



Effects of hip torque during step-to-step transition on center-of-mass dynamics during human walking examined with numerical simulation

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ARTICLE INFO

Article history:

Accepted 13 April 2019

Keywords:

Human walking
Step-to-step transition
Powered walking model
Hip torque
Energetics
Stability

ABSTRACT

Besides the leg force actuator, humans also use a hip torque actuator during the step-to-step transition to redirect the velocity of CoM (Center of Mass). Although the leg force actuator has been widely studied, few researches analyze the hip torque actuator during the step-to-step transition. In this paper, we build a powered walking model which consists of a point mass linked with two compliant legs. Each leg has a spring and a damper in parallel. Two types of active actuators, the force actuator on the leg and the torque actuator at the hip, are added to simulate the leg force and hip torque actuator during the step-to-step transition. The cycle walk is solved by numerical simulations under different hip torque strength, and the energetics and stability are evaluated. The simulation results show that the hip torque actuator can reduce the energy cost and improve the stability of walking. Further analysis shows that the hip torque actuator can reduce mechanical works of both legs with small extra energy cost. To understand the principle of hip torque actuator, the CoM dynamics is analyzed. It is shown that the hip torque actuator is efficient on the redirection of CoM. Thus, it can improve the stability and reduce required forces of both legs, which decreases the energy cost. Our work provides a fundamental understanding of the hip torque during the step-to-step transition, and may help improve the design of bipedal robots and prosthesis.

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1. Introduction

Human walking is a mechanically complex task which utilizes active forcing by muscle actuators, and robots use different types of actuators to propel walking. Inspired by passive-dynamic walkers, Collins et al. (2005) design several robots which are energy efficient and have very natural gaits. To further understand these robots, Alexander (2005) argues that the main energy interaction occurs at the instant when the walker moves from one step to the next, and at this instant, the velocity of CoM (Center of Mass) must be redirected. This instant can be demonstrated well by the step-to-step transition (Kuo et al., 2005; Adamczyk and Kuo, 2009).

The step-to-step transition is an energetic consequence of human walking, which is inspired by the simple walking model, consisting of a point mass and two rigid, massless leg. In the simple model, the CoM moves along an arc during each step with the conservation of mechanical energy, and substantial energy performs to redirect the CoM velocity at the step-to-step transition between arcs (Kuo et al., 2005; Adamczyk and Kuo, 2009). Many researchers

study this step-to-step transition with the simple walking model or human experiments (Donelan et al., 2002; Yeom and Park, 2011; Kim and Park, 2012). While, these researches mainly focus on the leg-force actuator which is described by Alexander (2005).

Two different actuators are applied on the redirection of CoM velocity, according to Alexander (2005): the leg force and the hip torque. The first actuator, legforce actuator, is further studied, which is turned out to be a high energy efficiency actuator (Kuo, 2001; Kuo et al., 2005). However, the research about the second actuator is much less. This may be limited by the instant step-to-step transition of the simple model, which is difficult to analyze effects of the hip torque actuator with the leg force actuator simultaneously. For human experiments, it is difficult to control the strength of hip torque actuator in nature gaits. Former researches mainly focus on the hip torque during the single support phase (Kuo, 2001; Wisse et al., 2005). Nevertheless, few researches focus on the hip torque actuator during the step-to-step transition. In this paper, a powered walking model is built to explain effects of the hip torque actuator during the step-to-step transition on the CoM dynamics: the model consists of a point mass linked with two massless, compliant legs and can be propelled by leg force or hip torque actuators to simulate the process of step-to-step

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transition. During the double support phase, a torque is applied between two massless legs. It can be treated as the step-to-step transition of this model, which we are interested in. It should be noted that the torque can only be applied during the double support phase, when both legs contact with the ground. Because both legs are massless, applying torque at the single support phase will cause the angular acceleration of the swing leg becoming infinity in theory according to the Euler equation, which does not happen in real human walking.

Cycle walks of the model are solved by numerical simulations. The energetics and stability of walking are analyzed under different actuators conditions. Because of the compliant leg model, the step-to-step transition is not instant. Thus, the process of CoM velocity redirection is shown to provide more principles. From the simulation, we find that the hip torque actuator can reduce the energy cost and improve the stability of walking. Further analysis shows that the hip torque actuator can reduce mechanical works of both legs with small extra energy cost. Our works prove that the hip torque actuator is an efficient actuator, in spite of its limited strength. These findings may be helpful for the understanding of human walking and the design of bipedal robots and prosthesis.

