x_6 - \phi_c) - p_{b2}(\dot{x}_5 - \dot{x}_6)^2\Big), \\ g(z) = \begin{cases} 0, & \text{for } z \le 0 \\ 1 & \text{otherwise.} \end{cases} \end{cases} \tag{2}$$

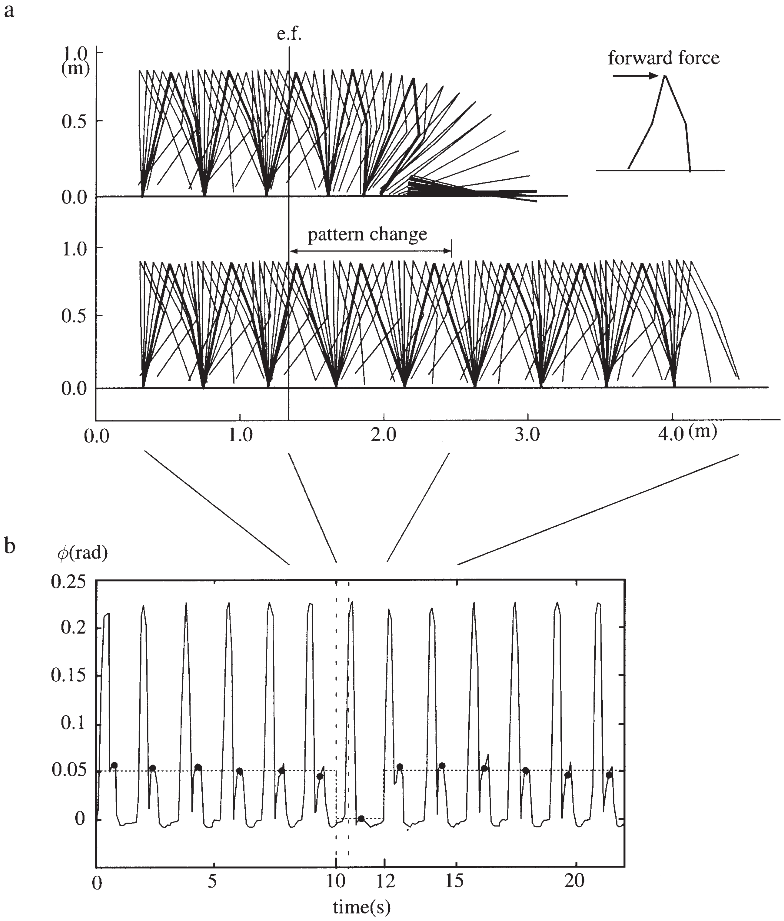

Fig. 3. (a) Stick figure of the walking motion when the forward force of 150 N was applied to the hip during 0.5 s from the indicated solid line. The upper stick figure shows the case where the equilibrium angle of the knee joint angle in the BSP (ϕ_c) was always set at 0.05π rad. The bottom stick figure shows the case where ϕ_c was set at 0.0π rad during 2.0 s after the force was given, and except for this time, ϕ_c was set at 0.05π rad. This modulation enabled the walking system to overcome the given external force by inducing the walking pattern change. (b) The motion of the left knee joint angle ϕ in the walking shown in the bottom figure in (a). Between the two broken lines, the force was given. The dotted line denotes ϕ_c. ● denotes the knee joint angle ϕ at the BSP. (c) Stick figure of the walking motion when the backward force of 150 N was given to the hip during 0.5 s from the indicated solid line. The upper stick figure shows the case where the equilibrium angle of the knee joint angle in the BSP (ϕ_c) was always set at 0.05π rad. The bottom stick figure shows the case where ϕ_c was set at 0.08π rad during 2.0 s after the force was given, and except for this 2.0 s ϕ_c was set at 0.05π rad. This modulation enabled the walking system to overcome the given external force by inducing the walking pattern change. (d) The motion of the left knee joint angle ϕ in the walking shown in the bottom figure in (c). Between the two broken lines, the force was given. The dotted line denotes ϕ_c. ● denotes the knee joint angle ϕ at the BSP.