2. Methods

The model represents the body as a point mass at the hip joint linked with two massless, compliant legs (Fig. 1). This model is developed from the spring leg model, and has dampers and active actuators to break the conservation of mechanical energy. The spring leg model is a simple mathematical model intended to mimic running (Blickhan, 1989; McMahon and Cheng, 1990; Geyer et al., 2005), walking (Geyer et al., 2006) and skipping (Andrada et al., 2016). Spring leg model can describe the basic mechanics of human walking such as the M-shaped GRF (Ground Reaction Force) curve, the oscillatory motion of CoM, and the double support phase (Geyer et al., 2006; Whittington and Thelen, 2009; Kim and Park, 2011). The step-to-step transition is non-instant in the spring leg model, which is different from the simple model. This means the spring leg can show the process of the redi-

rection of CoM velocity. On the basis of the spring leg model, Kim and Park (2011) add dampers in the spring legs, which breaks the conservation of mechanical energy, and use this spring-damper-leg model to reproduce the GRF curve of human walking (Lee et al., 2014; Hong et al., 2013). Dampers in legs always dissipate energy, which prevents the model from cycle walking. To realize cycle walks, we propose a powered walking model in which active actuators are added in the spring-damper-leg model.

2.1. The dynamic equations of model

The powered spring-damper-leg model mentioned above consists of a point mass (M) and two massless compliant legs. Each leg comprises a spring with the stiffness k , a damper with the damping factor Cv , and a telescoping axial actuator with force strength F_a in parallel, which compose the leg-force actuator. A rotational actuator with strength τ is added between two legs to simulate the hip torque actuator (Fig. 1). The configuration of each leg is described by the leg angle q_i and leg length l_i ($i = 1, 2$). $i = 1$ refers to the stance leg in single support phase, and the trailing leg in double support phase. $i = 2$ refers to the swing leg in single support phase and the leading leg in double support phase. The following equations are expressed in dimensionless terms, by using overall the mass of CoM (M), original leg length (l_0), gravitational acceleration (g), and time factor $t = \sqrt{l_0/g}$ as based units. For example, $l_1 - l_0$ is expressed by $l_1 - 1$ in following equations. Therefore, velocities will be made dimensionless by the factor $\sqrt{g/l_0}$, the force will be made by Mg , the torque factor is Mgl_0 , and the stiffness factor is Mg/l_0 , etc. The damper is described by the damping ratio $\zeta = Cv/(2\sqrt{kM})$.

The dynamic equations of model are derived by the Lagrangian formulation, and final equations are compared with those from the Newton-Euler formulation. During the single support phase, the force actuator is switched off, and the swing leg is set at a certain step angle α . The equations of the single support phase are:

$$\begin{aligned} \ddot{q}_1 &= \frac{1}{l_1} (\sin q_1 - 2\dot{q}_1 \dot{l}_1) \\ \ddot{l}_1 &= -k(l_1 - 1) - 2\zeta\sqrt{k}l_1 + \dot{q}_1^2 l_1 - \cos q_1. \end{aligned} \quad (1)$$

During the double support phase, the torque actuator and the force actuator of trailing leg are switched on. A constant force F_a and a constant torque τ are implemented (Fig. 2). This constant formulation is chosen to represent a simple, open-loop actuation. This simplification makes it easy to model the system mathematically and provides a fundamental understanding of the hip torque. The equations of the double support phase are:

$$\begin{aligned} \ddot{l}_1 &= -k(l_1 - 1) - k(l_2 - 1) \cos q + \dot{q}_1^2 l_1 + F_a \\ &\quad - 2\zeta\sqrt{k}l_1 - 2\zeta\sqrt{k}l_2 \cos q - \frac{\tau}{l_2} \sin q - \cos q_1 \\ \ddot{q}_1 &= \frac{1}{l_1} [-k(l_2 - 1) \sin q - 2\zeta\sqrt{k}l_2 \sin q \\ &\quad + \frac{\tau}{l_2} \cos q + \frac{\tau}{l_1} + 2\dot{q}_1 \dot{l}_1 - \sin q_1], \end{aligned} \quad (2)$$

where q is the angle between two legs:

$$q = q_2 - q_1. \quad (3)$$

q_2 and l_2 are the angle and length of leading leg as we mentioned before, and can be solved from the geometric constraint:

$$\begin{aligned} l_2 &= \sqrt{l_1^2 + L_s^2 + 2l_1 L_s \sin q_1} \\ q_2 &= \arctan[(L_s + l_1 \sin q_1)/(l_1 \cos q_1)], \end{aligned} \quad (4)$$

where L_s is the step length and can be calculated at Leading Leg Impact:

$$L_s = \sin \alpha - (l_1 \sin q_1)|_{t=t_{LI}}, \quad (5)$$

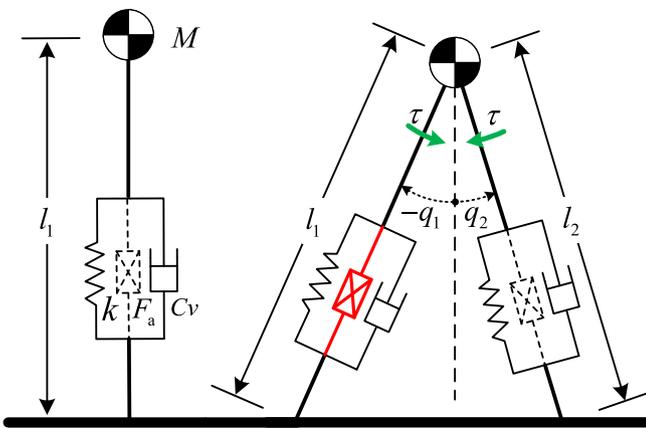


Fig. 1. Schematic of compliant walking model. The model consists of a point mass (M) linked with two massless compliant legs and can be propelled by force/torque actuators. Each leg is represented by a spring with stiffness k , a damper with damper factor Cv , and a telescoping axial actuator with force strength F_a . A rotational actuator with torque τ is exerted between two legs. During double support phase, the torque actuator and the force actuator of the trailing leg are switched on, which are marked in green and red color, respectively. In other time, force actuators are switched off, which are marked in dashed line. We use l_1 and q_1 to describe the configuration of the model. l_2 and q_2 can be calculated from the geometric constraint. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

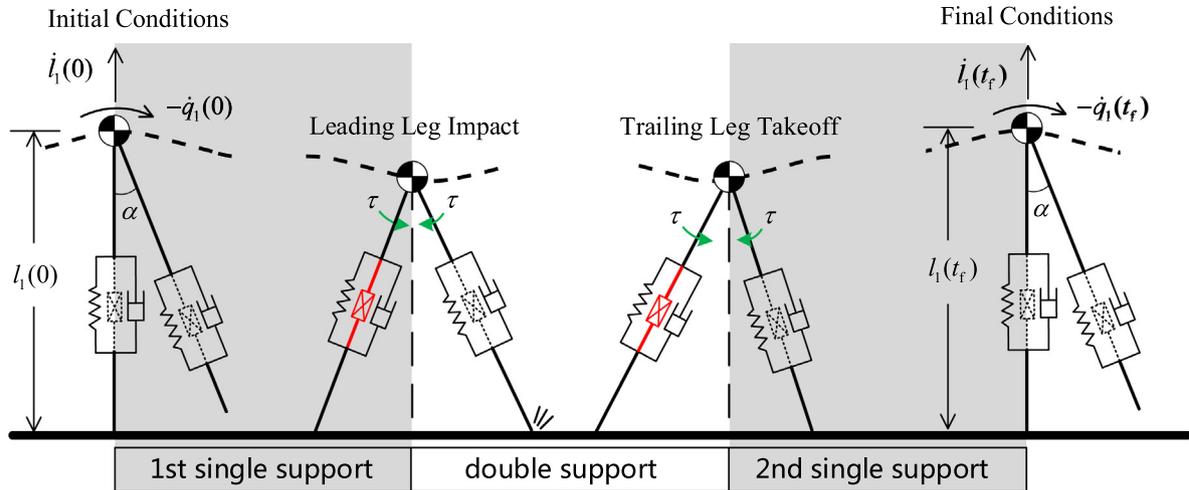


Fig. 2. Schematic of a full walking cycle. Each cycle starts with the support leg in an upright configuration and the swing leg oriented at the step angle α . The double support phase starts when the leading leg contacts the ground and then continues until the Trailing Leg Take-off. During the double support phase, the hip torque actuator and the force actuator of trailing leg are switched on, marked in green and red color, respectively. The second single support phase continued until the leg reaches the upright configuration, signaling the end of a step. The limit cycle solution is searched in which the final conditions (\dot{q}_1, l_1, l_1) replicate the initial conditions. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

where t_{li} is the time of the Leading Leg Impact.

For each walking cycle, the model is initially positioned in upright single support and the swing leg set at the step angle α (Fig. 2). During the first single support phase, the equations of motion (Eq. (1)) are numerically integrated (ode, matlab, the same as follow) forward in time until the event, Leading Leg Impact,

$$q_1 l_1 - \cos \alpha = 0, \quad (6)$$

is detected. Then the model changes into the double support phase, and the Eq. (2) is numerically integrated until the event, Trailing Leg Take-off:

$$l_1 = 1, \quad (7)$$

is detected. We call Eq. (7) the leg length condition. It should be noted that, we do not use the vertical GRF condition ($F_y = 0$, where F_y is the vertical GRF of the leg (Shen and Seipel, 2012; Shen et al., 2014)). In the spring leg model without dampers or force actuators, these two conditions are the same. In this model, these two conditions are usually not at the same time, unless the force actuator just compensates the damper. If the trailing leg leaves the ground not at initial length, some elastic energy is kept in the spring. This will cause some problem in calculating energy cost. After the Trailing Leg Take-off, the trailing leg changes into new swing leg, and the leading leg becomes the new stance leg. At this instant, the state of the model has a change:

$$[q_1(t), \dot{q}_1(t), l_1(t), \dot{l}_1(t)]_{t=t_{TO}^+} = [q_2(t), \dot{q}_2(t), l_2(t), \dot{l}_2(t)]_{t=t_{TO}} \quad (8)$$

where t_{TO} is the time of the Trailing Leg Take-off. Then the model changes into the second single support phase, and numerical integration continues until the stance leg reaches back to the upright configuration. During these integrations, the vertical GRF of stance leg is monitored to ensure that the leg is contacted with the ground. Once the vertical GRF becomes negative, which means the initial condition is invalid and breaks the walking constrain, we will reject it and change another set of initial conditions. By the integration of nonlinear equations, the state of the $(n+1)$ th step is acquired from the n th step with certain model parameters (Fig. 2), and therefore a three-dimensional Poincaré return map is described as:

$$\mathbf{y}_{n+1} = \mathbf{P}(\mathbf{y}_n, k, \zeta, \alpha, F_a, \tau), \quad (9)$$

where $\mathbf{y}_i = [\dot{q}_1(0), l_1(0), \dot{l}_1(0)]^T$ is the state of i th step.

2.2. Fixed point, stability and energetics

In simulations, the step angle α is set to 0.5 rad (28.6°), and the dimensionless v_{ave} is set to 0.35. The leg dimensionless stiffness k and the damping ratio ζ are according to the two parameters above. We choose $k = 14$ and $\zeta = 0.08$ from early researches (Geyer et al., 2006; Whittington and Thelen, 2009; Kim and Park, 2011).

When the Eq. (9) is at a certain fixed point ($\mathbf{y}_{n+1} = \mathbf{y}_n = \mathbf{y}^*$), the model will exhibits a periodic motion corresponding to \mathbf{y}^* . Given a set of model parameters (k, ζ, α) , hip torque strength τ , and a desired walking speed, a nonlinear solver (matlab, fsolve) is used to find the fixed point of Eq. (9).

The stability of the walk can be characterized by the eigenvalues of the Jacobian matrix based on the Poincaré return map. We define $\lambda_i (i = 1, 2, 3)$ as the norm of the eigenvalues of the Jacobian matrix $\partial \mathbf{P} / \partial \mathbf{y}_n$. If the maximum eigenvalue λ_{max} is less than 1, it means that all three eigenvalues are in the unit circle, and this fixed point is stable. The smaller the λ_{max} , the more stable the fixed point. Thus, we use λ_{max} to evaluate the stability of the cycle walk.

We use the parameter COT (Cost of Transport) to describe the energetics of the walking, which is defined as (energy cost)/(body weight * distance traveled). We do not use the energy of the force and torque actuator as the energy cost of this walking model. Because the spring-mass model is an energy conserved model, the amount of energy dissipation by the damper is very small compared to the mechanical energy transfer between the gravitational and elastic potential energy. Srinivasan and Ruina (2006) have used muscle work as energy cost in a compliant leg model, under the approximation that energy cost is proportional to muscle work (Alexander, 1980). In this model, we treat all components (the springs, dampers and force/torque actuators) as the modeling of muscles. Thus, both positive work and negative work cost energy. The muscle work (C) can be estimated from the mechanical work: $C = \beta_1 W^+ + \beta_2 W^-$, where W^+ and W^- are values of positive and negative mechanical works, and β_1 and β_2 are all constants. Because humans have elastic tendon and soft tissues (Zelik and Kuo, 2010), humans usually cost lower muscle work in negative

mechanical work, which means $\beta_1 > \beta_2$. The mechanical energy of the model comes back to the initial value after a cycle walk. Therefore, values of positive and negative mechanical works should be the same ($W^+ = W^-$), and we get $C = (\beta_1 + \beta_2)W^+$, meaning that the muscle work is proportional to the positive mechanical work. Thus, the COT is defined as:

$$\text{COT} = \frac{W^+}{mgL_s}. \quad (10)$$

W^+ is the positive mechanical work of model, and can be calculated by integrating mechanical powers:

$$W^+ = \int [-\tau\dot{\theta}]^+ dt + \sum_{i=1}^2 \int [F_i(t)\dot{l}_i]^+ dt, \quad (11)$$

where $[\]^+$ is non-zero only for positive values ($[x]^+ = x$ if $x > 0$ and $[x]^+ = 0$ if $x \leq 0$). $F_i(t)$ is the GRF along the leg direction, including the force of spring, damper and actuator. Other parameters ($l_i(t)$, τ and θ) are shown in Fig. 3.