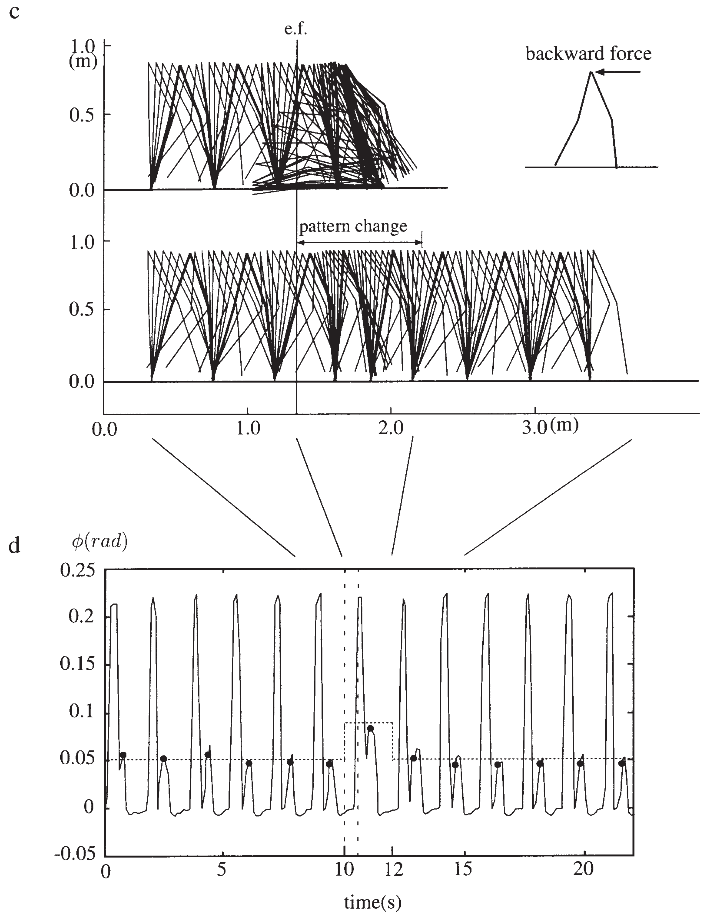

Fig. 3. (continued).

where, T_{bsp_l} and T_{bsp_r} are torques generated in the knee joint of the left and right leg, respectively; ϕ_c is the equilibrium angle in the knee joints, the value of which will be given in the next section; p_{b1} and p_{b2} are constants (in this paper, p_{b1} = 400 and p_{b2} = 40 are assumed). Through the modulation of the natural frequency and the connections from the CPG to the posture controller, this motor outputs were set up to be generated in each BSP.

It is known that Purkinje cells in the cerebellum receive the outputs from the CPG (ARSHAVSKY *et al.*, 1983; ITO, 1984). And, as mentioned in introduction, Purkinje cells in the cerebellum strongly participate in the moderate extension of the legs in the BSP. Thus, the posture controller presented here may correspond to Purkinje cells in lobule V of the paravermal part and the vermal part, and so on, of the cerebellum.

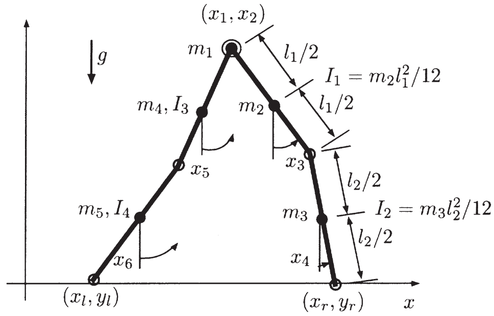

Fig. 4. Model of bipedal body as an interconnected chain of 5 rigid links (a point mass m_1 on the hip and four rigid bodies I_i (i = 1, 4)).