3. Results

To find effects of the hip torque, we first find the fixed point under different torque strength. Then, the maximum eigenvalue and the COT of each fixed point are calculated to evaluate the energy cost and stability. Simulation results are shown in Fig. 4. The X-axis τ is the strength of the hip torque actuator. $\tau = 0$ means that the model only uses the leg force actuator.

Simulation results show that the hip torque actuator can improve the walking stability and lower the energy cost. Fig. 4(A) shows the corresponding maximum eigenvalue under different torque strengths. The shaded region represents the stable region where the maximum eigenvalue is less than 1. With leg force only ($\tau = 0$), λ_{\max} is larger than 1, which means walking is unstable. With the torque strength increasing, the λ_{\max} decreases and enters the stable region eventually. As Fig. 4(B) shows, COT decreases with the torque strengthening. From $\tau = 0$ (the leg force actuator only), to $\tau = 0.6$, COT decreases by 33.89%.

From simulation results above, we can find that the hip torque actuator has beneficial effects on the energetics and stability of walking. To understand effects of the hip torque on the energetics,

energy cost of the leg force (W_{Leg}) and the hip torque (w_τ) are calculated, as Fig. 5(A) shows. When $\tau = 0$, all energy cost comes from the leg force. Although the energy dissipation on the hip torque increases with the torque strength increasing, the cost of the leg force decreases furtherly. The decrease of W_{Leg} , ΔW_{Leg} , is larger than the W_τ , which results in the decrease of the total mechanical cost. Fig. 5(B) shows values of $\Delta W_{\text{Leg}}/W_\tau$ under different hip torque strength. Under all torque strengths, values of $\Delta W_{\text{Leg}}/W_\tau$ are larger than 4.16, and the average is 4.31. The hip torque actuator has high efficiency on energetics, and can reduce the total mechanical work with small extra energy cost. The energy cost reduced by the hip torque is about four times that of the energy dissipated on it.

From the simulations, we find that the hip torque can reduce the energy cost and increase the stability of walking. The energy cost is calculated by the muscle work of the model. The specific analysis shows that the energy cost reduced by the hip torque is about four times that of the energy dissipated on it. The stability is quantified by the maximum eigenvalue λ_{\max} , which describes the stability of the fixed point of the Poincaré return map. That means the model can be stabilized at a certain limit cycle state more easily with the hip torque actuator.

4. Discussion

This paper demonstrates that the hip torque actuator during the step-to-step transition can improve the energetics and stability of walking. Effects of hip torque on mechanical works are analyzed, which shows that the hip torque actuator can reduce works of both legs with little extra energy cost.

To further understand the principle of hip torque, the CoM dynamics is analyzed. The hip torque can be equivalent to a force (f_τ) on the CoM, with its direction near to the vertically up direction, as shown in Fig. 6(A). Because of the torque on each leg, a set of forces perpendicular to two legs ($f_{\tau R} = \tau/l_1$ and $f_{\tau L} = \tau/l_2$) are applied on the CoM. Because $l_1 \approx l_2$ during the double support phase, we can get $f_{\tau R} \approx f_{\tau L}$. Thus, the direction of f_τ , which is the resultant force of $f_{\tau R}$ and $f_{\tau L}$, is near to the angular bisector of the two legs, and is near to the vertically up direction (Fig. 6(A)). We can estimate the amplitude of this hip torque force: $f_\tau \approx 2\tau \sin \alpha/l$.

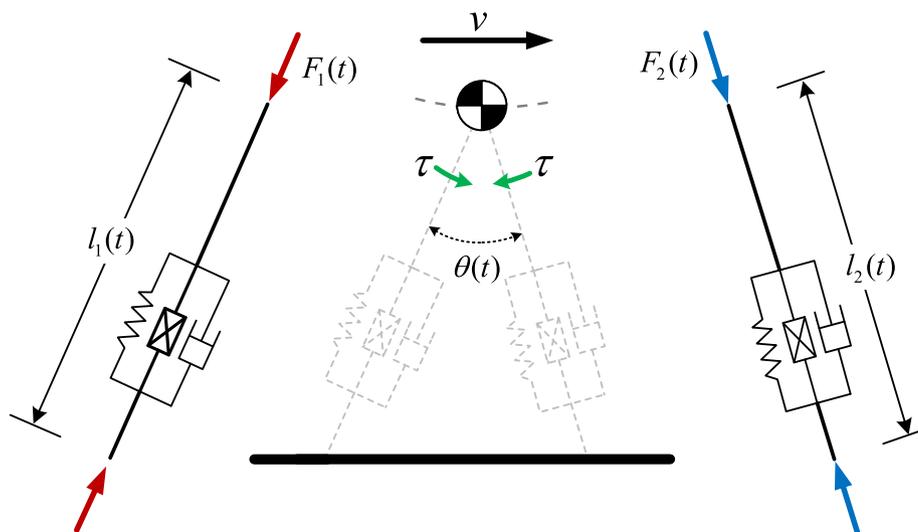


Fig. 3. Diagram of the COT calculation. The energy cost of walking is estimated by the positive mechanical work of the model. All components (the springs, dampers and force/torque actuators) are treated as the modeling of muscles. The energy cost can be calculated by integrating mechanical powers: $W^+ = W_{\text{Leg}} + W_\tau$, where $W_{\text{Leg}} = \sum_{i=1}^2 \int [F_i(t)\dot{l}_i]^+ dt$, and $W_\tau = \int [-\tau\dot{\theta}]^+ dt$. $F_i(t)$ is the GRF along the leg direction, including the force of spring, damper and actuator.

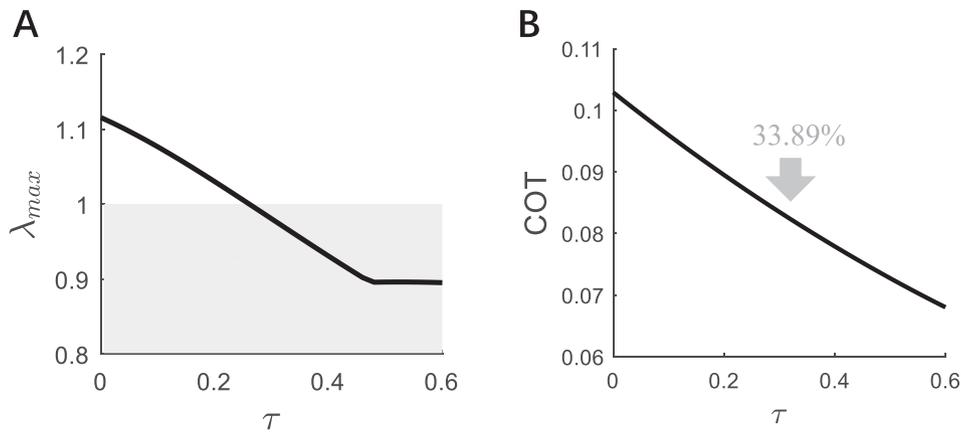


Fig. 4. Simulation results of compliant walking model. (A) The maximum eigenvalue under different hip torques. The shaded region represents the stable region where the maximum eigenvalue is less than 1. With leg force only ($\tau = 0$), λ_{max} is larger than 1, which means the walking is unstable. With the torque increasing, the λ_{max} decreases and enters the stable region. (B) The corresponding COT under different hip torques. The COT decreases with the torque strengthening (from $\tau = 0$ to $\tau = 0.6$, COT_a decrease about 33.89%), which means the hip torque can reduce the energy cost of walking.