3. Results

We considered the case in which the hip is suddenly pushed forward or backward by an external force. The following conditions were assumed; normally, the equilibrium angle of knee joints ϕ_c (in Eq. (2)) was set at 0.05π rad, which we call the normal condition; after the walking became steady, an external force was given during 0.5*s*; ϕ_c was modulated into $\phi_c = \phi_m$ only a moment around the BSP during 2.0*s* after the external force began to act on (only, the BSP was autonomously detected by the walking system), which we call the modulated condition.

First, giving the force forward, the robustness of the walking system was tested under the normal condition, i.e., under no modulation of the knee joint angle ($\phi_m = 0.05\pi$ rad) even though the force was given). In such a case, the walking could be maintained when the force was weaker than 20 N (*Newton*); when the force was stronger than the value, the walking system fell forward (see Fig. 3a). Then, the robustness of the walking system was tested under the modulated condition; ϕ_m was set at 0.0 rad against the forward force stronger than 20N. In this case, the walking could be maintained when the force was weaker than 150 N (see Fig. 3b). This result was little changed although the phase when the external force was given was variously changed. This result agrees with that in a biomechanical experiment on human walk (FORNER CORDERO *et al.*, 2004) that shows disturbances induced by forward forces are coped with the large stride.

Next, the adaptability to the backward force was investigated. Set at $\phi_m = 0.08\pi$ rad, similar results were obtained (Figs. 3c–d).

4. Discussion

Our model was shown to induce walking to fit to strong external forces which significantly disturbed the body speed and pushed down the walking system without the fitting mechanism. As mentioned in introduction section, our model can be expected to overcome various perturbations to the body speed regardless of kinds of the perturbation, i.e., such as physical (body) or environmental changes. The formulation of ϕ_c in Eq. (2) as a function of the body speed will lead to the refinement of the model to the description of an autonomously adaptive walking system.

Theoretical studies (NISHIURA *et al*., 2003; TERAMOTO *et al*., 2004) have pointed out that border states (neutral states) latent in dynamical systems such as reaction-diffusion systems play a key role in changes of the system behaviors. As shown in Fig. 1b, the modulation of the leg extension in the BSP can bring about the control of the natural deceleration degree of the body motion. Under the no perturbation condition, when the knee joint angle ϕ of the leg which begins the stance phase is smaller than the border state ϕ_b, the body is too decelerated and falls backward (data not shown). Conversely, when ϕ is larger than the another border state ϕ_f, the body falls forward. That is, the neutral states (border states) exist in ϕ, and from a mathematical point of view, these neutral states will lead the system into the unstable periodic solutions. Thus ϕ determines the behavior of *the integrated system composed of CPG and body*, i.e., ϕ works as the initial state. When the initial state ϕ is in the width between the two neutral states (ϕ_b and ϕ_f), the system can continue to walk. Normally the ϕ can stably be in the width between the neutral states, i.e., the two neutral states (ϕ_b and ϕ_f) normally are latent. However perturbations shift the neutral states and cause ϕ to go out from the width between the neutral states if ϕ is not changed. We call this "surfacing of the neutral state".

Our strategy for the flexible locomotor control of *the integrated system composed of CPG and body* can be described as follows; (1) the system state in the phase when the neutral state (ϕ_b or ϕ_f) can surface should be regarded as the initial state which determines the behavior of the system; (2) the initial state should be renewed so that it is kept from going out of the side where limit cycle exists. This initial state will be one of the constraints of the walking system. The question of how biological systems autonomously produce the constraints by themselves (SHIMIZU, 1994) is important. This study hints that the constraint will be generated as a function of the neutral state.

We would like to thank Dr. Masato Iida and Dr. Luc Berthouze for useful discussion and technical assistance.

Appendix A. The Equations of Motion for the Body

All variables and conventions correspond to those shown in Fig. 4. By using the Newton-Euler method, motion of the body (TAGA, 1991) can be written as follows.

$$P(\mathbf{x})\ddot{\mathbf{x}} = \mathbf{Q}\big(\mathbf{x}, \dot{\mathbf{x}}, T_r(\mathbf{y})\big),$$

therefore,

$$\ddot{\mathbf{x}} = \left[P(\mathbf{x})\right]^{-1}\mathbf{Q}\big(\mathbf{x}, \dot{\mathbf{x}}, T_r(\mathbf{y})\big),$$

where,

$$\mathbf{x} = \begin{pmatrix} x_1 \\ x_2 \\ x_3 \\ x_4 \\ x_5 \\ x_6 \end{pmatrix}$$

$$\begin{aligned}
&p_{11} = \sum_{n=1}^{5} m_n, && p_{21} = 0,\\
&p_{12} = 0, && p_{22} = \sum_{n=1}^{5} m_n,\\
&p_{13} = (0.5m_2 + m_3)I_1 \cos(x_3), && p_{23} = (0.5m_2 + m_3)I_1 \sin(x_3),\\
&p_{14} = 0.5m_3 l_2 \cos(x_4), && p_{24} = 0.5m_3 l_2 \sin(x_4),\\
&p_{15} = (0.5m_4 + m_5)l_3 \cos(x_5), && p_{25} = (0.5m_4 + m_5)l_3 \sin(x_5),\\
&p_{16} = 0.5m_5 l_4 \cos(x_6), && p_{26} = 0.5m_5 l_4 \sin(x_6),\\
&p_{31} = (0.5m_2 + m_3)I_1 \cos(x_3), && p_{41} = 0.5m_3 l_2 \cos(x_4),\\
&p_{32} = (0.5m_2 + m_3)I_1 \sin(x_3), && p_{42} = 0.5m_3 l_2 \sin(x_4),\\
&p_{33} = 0.25m_2 l_1^2 + m_3 l_1^2 + I_1, && p_{43} = 0.5m_3 l_1 l_2 \cos(x_3 - x_4),\\
&p_{34} = 0.5m_3 l_1 l_2 \cos(x_4 - x_3), && p_{44} = I_2 + 0.25m_3 l_2^2,\\
&p_{35} = 0, && p_{45} = 0,\\
&p_{36} = 0, && p_{46} = 0,\\
&p_{51} = (0.5m_4 + m_5)I_3 \cos(x_5), && p_{61} = 0.5m_5 l_4 \cos(x_6),\\
&p_{52} = (0.5m_4 + m_5)I_3 \sin(x_5), && p_{62} = 0.5m_5 l_4 \sin(x_6),\\
&p_{53} = 0, && p_{63} = 0,\\
&p_{54} = 0, && p_{64} = 0,\\
&p_{55} = (0.25m_4 + m_5)l_3^2 + I_3, && p_{65} = 0.5m_5 l_3 l_4 \cos(x_5 - x_6),\\
&p_{56} = 0.5m_5 l_3 l_4 \cos(x_6 - x_5), && p_{66} = 0.25m_5 l_4^2 + I_4,
\end{aligned}$$

$$\begin{aligned}
q_1 = &(0.5m_2 + m_3)l_1 \sin(x_3)\dot{x}_3^2 + 0.5m_3 l_2 \sin(x_4)\dot{x}_4^2\\
&+(0.5m_4 + m_5)l_3 \sin(x_5)\dot{x}_5^2 + 0.5m_5 l_4 \sin(x_6)\dot{x}_6^2 + F_{g1} + F_{g3}
\end{aligned}$$

$$q_2 = -(0.5m_2 + m_3)l_1 \cos(x_3)\dot{x}_3^2 - 0.5m_3 l_2 \cos(x_4)\dot{x}_4^2$$
$$-(0.5m_4 + m_5)l_3 \cos(x_5)\dot{x}_5^2 - 0.5m_5 l_4 \cos(x_6)\dot{x}_6^2 + F_{g1} + F_{g2} - \sum_{n=1}^{5} m_n g$$

$$q_3 = 0.5m_3 l_1 l_2 \sin(x_4 - x_3)\dot{x}_4^2 + F_{g1} l_1 \cos(x_3) + F_{g2} l_1 \sin(x_3)$$
$$-(m_2 + 2m_3)0.5 g l_1 \sin(x_3) + T_{rp1} + T_{r1} - T_{r2} - T_{r4}$$