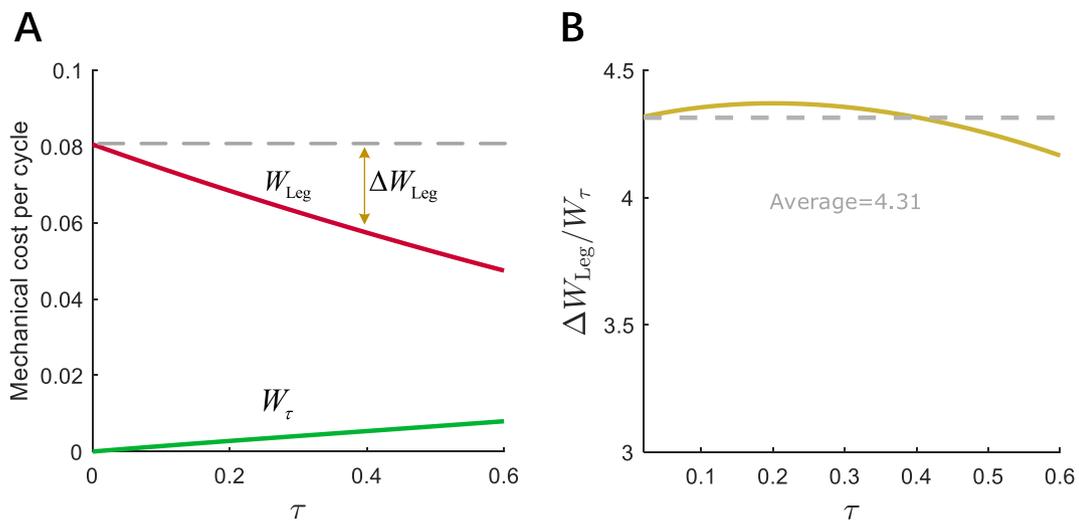


Fig. 5. Simulation results of mechanical cost through a cycle walk. (A) The corresponding mechanical cost of different hip torque strengths. Although the mechanical cost of the hip torque (W_τ) increases with the hip torque strengthening, the cost of legs (W_{Leg}) decrease furtherly. The decrease of W_{Leg} , ΔW_{Leg} , is larger than the W_τ , which results in the decrease of total mechanical cost. (B) Corresponding values of $\Delta W_{Leg}/W_\tau$ under different hip torque strengths. The average value of $\Delta W_{Leg}/W_\tau$ is 4.31, indicating that the hip torque has high efficiency on energetics. The energy cost reduced by the hip torque is about four times that of the energy dissipated on it.

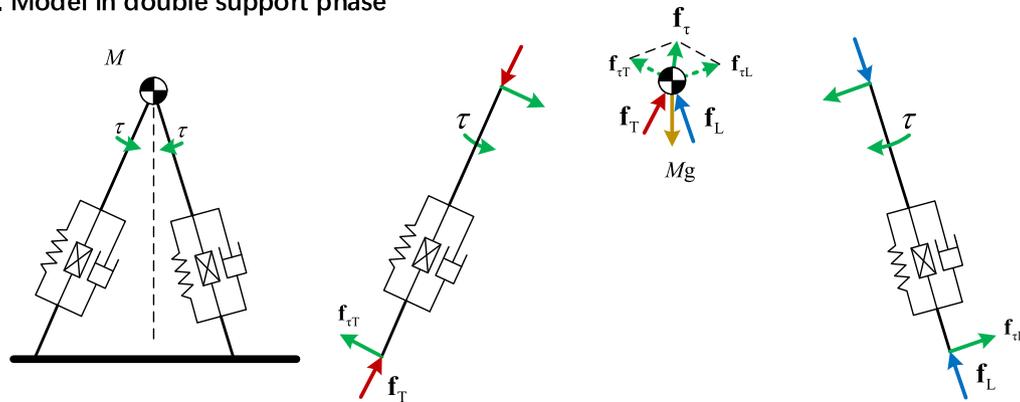
By applying the hip torque actuator, forces of both legs can be reduced resulting in the decrease of energy cost. For the walking without the hip torque, both legs come into contact with the ground and exert leg forces, which direct along each leg. These two leg forces, as well as the gravitational force, propel the redirection of the CoM together (Fig. 6(B)). For the walking with the hip torque actuator, the torque can be equivalent as a force directing upward, which is near to the direction of $\Delta \mathbf{v}$ ($\Delta \mathbf{v} = \mathbf{v}^+ - \mathbf{v}^-$). Thus, the hip torque force is more efficient on the redirection of CoM, and can substitute some forces of both legs (Fig. 6(C)), which leads to the decrease of total energy cost. Although the hip torque cost extra energy, the needed energy is much smaller than the decrease of leg force work.

Although the hip torque has beneficial effects both on the energetics and stability, its effects are limited by the torque strength and the step angle. First, it is difficult for the muscle-tendon system to generate large torques for the human. The value of the hip torque normal walking is approximately 0.15 in dimensionless terms from human walking experiments (Hof, 2000). Second, the step

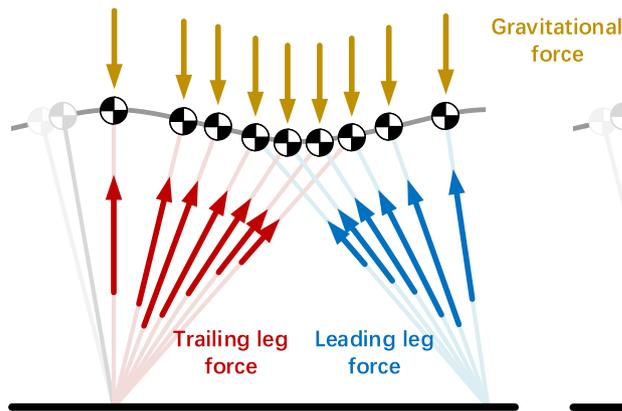
angle α in normal walking is small. For example, the step angle is 0.22–0.63 rad reported from human walking experiments (Donelan et al., 2002). These two factors limit the amplitude of the torque force ($f_\tau \approx 2\tau \sin \alpha/l$), and then limit effects of the hip torque.