$$q_4 = 0.5m_3 l_1 l_2 \sin(x_3 - x_4)\dot{x}_3^2 - 0.5m_2 g l_2 \sin(x_4) + F_{g1} l_2 \cos(x_4)$$
$$+F_{g2} l_2 \sin(x_4) + T_{rp2} + T_{r2} - T_{r3}$$

$$q_5 = 0.5m_5 l_3 l_4 \sin(x_6 - x_5)\dot{x}_6^2 - 0.5(m_4 + 2m_5) g l_3 \sin(x_5) + F_{g3} l_3 \cos(x_5)$$
$$+F_{g4} l_3 \sin(x_5) + T_{rp3} + T_{r4} - T_{r5} - T_{r1}$$

$$q_6 = 0.5m_5 l_3 l_4 \sin(x_5 - x_6)\dot{x}_5^2 - 0.5m_4 g l_4 \sin(x_6)$$
$$+F_{g3} l_4 \cos(x_6) + F_{g4} I_4 \sin(x_6) + T_{rp4} + T_{r5} - T_{r6}.$$

Horizontal and vertical forces on the ankles are given by:

$$F_{g1} = \begin{cases} -k_g(x_r - x_{r0}) - b_g \dot{x}_r & \text{for } y_r - y_g(x_r) < 0 \\ 0 & \text{otherwise,} \end{cases}$$

$$F_{g2} = \begin{cases} -k_g(y_r - y_{r0}) + b_g f(-\dot{y}_r) & \text{for } y_r - y_g(x_r) < 0 \\ 0 & \text{otherwise,} \end{cases}$$

$$F_{g3} = \begin{cases} -k_g(x_l - x_{l0}) - b_g \dot{x}_l & \text{for } y_l - y_g(x_l) < 0 \\ 0 & \text{otherwise,} \end{cases}$$

$$F_{g4} = \begin{cases} -k_g(y_l - y_{l0}) + b_g f(-\dot{y}_l) & \text{for } y_l - y_g(x_l) < 0 \\ 0 & \text{otherwise,} \end{cases}$$

where $y_g(x)$ is the function which represents the terrain. When the ground is level, $y_g(x) = 0$. (x_r, y_r) and (x_l, y_l) represent the positions of the ankles, which are given by:

$$(x_r, y_r) = (x_1 + l_1 \cos x_3 - l_2 \cos x_4, x_2 - l_1 \sin x_3 - l_2 \sin x_4),$$

$$(x_l, y_l) = (x_1 + l_1 \cos x_5 + l_2 \cos x_6, x_2 - l_1 \sin x_5 - l_2 \sin x_6).$$

Passively generated torques at each joint are given by:

$$T_{rp1} = k_r f(x_4 - x_3) - b_r g(x_4 - x_3)(x_4 - x_3) - b(\dot{x}_3 - \dot{x}_5) - b(\dot{x}_3 - \dot{x}_4),$$

$$T_{rp2} = -k_r f(x_4 - x_3) + b_r g(x_4 - x_3)(x_4 - x_3) - b(\dot{x}_4 - \dot{x}_3) - b\dot{x}_4,$$

$$T_{rp3} = k_r f(x_6 - x_5) - b_r g(x_6 - x_5)(x_6 - x_5) - b(\dot{x}_5 - \dot{x}_3) - b(\dot{x}_5 - \dot{x}_6),$$

$$T_{rp4} = -k_r f(x_6 - x_5) + b_r g(x_6 - x_5)(x_6 - x_5) - b(\dot{x}_6 - \dot{x}_5) - b\dot{x}_6,$$

where k and b are the positive constants.

Actively generated torques at each joint are given by:

$$T_{r1} = p_1(y_1 - y_2) + g(y_1)\left(-p_e\left(f(x_3 - x_5 - x_0)\right)^2 - p_v(\dot{x}_3 - \dot{x}_5)\right),$$

$$T_{r2} = p_2(y_3 - y_4) + T_{bspl},$$

$$T_{r3} = (p_3 y_3 - p_4 y_4)g(F_{g2}),$$

$$T_{r4} = p_1(y_7 - y_8) + g(y_7)\left(-p_e\left(f(x_5 - x_3 - x_0)\right)^2 - p_v(\dot{x}_5 - \dot{x}_3)\right),$$

$$T_{r5} = p_2(y_9 - y_{10}) + T_{bspr},$$

$$T_{r6} = (p_3 y_{11} - p_4 y_{12})g(F_{g4}),$$

where p is the positive constants; α_l and α_r are coefficients which express the ankle joint torque level of the left and right leg, respectively. Depending the ith neuronal output y_i, the mechanical visco-elastic torque at hip joint was assumed to be produced when the hip joint angle between the left and right thigh was beyond a threshold angles x_0.

The sensory feedback F_i to the ith neuron is given as follows (TAGA, 1991):

$$\begin{cases} F_i(i=1\cdots 12) \text{ are given} \\ F_1=F, \quad F_2=F', \quad F_7=F', \quad F_8=F, \quad F_i=0 \text{ (else)}. \\ \text{Here,} \\ F=f(-x_3)-f(-x_5), \; F'=f(-x_5)-f(-x_3). \end{cases} \tag{A1}$$

Besides,

$$f(z)=\max(0,z), \quad g(z)=\begin{cases} 0, & \text{for } z \le 0 \\ 1 & \text{otherwise.} \end{cases}$$

Appendix B. Simulation Parameters

1. Body

$m_1 = 48.0$, $m_2 = 7.0$, $m_3 = 4.0$, $m_4 = 7.0$, $m_5 = 4.0$,
$l_1 = 0.4$, $l_2 = 0.5$, $l_3 = 0.4$, $l_4 = 0.5$,
$I_1 = m_2 l_1^2/3$, $I_2 = m_3 l_2^2/3$, $I_3 = m_4 l_3^2/3$, $I_4 = m_5 l_4^2/3$,
$k_g = 30000.0$, $k_r = 2000.0$, $b_g = 3000.0$, $b_r = 200.0$, $b = 1.0$,
$p_1 = 21.0$, $p_2 = 21.0$, $p_3 = 21.0$, $p_4 = 5.0$, $p_e = 500.0$, $p_v = 50.0$,
$p_{b1} = 400.0$, $p_{b2} = 40.0$,
$x_0 = 0.1\pi$ rad, $g = 9.8$ m/s^2.

2. Neural System (Central Pattern Generator and Posture Controller)

$\tau_i(i = 1 \cdots 16)$ are given
$\tau_1 4 = \tau_{16} = 1/100$, $\tau_i = 1/30$ (else).

$\tau'_i(i = 1 \cdots 16)$ are given
$\tau'_1 4 = \tau'_1 6 = 10{,}0$, $\tau_i = 10/3$ (else).

$u_0 = 0.3$, $\alpha_u = 1.0$, $a = 0.7$, $b = 0.8$.
$w_{1\,5}$, $w_{2\,6}$, $w_{2\,4}$, $w_{4\,3}$, $w_{5\,6}$, $w_{6\,5}$, $w_{1\,7}$, $w_{7\,1} = -1.0$,
$w_{7\,12}$, $w_{8\,12}$, $w_{8\,10}$, $w_{10\,9}$, $w_{11\,12}$, $w_{12\,11}$, $w_{2\,8}$, $w_{8\,2} = -1.0$,
$w_{1\,15}$, $w_{7\,13}$, $w_{13\,14}$, $w_{15\,16} = -1.0$,
$w_{1\,2}$, $w_{2\,1}$, $w_{7\,8}$, $w_{8\,7} = -2.0$,
$w_{1\,3}$, $w_{2\,3}$, $w_{7\,9}$, $w_{8\,9} = 1.0$,
otherwise $w_{ij} = 0.0$.

3. Initial Condition

$$x_1 = 0.0,\ \ x_2 = l_1 + l_2,$$
$$x_3, x_4, x_5, x_6 = 0.0,$$
$$\dot{u}_i = 0.0,\quad \dot{v}_i = 0.0.$$

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