The model in this study has springs, dampers, and two types of actuators. The spring leg model can reproduce the dynamics of human walking by changing the leg stiffness and step angle (Geyer et al., 2006). Dampers are added to restrain the excessive motion of the CoM (Kim and Park, 2011), which can restrain the excessive oscillation of model and eliminate multiple solutions of the cycle walk. Shen and Seipel (2012) argue that the damper has fundamental effects on the spring leg model, and can enlarge the basin of attraction, which makes it easier to seek the fixed point. Kim and Park (2011) prove that the damping ratio is small for human walking, which means the main dynamic behavior is determined by the spring. Because of the energy dissipation of dampers, two types of actuators are added to compensate the energy loss. Both the force and torque actuators are in constant

A. Model in double support phase



B. Walking with leg force only



C. Walking with hip torque

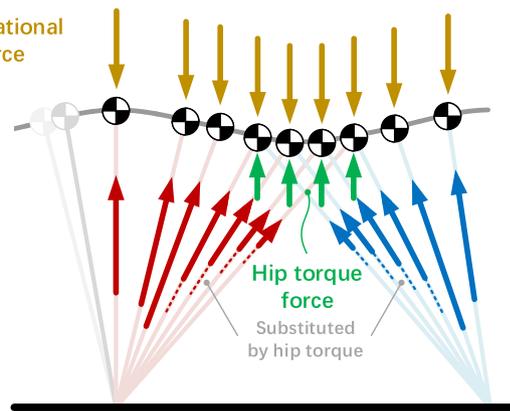


Fig. 6. Diagrams of hip torque and its effects on CoM dynamics. (A) Force diagram of model in step-to-step transition. The hip torque can be treated as a force approximately pointing upwards on the CoM. Thus, four forces act on the CoM: f_T (trailing leg force), f_L (leading leg force), Mg (gravitational force), and f_τ (hip torque force). (B) shows the force diagram without hip torque. The CoM is affected by f_T , f_L and Mg , which redirect the CoM velocity. (C) shows the force diagram with the hip torque. With the hip torque, the forces of both legs can be less than in (B) during the step-to-step transition, which will cost less redirection work.

pattern without any feedback control, which is easy to model mathematically and provides a fundamental understanding.

Because of the hip torque, the GRF during the double support phase is not along the leg direction, and GRFs of two legs intersect on a point above the CoM. This is similar to the virtual pivot point (VPP) in human walking (Maus et al., 2010), which is sufficient for achieving and maintaining stability in single support phase. Müller et al. (2017) show that the forces intersect in a point above the CoM but below the CoM of the upper body, meaning that the force direction pattern promotes the whole body stability, but not upper body stability. In our model, the average position of the intersect point is determined by the strength of the hip torque: the average position becomes higher with the hip torque strengthening. According to our simulations, the corresponding stability is also improved when the hip torque is strengthened. This may indicate that human also uses VPP to maintain walking stability not only in single support phase, but also in double support phase.

This paper demonstrates that the hip torque actuator during step-to-step transition can improve the energetics and stability of walking. These findings may be helpful for the understanding of human walking and the improvement of the bipedal robots, prosthesis and exoskeleton. For example, prosthesis users can use hip-actuated exoskeleton (e.g., Asbeck et al. (2015)) to enhance the hip torque of walking, which can reduce the required force of the prosthesis. Such an application can also be used in walking with load. The assistance at the hip joint can help the redirection of CoM velocity during the step-to-step transition, which can improve the load capability (Wu et al., 2016). In this study, all

simulations are under only one set of walking parameters (the walking speed and step angle), and ignore the influence of feet. The model parameters, the leg stiffness k and the damping ratio ζ , correspond with them (Kim and Park, 2011). If we change the walking speed or the step angle, the model parameters must be changed accordingly. Then, it is unclear whether the results are from the hip torque strength or the change of the model parameters, because these two parameters will determine the dynamics of compliant leg model (Geyer et al., 2006; Shen and Seipel, 2012). The model also ignores the influence of feet for simplification. In the preliminary analysis, we can indicate that effects of hip torque will be reduced, if feet are considered. The equivalent step angle will decrease with feet (Adamczyk et al., 2006), and the hip torque force will be reduced because of the small step angle. It should be noted that the hip torque can also reduce the peaks of leg forces, similar to the roller feet (Whittington and Thelen, 2009). For further study, more experiments and analysis are needed to fully verify the change of the model parameters and the influence of feet.

Conflict of interest

Authors have no conflict of interest to report in this research.

Acknowledgement

This work was supported in part by National Key R&D Program of China under Grant 2018YFB1305400, National Natural Science

Foundation of China under Grant 61533004 and Grant U1613206, Guangdong Innovative and Entrepreneurial Research Team Program under Grant 2016ZT06G587, and Centers for Mechanical Engineering Research and Education at MIT and SUSTech.

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